Psychophysics, Fitting, and Signal Processing for Combined Hearing Aid and Cochlear Implant Stimulation

Tom Francart[1,2] and Hugh J. McDermott[2,3]

The addition of acoustic stimulation 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 who 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.

(Ear & Hearing 2013;34;685–700)

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 (HA) 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 HA. If the residual hearing is in the nonimplanted 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 present 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 to 2 kHz (von Ilberg et al. 2011). In nonimplanted ears there is much variability, but typical cutoff frequencies are around 1 to 2 kHz, when the cutoff frequency is defined as the frequency above which unaided thresholds are higher than 90 to 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 nonimplanted ear, there is binaural stimulation. Having two ears gives rise to several effects that can benefit speech recognition in noise (Bronkhorst 2000; Akeroyd 2006). These include binaural redundancy, head shadow, and squelch. Of these, squelch* is widely considered to be the only one based on true binaural cues, that is, 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 localize 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 from the acoustic stimulus, which is not available through the CI (such as pitch) and vice versa to achieve better performance. In addition, 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 & 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 (Kong et al. 2005; Dorman et al. 2008). 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,

[1]KU Leuven Department of Neurosciences, ExpORL, Belgium; [2]The Bionics Institute, Melbourne, Victoria, Australia; and [3]Department of Otolaryngology and Medical Bionics, The University of Melbourne, Victoria, Australia.

* Binaural 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.

0196/0202/13/3406-0685/0 • Ear and Hearing • Copyright © 2013 by Lippincott Williams & Wilkins • Printed in the U.S.A.

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). In general, 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 with 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.

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 HA. 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 Hybrid™ (Büchner et al. 2009) and the Med-El DUET system (Helbig et al. 2008; Helbig & Baumann 2010). While the current integrated hybrid systems solve some of the problems associated with separate devices discussed later in the article, 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 by current sound processing strategies, and finally discuss transmission of binaural cues through bimodal devices.

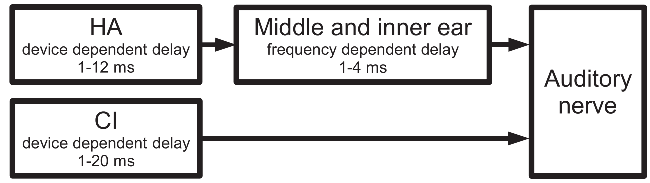

Fig. 1. Illustration of the temporal synchronization problem. HA indicates hearing aid. Adapted with permission from Francart (2008).

Temporal Synchronization, Place Mismatch, and Preprocessing

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 to 20 msec†. 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 an HA, it also undergoes a processing delay (Stone & Moore 1999), 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 msec is probably not a problem for speech perception (Stone & 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 μsec in one ear (Javer & 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 μsec.

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.

† Processing 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.

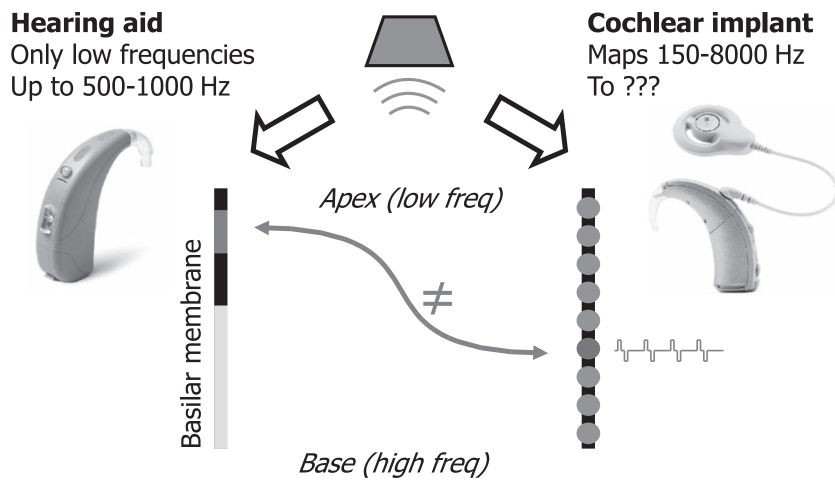

Fig. 2. Illustration of place mismatch. The same sound, for example, a 1 kHz tone, processed by a cochlear implant and hearing aid can end up causing neural excitation in different places in the two cochleae. Adapted with permission from Francart (2008).

Consider, for instance, a 200 Hz pure tone. When Cochlear's ACE strategy is used, 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 cochlear place of the most apical electrode varies widely across subjects. Ketten et al. (1998) found corresponding frequencies ranging from 387 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 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 Frequency Allocation), but large mismatches can be a problem for binaural cue perception (see Binaural Cue Perception).

CI sound processors and especially HAs nowadays have a large range of preprocessing options, such as compression of the acoustic signal or automatic gain control (Moore 2008), noise reduction (Bentler & Chiou 2006), and feedback suppression (Spriet et al. 2010). If there are large across-ear differences in preprocessing, 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 lateralization 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 Binaural Cue Transmission Through Bimodal Devices), 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.

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 on the one hand envelopes, with rates up to a few hundred hertz, and on the other hand temporal fine structure with rates 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 (NH) listeners are sensitive to ITDs with thresholds ranging from 10 to 100 μsec, depending on the type of signal (see Binaural Cue Transmission Through Bimodal Devices). 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. For example, most listeners cannot perceive temporal modulations beyond 300 to 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 cutoff 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-msec 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 nonlinear maxima selection process introduces extratemporal distortion. For a carrier rate of 900 pps, the timing information in the envelope is quantized to a multiple of 1.1 msec (the period corresponding to the carrier rate), which is undesirable, given a desired ITD resolution on the order of 10 to 100 μsec.

Advanced Bionics devices commonly use the HiResolution strategy, which is similar to CIS but uses high carrier rates (around 2900 pps). HiResolution uses half-wave rectification for envelope detection, but with a fairly high cutoff 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 μsec 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 & 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 to 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 NH, 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. In addition, in the CIS channels, the same envelope temporal cue quantization effect is present as discussed earlier for ACE.

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 an 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 Binaural Cue Perception). 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.

Interaural Time Differences. • NH listeners can use ongoing temporal cues that are present both in the fine structure and the envelope of sound signals (Henning 1974; Yost 1974; Bernstein & Trahiotis 1985). Most current sound processors do not transmit undistorted temporal fine structure (see previous section), and even with specially crafted stimuli, CI-listeners usually cannot 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 nonlinear processes 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 msec. Given that realistic ITD cues range from 0 to around 700 μsec, this can result in relatively large temporal distortions.

Interaural Level Differences • In addition, good ITD perception depends on consistent ILD cues or at least loudness balance between the ears (Domnitz & Colburn 1977; Wightman & 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 multichannel compression HA 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. In addition, 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 (Hz). Unfortunately, many bimodal listeners only have usable residual hearing up to 1000 to 2000 Hz (Dorman & 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 nonmonotonic beyond approximately 45° off-center (Shaw 1974; Francart et al. 2011b), which makes it impossible to distinguish between sound directions beyond 45° at one side based on ILD alone.

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 NH listeners, the utility of spectral cues for localization in the horizontal plane is very limited (Wightman & Kistler 1997). In addition, 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 localization 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 Binaural Cue Perception), 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 of 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, that is, relatively flat for angles larger than approximately 45° (Francart et al. 2011b); and
6. it has been shown that many bilateral CI users do not use ITD cues with commercial devices (Seeber & Fastl 2008; Aronoff et al. 2010). 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 11 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.

PSYCHOPHYSICAL STUDIES

While many authors have studied performance with EAS with commercial devices, the number of fundamental studies that use 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.

Matching Pitch and Cochlear Place of Excitation

As mentioned in Current Devices, 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 & Hafter 1981) and ILD discrimination (Francart & Wouters 2007). In EAS users, place mismatch varies, and is difficult to characterize. Attempts to determine degree of mismatch have included behavioral 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 2 octaves) than would be predicted for those electrodes based on the Greenwood cochlear place-to-frequency map (Blamey et al. 1996; Boëx et al. 2006; Dorman et al. 2007). 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 NH (or near-NH) in the nonimplanted 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, for example, 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 nonimplanted ear, and found that pitch matches obtained for either side were similar. They proposed the use of pitch matching to obtain the crossover 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 crossover frequency to the pitch of the most apical electrode is the best approach (see Frequency Allocation).

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. 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 toward better ITD sensitivity for certain acoustic frequency ranges when combined with a certain electrode. This tendency shifted consistently with electrode number; that is, 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 is extremely time consuming.

Smith and Delgutte (2007) measured the binaural interaction component of the electrical auditory brainstem response (ABR; i.e., the difference between the electrical brainstem 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 computed tomography 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 computed tomography 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. In addition, it is unclear whether any matching between acoustic hearing and the electric filter bank is desirable for optimal speech perception (see Frequency Allocation). 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.

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 NH 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; Zeng and Shannon (1992) performed binaural loudness-balancing experiments between electric and acoustic stimulation in auditory brainstem 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 microamperes. Eddington et al. (1978) also found a linear decibel-versus-microampere relationship in 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 nonimplanted ear. Francart et al. (2008) confirmed this finding in 10 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 & 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 found in the study by 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 1 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 LGFs and loudness summation for a group of bimodal listeners. The stimuli were 1/3 octave-wide noisebands, 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 to 500 Hz noiseband 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, LGFs 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 Signal Processing Techniques).

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 Current Devices.

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 & Wouters 2007). NH listeners are also very sensitive to ITDs in several different features of the acoustic signal: the onset, temporal fine structure (Feddersen et al. 1957; Yost 1974; Brown & Yost 2011), and envelope modulations (Henning 1974; Middlebrooks & Green 1990), with thresholds ranging from as low as 10 μsec for pure tones, to around 40 μsec for vowels (Akeroyd 2003), and around 100 μsec for modulated high-frequency tones (Bernstein & 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 to 250 μsec (compared with 10–50 μsec 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 to 200 Hz, which corresponds to the rate limitation for envelope-ITD perception found in NH listeners (Bernstein & 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 Temporal Synchronization, Place Mismatch, and Preprocessing 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 msec. 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 & Ruggero 2001). Francart et al. (2009a) compared this value with the difference between average ABR and electrical ABR 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 Psychosocial Studies we review a number of speech processing algorithms designed to overcome this deficiency.

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, 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 earlier discussion about loudness growth and transmission of ILD cues 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. Büchner 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 cutoff frequency, but significant improvements were found over CI-only stimulation, even for the 500 Hz cutoff frequency. Zhang et al. (2010) tested monosyllabic word recognition in quiet and sentence recognition in noise. The acoustic stimuli presented to the nonimplanted ear were either low-pass filtered at 125, 250, 500, or 750 Hz, or unfiltered (wideband). 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/octave and therefore did not completely eliminate all information above its cutoff frequency. Cullington and Zeng (2010) tested a CI recipient with NH in the contralateral ear, and measured the effect of low-pass and high-pass filtering of 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 downward 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 an 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 3 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 are 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.

FITTING METHODS

When attempting to fit an 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 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.

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 intersubject 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 Matching Pitch and Cochlear Place of Excitation). In addition, depending on insertion depth, frequencies up to 1 to 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 cutoff frequency of the residual hearing are mapped to electrodes. In panel 4 of Figure 3, only acoustic frequencies beyond the cutoff 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 (Med-El, Inssbruck, Austria) and the Cochlear Hybrid processor (Cochlear, Sydney, Australia). 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 to 2 kHz) in a conventional way and do not amplify at higher frequencies. The upper cutoff frequency of acoustic stimulation is based on the audiogram of the residual hearing.

The cutoff 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., Kiefer et al. 2005; James et al. 2006; 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 cutoff 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 to 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 behavioral 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 nonoverlapping-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.

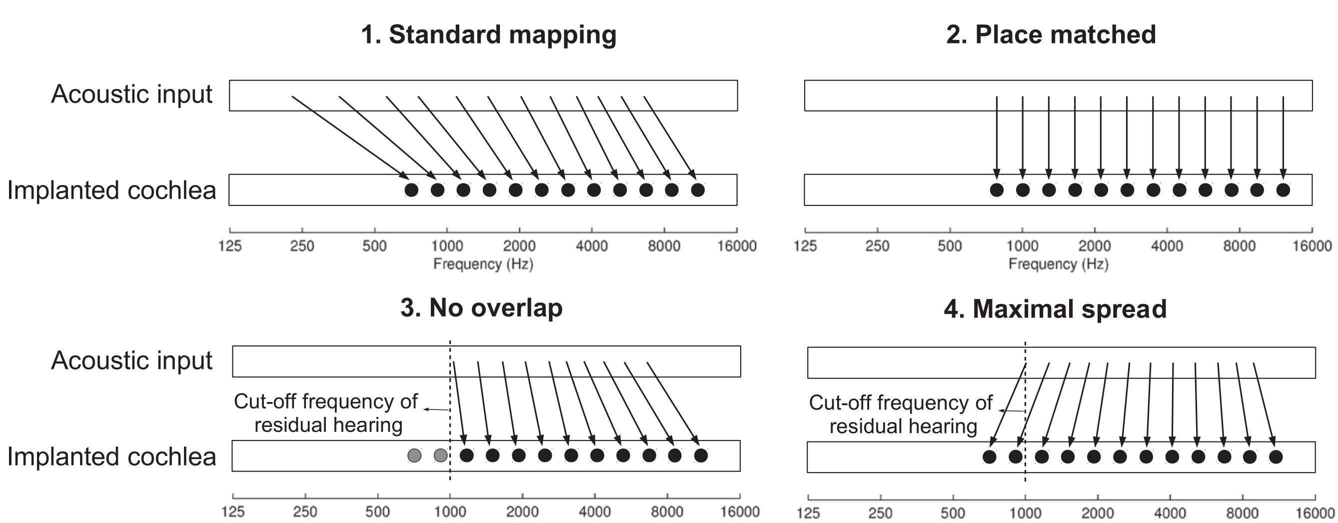

Fig. 3. Conceptual diagram of different frequency-to-electrode allocations in the cochlear implant sound processor. Note that actual electrode positions in the cochlea are different for each recipient, and standard frequency-to-electrode allocations depend on the processor used.

Speech perception results were only reported for the preferred map. Using a very similar protocol, Fraysse et al. (2006) found that seven of nine subjects much preferred the nonoverlapping-shifted, while two 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 cutoff frequency of the audiogram. The cutoff 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 to 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 to 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 nonimplanted ear (3 subjects) or both ears (2 subjects). In the CI-only condition, however, performance was lower for the nonoverlapping 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 & Shannon 1999; Rosen et al. 1999; Fu et al. 2002). 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, for example, by disabling apical electrodes or only mapping frequencies beyond 1 to 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 reimplantation 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.

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 the first step, the HA's frequency response is fine tuned based on the user's sound-quality judgments in a paired-comparisons procedure among 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 nonlinear HAs this is done for speech at 55 and 80 dB SPL. It has been reported that with this type of fitting speech perception and localization tests 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 cutoff frequency until the cutoff frequency was 1.5 to 2 times the estimated lower edge frequency of the dead region. With further increases in cutoff frequency, speech reception either remained constant or declined. On average it seems best not to amplify beyond 1.7 times the lower cutoff 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 cutoff 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 among 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 nonlinear processing (e.g., maxima selection) in most current sound processors additionally complicates such fitting procedures.

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.

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 (James et al. 2002; Blamey 2005). ADRO keeps track of the statistics of the intensity at the *output* of the HA in each frequency band and slowly adjusts the corresponding gain to keep the intensity distribution at the output within a certain range, for example, such that the 30th and 90th percentiles correspond to soft and comfortable levels, respectively. 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 an 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 nonimplanted 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 that 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.

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 downward and mixed with any lower-frequency signal that may be present at the same time. In other HAs, nonlinear frequency compression is applied (Simpson et al. 2005; Glista et al. 2009). 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 (Phonak, Zurich, Switzerland) (nonlinear 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 (Widex, Lynge, Denmark) using 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.

Experimental Signal Processing Schemes

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

On the basis of the works 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 & Glasberg 1997). A block diagram is shown in Figure 4. SCORE uses an NH loudness model to estimate how an 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, 2012b) 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 & 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 hearing is usually limited to low frequencies, but the head-shadow effect and therefore also the ILD are small at low frequencies; (2) LGFs are generally different for acoustic and electric stimulation; and (3) the ILD-versus-angle function is nonmonotonic from approximately 45° to 60° off-center (Shaw 1974; Francart et al. 2011b) 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 issue 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° to

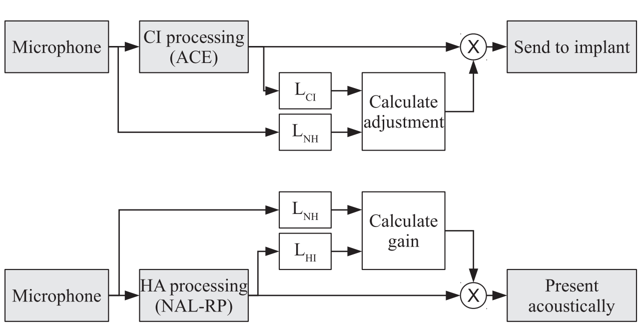

Fig. 4. Overview of SCORE signal processing strategy, reproduced from Francart and McDermott (2012b). The gray blocks indicate existing signal processing. ACE indicates advanced combination encoder; CI, cochlear implant; HA, hearing aid; NAL-RP, The National Acoustics Laboratories fitting rule, Revised for Profound hearing loss.

10° absolute error, relative to a mean absolute error of 28° in the condition without ILD enhancement.

Bimodal listeners with sufficient residual hearing are sensitive to ITD cues in laboratory experiments (see Binaural Cue Perception). While several strategies have been developed to improve ITD perception with bilateral CIs (e.g., van Hoesel & Tyler 2003; Smith 2009), 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. In addition, the across-channel timing of the electric signal becomes a problem. In acoustic hearing, the outputs of auditory filters with decreasing center 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 × 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. (2013) proposed the modulation enhancement strategy, 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 with 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.

SUMMARY AND FUTURE DIRECTIONS

Mainly due to progressive relaxation of implantation criteria there is now a large population of CI users with residual hearing in the implanted or nonimplanted 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 yet clear 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, no convincing evidence of benefit has been obtained, and there are some procedural issues. So it does not seem appropriate to use these procedures in clinical practice.

NH listeners have ILD and ITD thresholds of around 1 dB and 40 μsec and the physical cues in realistic signals range up to 20 dB and 700 μsec. In ideal circumstances, bimodal listeners with sufficient residual hearing can be sensitive to ILDs and ITDs with thresholds of around 1 dB and 150 μsec. 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 ITDs 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. In addition, one should not amplify signal frequencies beyond 1.7 times the estimated lower cutoff 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 EAS.

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 these strategies can be implemented in commercial systems.

In summary, we are of the opinion the following steps should be undertaken to improve hearing with combined HAs and CIs, in particular binaural cue perception, leading to improved sound source localization 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.

ACKNOWLEDGMENTS

The Bionics Institute acknowledges the support it receives from the Victorian Government through its Operational Infrastructure Support Program.

Tom Francart was sponsored by a Post Doctoral Fellowship of the Fund for Scientific Research of the Flemish Government and a Marie Curie International Outgoing Fellowship of the European Commission, grant agreement number PIOF-GA-2009–252730.

The authors declare no conflict of interest.

Address for correspondence: Tom Francart, ExpORL, Department of Neurosciences, KU Leuven, Herestraat 49 bus 721, B-3000 Leuven, Belgium. E-mail: tom.francart@med.kuleuven.be

Received January 3, 2013; accepted May 9, 2013.

REFERENCES

Adunka, O. F., Buss, E., Clark, M. S., et al. (2008). Effect of preoperative residual hearing on speech perception after cochlear implantation. *Laryngoscope*, *118*, 2044–2049.

Akeroyd, M. A. (2003). Threshold differences for interaural time delays carried by double vowels. *J Acoust Soc Am*, *114*(4 Pt 1), 2167–2177.

Akeroyd, M. (2006). The psychoacoustics of binaural hearing. *Int J Audiol*, *45* Suppl 1, 25–33.

Aronoff, J. M., Yoon, Y. S., Freed, D. J., et al. (2010). The use of interaural time and level difference cues by bilateral cochlear implant users. *J Acoust Soc Am*, *127*, EL87–EL92.

Baumann, U., Rader, T., Helbig, S., et al. (2011). Pitch matching psychometrics in electric acoustic stimulation. *Ear Hear*, *32*, 656–662.

Bentler, R., & Chiou, L. K. (2006). Digital noise reduction: An overview. *Trends Amplif*, *10*, 67–82.

Bernstein, L. R., & Trahiotis, C. (1985). Lateralization of low-frequency, complex waveforms: The use of envelope-based temporal disparities. *J Acoust Soc Am*, *77*, 1868–1880.

Bernstein, L. R., & Trahiotis, C. (2002). Enhancing sensitivity to interaural delays at high frequencies by using "transposed stimuli". *J Acoust Soc Am*, *112*(3 Pt 1), 1026–1036.

Blamey, P. J. (2005). Adaptive dynamic range optimization (ADRO): A digital amplification strategy for hearing aids and cochlear implants. *Trends Amplif*, *9*, 77–98.

Blamey, P. J., Dooley, G. J., James, C. J., et al. (2000). Monaural and binaural loudness measures in cochlear implant users with contralateral residual hearing. *Ear Hear*, *21*, 6–17.

Blamey, P. J., Dooley, G. J., Parisi, E. S., et al. (1996). Pitch comparisons of acoustically and electrically evoked auditory sensations. *Hear Res*, *99*, 139–150.

Blauert, J. (1997). *Spatial Hearing*. Cambridge, MA: MIT Press.

Boëx, C., Baud, L., Cosendai, G., et al. (2006). Acoustic to electric pitch comparisons in cochlear implant subjects with residual hearing. *J Assoc Res Otolaryngol*, *7*, 110–124.

Bonham, B. H., & Litvak, L. M. (2008). Current focusing and steering: Modeling, physiology, and psychophysics. *Hear Res*, *242*, 141–153.

Bronkhorst, A. (2000). The cocktail party phenomenon: A review of research on speech intelligibility in multiple-talker conditions. *Acta Acustica Unit Acustica*, *86*(1), 117–128.

Brown, C. A., & Bacon, S. P. (2009). Low-frequency speech cues and simulated electric-acoustic hearing. *J Acoust Soc Am*, *125*, 1658–1665.

Brown, C. A., & Bacon, S. P. (2010). Fundamental frequency and speech intelligibility in background noise. *Hear Res*, *266*, 52–59.

Brown, C. A., Scherrer, N. M., Bacon, S. P. (2010). Shifting fundamental frequency in simulated electric-acoustic listening. *J Acoust Soc Am*, *128*, 1272–1279.

Brown, C. A., & Yost, W. A. (2011). Interaural spectral asymmetry and sensitivity to interaural time differences. *J Acoust Soc Am*, *130*, EL358–EL364.

Buechner, A., Brendel, M., Lesinski-Schiedat, A., et al. (2010). Cochlear implantation in unilateral deaf subjects associated with ipsilateral tinnitus. *Otol Neurotol*, *31*, 1381–1385.

Büchner, A., Schüssler, M., Battmer, R. D., et al. (2009). Impact of low-frequency hearing. *Audiol Neurootol*, *14* Suppl 1, 8–13.

Byrne, D., Dillon, H., Ching, T., et al. (2001). NAL-NL1 procedure for fitting nonlinear hearing aids: Characteristics and comparisons with other procedures. *J Am Acad Audiol*, *12*, 37–51.

Byrne, D., Parkinson, A., Newall, P. (1990). Hearing aid gain and frequency response requirements for the severely/profoundly hearing impaired. *Ear Hear*, *11*, 40–49.

Carlyon, R. P., Macherey, O., Frijns, J. H., et al. (2010). Pitch comparisons between electrical stimulation of a cochlear implant and acoustic stimuli presented to a normal-hearing contralateral ear. *J Assoc Res Otolaryngol*, *11*, 625–640.

Ching, T. Y., Incerti, P., Hill, M. (2004). Binaural benefits for adults who use hearing aids and cochlear implants in opposite ears. *Ear Hear*, *25*, 9–21.

Ching, T. Y., Psarros, C., Hill, M., et al. (2001). Should children who use cochlear implants wear hearing aids in the opposite ear? *Ear Hear*, *22*, 365–380.

Ching, T., Scollie, S., Dillon, H., et al. (2010). A cross-over, double-blind comparison of the nal-nl1 and the dsl v4.1 prescriptions for children with mild to moderately severe hearing loss. *Int J Audiol*, *49* Suppl 1, 4–15.

Ching, T. Y., van Wanrooy, E., Dillon, H. (2007). Binaural-bimodal fitting or bilateral implantation for managing severe to profound deafness: A review. *Trends Amplif*, *11*, 161–192.

Cohen, L. T., Xu, J., Xu, S. A., et al. (1996). Improved and simplified methods for specifying positions of the electrode bands of a cochlear implant array. *Am J Otol*, *17*, 859–865.

Cullington, H. E., & Zeng, F. G. (2010). Bimodal hearing benefit for speech recognition with competing voice in cochlear implant subject with normal hearing in contralateral ear. *Ear Hear*, *31*, 70–73.

Domnitz, R. H., & Colburn, H. S. (1977). Lateral position and interaural discrimination. *J Acoust Soc Am*, *61*, 1586–1598.

Dorman, M. F., & Gifford, R. H. (2010). Combining acoustic and electric stimulation in the service of speech recognition. *Int J Audiol*, *49*, 912–919.

Dorman, M. F., Gifford, R. H., Spahr, A. J., et al. (2008). The benefits of combining acoustic and electric stimulation for the recognition of speech, voice and melodies. *Audiol Neurootol*, *13*, 105–112.

Dorman, M. F., Smith, L., Parkin, J. L. (1993). Loudness balance between acoustic and electric stimulation by a patient with a multichannel cochlear implant. *Ear Hear*, *14*, 290–292.

Dorman, M. F., Spahr, T., Gifford, R., et al. (2007). An electric frequency-to-place map for a cochlear implant patient with hearing in the nonimplanted ear. *J Assoc Res Otolaryngol*, *8*, 234–240.

Dunn, C. C., Perreau, A., Gantz, B., et al. (2010). Benefits of localization and speech perception with multiple noise sources in listeners with a short-electrode cochlear implant. *J Am Acad Audiol*, *21*, 44–51.

Dunn, C. C., Tyler, R. S., Witt, S. A. (2005). Benefit of wearing a hearing aid on the unimplanted ear in adult users of a cochlear implant. *J Speech Lang Hear Res*, *48*, 668–680.

Eddington, D. K., Dobelle, W. H., Brackmann, D. E., et al. (1978). Auditory prostheses research with multiple channel intracochlear stimulation in man. *Ann Otol Rhinol Laryngol*, *87*(6 Pt 2), 1–39.

El Fata, F., James, C. J., Laborde, M. L., et al. (2009). How much residual hearing is 'useful' for music perception with cochlear implants? *Audiol Neurootol*, *14* Suppl 1, 14–21.

Feddersen, W., Sandel, T., Teas, D., et al. (1957). Localization of high-frequency tones. *J Acoust Soc Am*, *29*(9), 988–991.

Francart, T., Brokx, J., Wouters, J. (2008). Sensitivity to interaural level difference and loudness growth with bilateral bimodal stimulation. *Audiol Neurootol*, *13*, 309–319.

Francart, T., Brokx, J., Wouters, J. (2009a). Sensitivity to interaural time differences with combined cochlear implant and acoustic stimulation. *J Assoc Res Otolaryngol*, *10*, 131–141.

Francart, T., Lenssen, A., Wouters, J. (2011a). Sensitivity of bimodal listeners to interaural time differences with modulated single- and multiple-channel stimuli. *Audiol Neurootol*, *16*, 82–92.

Francart, T., Lenssen, A., Wouters, J. (2013). Modulation emphasis improves perception of interaural time differences in vowels with bimodal stimulation. *J Assoc Res Otolaryngol*, submitted.

Francart, T., & McDermott, H. J. (2012a). Development of a loudness normalisation strategy for combined cochlear implant and acoustic stimulation. *Hear Res*, *294*, 114–124.

Francart, T., & McDermott, H. J. (2012b). Speech perception and localisation with SCORE bimodal: A loudness normalisation strategy for combined cochlear implant and hearing aid stimulation. *PLoS One*, *7*, e45385.

Francart, T., Van den Bogaert, T., Moonen, M., et al. (2009b). Amplification of interaural level differences improves sound localization in acoustic simulations of bimodal hearing. *J Acoust Soc Am*, *126*, 3209–3213.

Francart, T., & Wouters, J. (2007). Perception of across-frequency interaural level differences. *J Acoust Soc Am*, *122*, 2826–2831.

Francart, T. (2008). Perception of binaural localization cues with combined electric and acoustic hearing. Ph.D. thesis, K.U. Leuven.

Francart, T., Lenssen, A., Wouters, J. (2011b). Enhancement of interaural level differences improves sound localization in bimodal hearing. *J Acoust Soc Am*, *130*, 2817–2826.

Fraysse, B., Macías, A. R., Sterkers, O., et al. (2006). Residual hearing conservation and electroacoustic stimulation with the nucleus 24 contour advance cochlear implant. *Otol Neurotol*, *27*, 624–633.

Fu, Q. J., & Shannon, R. V. (1999). Effects of electrode configuration and frequency allocation on vowel recognition with the Nucleus-22 cochlear implant. *Ear Hear*, *20*, 332–344.

Fu, Q. J., Shannon, R. V., Galvin, J. J. 3rd. (2002). Perceptual learning following changes in the frequency-to-electrode assignment with the Nucleus-22 cochlear implant. *J Acoust Soc Am*, *112*, 1664–1674.

Gfeller, K., Turner, C., Oleson, J., et al. (2007). Accuracy of cochlear implant recipients on pitch perception, melody recognition, and speech reception in noise. *Ear Hear*, *28*, 412–423.

Gifford, R. H., Dorman, M. F., Shallop, J. K., et al. (2010). Evidence for the expansion of adult cochlear implant candidacy. *Ear Hear*, *31*, 186–194.

Glista, D., Scollie, S., Bagatto, M., et al. (2009). Evaluation of nonlinear frequency compression: Clinical outcomes. *Int J Audiol*, *48*, 632–644.

Grantham, D. W., Ricketts, T. A., Ashmead, D. H., et al. (2008). Localization by postlingually deafened adults fitted with a single cochlear implant. *Laryngoscope*, *118*, 145–151.

Green, T., Faulkner, A., Rosen, S. (2012). Frequency selectivity of contralateral residual acoustic hearing in bimodal cochlear implant users, and limitations on the ability to match the pitch of electric and acoustic stimuli. *Int J Audiol*, *51*, 389–398.

Greenwood, D. D. (1990). A cochlear frequency-position function for several species–29 years later. *J Acoust Soc Am*, *87*, 2592–2605.

He, S., Brown, C. J., Abbas, P. J. (2012). Preliminary results of the relationship between the binaural interaction component of the electrically evoked auditory brainstem response and interaural pitch comparisons in bilateral cochlear implant recipients. *Ear Hear*, *33*, 57–68.

Helbig, S., & Baumann, U. (2010). Acceptance and fitting of the DUET device—A combined speech processor for electric acoustic stimulation. *Adv Otorhinolaryngol*, *67*, 81–87.

Helbig, S., Baumann, U., Helbig, M., et al. (2008). A new combined speech processor for electric and acoustic stimulation—Eight months experience. *ORL J Otorhinolaryngol Relat Spec*, *70*, 359–365.

Henning, G. B. (1974). Detectability of interaural delay in high-frequency complex waveforms. *J Acoust Soc Am*, *55*, 84–90.

Hochmair, I., Nopp, P., Jolly, C., et al. (2006). MED-EL Cochlear implants: State of the art and a glimpse into the future. *Trends Amplif*, *10*, 201–219.

Hua, H., Johansson, B., Jönsson, R., et al. (2012). Cochlear implant combined with a linear frequency transposing hearing aid. *J Am Acad Audiol*, *23*, 722–732.

Iwaki, T., Blamey, P., Kubo, T. (2008). Bimodal studies using adaptive dynamic range optimization (ADRO) technology. *Int J Audiol*, *47*, 311–318.

James, C., Blamey, P., Shallop, J. K., et al. (2001). Contralateral masking in cochlear implant users with residual hearing in the non-implanted ear. *Audiol Neurootol*, *6*, 87–97.

James, C. J., Blamey, P. J., Martin, L., et al. (2002). Adaptive dynamic range optimization for cochlear implants: A preliminary study. *Ear Hear*, *23*(1 Suppl), 49S–58S.

James, C. J., Fraysse, B., Deguine, O., et al. (2006). Combined electroacoustic stimulation in conventional candidates for cochlear implantation. *Audiol Neurootol*, *11* Suppl 1, 57–62.

Javer, A. R., & Schwarz, D. W. (1995). Plasticity in human directional hearing. *J Otolaryngol*, *24*, 111–117.

Keidser, G., Rohrseitz, K., Dillon, H., et al. (2006). The effect of multichannel wide dynamic range compression, noise reduction, and the directional microphone on horizontal localization performance in hearing aid wearers. *Int J Audiol*, *45*, 563–579.

Ketten, D. R., Skinner, M. W., Wang, G., et al. (1998). *In vivo* measures of cochlear length and insertion depth of nucleus cochlear implant electrode arrays. *Ann Otol Rhinol Laryngol Suppl*, *175*, 1–16.

Kiefer, J., Pok, M., Adunka, O., et al. (2005). Combined electric and acoustic stimulation of the auditory system: Results of a clinical study. *Audiol Neurootol*, *10*, 134–144.

Kleine Punte, A., Vermeire, K., Hofkens, A., et al. (2011). Cochlear implantation as a durable tinnitus treatment in single-sided deafness. *Coch Impl Int*, *12* Suppl 1, 26–9.

Kong, Y. Y., Stickney, G. S., Zeng, F. G. (2005). Speech and melody recognition in binaurally combined acoustic and electric hearing. *J Acoust Soc Am*, *117*(3 Pt 1), 1351–1361.

Kuk, F., Keenan, D., Korhonen, P., et al. (2009). Efficacy of linear frequency transposition on consonant identification in quiet and in noise. *J Am Acad Audiol*, *20*, 465–479.

Laneau, J., Wouters, J., Moonen, M. (2006). Improved music perception with explicit pitch coding in cochlear implants. *Audiol Neurootol*, *11*, 38–52.

Lenssen, A., Francart, T., Brokx, J., et al. (2011). Bimodal listeners are not sensitive to interaural time differences in unmodulated low-frequency stimuli (L). *J Acoust Soc Am*, *129*, 3457–3460.

Lin, P., Turner, C. W., Gantz, B. J., et al. (2011). Ipsilateral masking between acoustic and electric stimulations. *J Acoust Soc Am*, *130*, 858–865.

Luntz, M., Shpak, T., Weiss, H. (2005). Binaural-bimodal hearing: Concomitant use of a unilateral cochlear implant and a contralateral hearing aid. *Acta Otolaryngol*, *125*, 863–869.

McDermott, H. (2011). Benefits of combined acoustic and electric hearing for music and pitch perception. *Semin Hear*, *32*, 103–114.

McDermott, H., & Henshall, K. (2010). The use of frequency compression by cochlear implant recipients with postoperative acoustic hearing. *J Am Acad Audiol*, *21*, 380–389.

McDermott, H. J., McKay, C. M., Richardson, L. M., et al. (2003). Application of loudness models to sound processing for cochlear implants. *J Acoust Soc Am*, *114*(4 Pt 1), 2190–2197.

McDermott, H. J., McKay, C. M., Vandali, A. E. (1992). A new portable sound processor for the University of Melbourne/Nucleus Limited multielectrode cochlear implant. *J Acoust Soc Am*, *91*, 3367–3371.

McDermott, H., Sucher, C., Simpson, A. (2009). Electro-acoustic stimulation. Acoustic and electric pitch comparisons. *Audiol Neurootol*, *14* Suppl 1, 2–7.

McDermott, H., & Varsavsky, A. (2009). Better fitting of cochlear implants: Modeling loudness for acoustic and electric stimuli. *J Neural Eng*, *6*, 065007.

McKay, C. M., Henshall, K. R., Farrell, R. J., et al. (2003). A practical method of predicting the loudness of complex electrical stimuli. *J Acoust Soc Am*, *113*(4 Pt 1), 2054–2063.

McKay, C. M., & McDermott, H. J. (1998). Loudness perception with pulsatile electrical stimulation: The effect of interpulse intervals. *J Acoust Soc Am*, *104*(2 Pt 1), 1061–1074.

McKay, C. M., Vandali, A. E., McDermott, H. J., et al. (1994). Speech processing for multichannel cochlear implants: Variations of the Spectral Maxima Sound Processor strategy. *Acta Otolaryngol*, *114*, 52–58.

Middlebrooks, J. C., & Green, D. M. (1990). Directional dependence of interaural envelope delays. *J Acoust Soc Am*, *87*, 2149–2162.

Milczynski, M., Wouters, J., van Wieringen, A. (2009). Improved fundamental frequency coding in cochlear implant signal processing. *J Acoust Soc Am*, *125*, 2260–2271.

Mills, A. (1960). Lateralization of high-frequency tones. *J Acoust Soc Am*, *32*(132), 132–134.

Mok, M., Grayden, D., Dowell, R. C., et al. (2006). Speech perception for adults who use hearing aids in conjunction with cochlear implants in opposite ears. *J Speech Lang Hear Res*, *49*, 338–351.

Moore, B. (2003). *An Introduction to the Psychology of Hearing* (6th ed.). Bingley, UK: Emerald Group Publishing Limited.

Moore, B. C. (2008). The choice of compression speed in hearing AIDS: Theoretical and practical considerations and the role of individual differences. *Trends Amplif*, *12*, 103–112.

Moore, B., Glasberg, B. (1997). A model of loudness perception applied to cochlear hearing loss. *Audit Neurosci*, *3*, 289–311.

Moore, B., Glasberg, B., Baer, T. (1997). A model for the prediction of thresholds, loudness, and partial loudness. *J Audio Engineer Soc*, *45*(4), 224–240.

Moore, B. C., Glasberg, B., Schlueter, A. (2010). Detection of dead regions in the cochlea: Relevance for combined electric and acoustic stimulation. *Adv Otorhinolaryngol*, *67*, 43–50.

Moore, B. C., Huss, M., Vickers, D. A., et al. (2000). A test for the diagnosis of dead regions in the cochlea. *Br J Audiol*, *34*, 205–224.

Mowry, S. E., Woodson, E., Gantz, B. J. (2012). New frontiers in cochlear implantation: Acoustic plus electric hearing, hearing preservation, and more. *Otolaryngol Clin North Am*, *45*, 187–203.

Nuetzel, J., Hafter, E. (1981). Discrimination of interaural delays in complex waveforms: Spectral effects. *J Acoust Soc Am*, *69*(4), 1112–1118.

Reiss, L. A., Gantz, B. J., Turner, C. W. (2008). Cochlear implant speech processor frequency allocations may influence pitch perception. *Otol Neurotol*, *29*, 160–167.

Reiss, L. A., Turner, C. W., Erenberg, S. R., et al. (2007). Changes in pitch with a cochlear implant over time. *J Assoc Res Otolaryngol*, *8*, 241–257.

Richard, C., Ferrary, E., Borel, S., et al. (2012). Interaction between electric and acoustic cues in diotic condition for speech perception in quiet and noise by cochlear implantees. *Otol Neurotol*, *33*, 30–37.

Robles, L., & Ruggero, M. A. (2001). Mechanics of the mammalian cochlea. *Physiol Rev*, *81*, 1305–1352.

Rosen, S. (1992). Temporal information in speech: Acoustic, auditory and linguistic aspects. *Philos Trans: Biol Scien*, *336*(1278), 367–373.

Rosen, S., Faulkner, A., Wilkinson, L. (1999). Adaptation by normal listeners to upward spectral shifts of speech: Implications for cochlear implants. *J Acoust Soc Am*, *106*, 3629–3636.

Sampaio, A. L., Araújo, M. F., Oliveira, C. A. (2011). New criteria of indication and selection of patients to cochlear implant. *Int J Otolaryngol*, *2011*, 573968.

Schafer, E. C., Amlani, A. M., Paiva, D., et al. (2011). A meta-analysis to compare speech recognition in noise with bilateral cochlear implants and bimodal stimulation. *Int J Audiol*, *50*, 871–880.

Seeber, B. U., & Fastl, H. (2008). Localization cues with bilateral cochlear implants. *J Acoust Soc Am*, *123*, 1030–1042.

Shannon, R. V. (1983). Multichannel electrical stimulation of the auditory nerve in man. I. Basic psychophysics. *Hear Res*, *11*, 157–189.

Shannon, R. V., Zeng, F. G., Kamath, V., et al. (1995). Speech recognition with primarily temporal cues. *Science*, *270*, 303–304.

Shaw, E. A. (1974). Transformation of sound pressure level from the free field to the eardrum in the horizontal plane. *J Acoust Soc Am*, *56*, 1848–1861.

Simpson, A., Hersbach, A. A., McDermott, H. J. (2005). Improvements in speech perception with an experimental nonlinear frequency compression hearing device. *Int J Audiol*, *44*, 281–292.

Simpson, A., McDermott, H. J., Dowell, R. C., et al. (2009). Comparison of two frequency-to-electrode maps for acoustic-electric stimulation. *Int J Audiol*, *48*, 63–73.

Smith, Z. (2009). Speech coding of interaural time differences for bilateral cochlear implants. In *Conference on Implantable Auditory Prostheses*.

Proceedings of the Conference on implantable auditory prosthesis, July 12-17, 2009, Lake Tahoe, CA.

Smith, Z. M., & Delgutte, B. (2007). Using evoked potentials to match interaural electrode pairs with bilateral cochlear implants. *J Assoc Res Otolaryngol*, *8*, 134–151.

Spriet, A., Moonen, M., Wouters, J. (2010). Evaluation of feedback reduction techniques in hearing aids based on physical performance measures. *J Acoust Soc Am*, *128*, 1245–1261.

Stakhovskaya, O., Sridhar, D., Bonham, B. H., et al. (2007). Frequency map for the human cochlear spiral ganglion: Implications for cochlear implants. *J Assoc Res Otolaryngol*, *8*, 220–233.

Stone, M. A., & Moore, B. C. (1999). Tolerable hearing aid delays. I. Estimation of limits imposed by the auditory path alone using simulated hearing losses. *Ear Hear*, *20*, 182–192.

Stone, M. A., & Moore, B. C. (2003). Tolerable hearing aid delays. III. Effects on speech production and perception of across-frequency variation in delay. *Ear Hear*, *24*, 175–183.

Sucher, C. M., & McDermott, H. J. (2009). Bimodal stimulation: Benefits for music perception and sound quality. *Coch Impl Int*, *10* (Suppl 1), 96–99.

Svirsky, M. A., Silveira, A., Neuburger, H., et al. (2004). Long-term auditory adaptation to a modified peripheral frequency map. *Acta Otolaryngol*, *124*, 381–386.

Turner, C. W., Reiss, L. A., Gantz, B. J. (2008). Combined acoustic and electric hearing: Preserving residual acoustic hearing. *Hear Res*, *242*, 164–171.

Van den Bogaert, T., Klasen, T. J., Moonen, M., et al. (2006). Horizontal localization with bilateral hearing aids: Without is better than with. *J Acoust Soc Am*, *119*, 515–526.

van Hoesel, R. J. (2012). Contrasting benefits from contralateral implants and hearing aids in cochlear implant users. *Hear Res*, *288*, 100–113.

van Hoesel, R. J., & Tyler, R. S. (2003). Speech perception, localization, and lateralization with bilateral cochlear implants. *J Acoust Soc Am*, *113*, 1617–1630.

Vandali, A. E., Sucher, C., Tsang, D. J., et al. (2005). Pitch ranking ability of cochlear implant recipients: A comparison of sound-processing strategies. *J Acoust Soc Am*, *117*, 3126–3138.

Vandali, A. E., Whitford, L. A., Plant, K. L., et al. (2000). Speech perception as a function of electrical stimulation rate: Using the Nucleus 24 cochlear implant system. *Ear Hear*, *21*, 608–624.

Varsavsky, A., McDermott, H. (2012). Application of real-time loudness models can improve speech recognition for cochlear implant users. *IEEE Trans Neural Syst Rehabil Eng*, *21*, 81–87.

Vermeire, K., Anderson, I., Flynn, M., et al. (2008). The influence of different speech processor and hearing aid settings on speech perception outcomes in electric acoustic stimulation patients. *Ear Hear*, *29*, 76–86.

Vickers, D. A., Moore, B. C., Baer, T. (2001). Effects of low-pass filtering on the intelligibility of speech in quiet for people with and without dead regions at high frequencies. *J Acoust Soc Am*, *110*, 1164–1175.

Vollmer, M., Hartmann, R., Tillein, J. (2010). Neuronal responses in cat inferior colliculus to combined acoustic and electric stimulation. *Adv Otorhinolaryngol*, *67*, 61–69.

von Ilberg, C. A., Baumann, U., Kiefer, J., et al. (2011). Electric-acoustic stimulation of the auditory system: A review of the first decade. *Audiol Neurootol*, *16* Suppl 2, 1–30.

Wiggins, I. M., & Seeber, B. U. (2011). Dynamic-range compression affects the lateral position of sounds. *J Acoust Soc Am*, *130*, 3939–3953.

Wightman, F. L., & Kistler, D. J. (1992). The dominant role of low-frequency interaural time differences in sound localization. *J Acoust Soc Am*, *91*, 1648–1661.

Wightman, F. L., & Kistler, D. J. (1997). Monaural sound localization revisited. *J Acoust Soc Am*, *101*, 1050–1063.

Wilson, B. S., Finley, C. C., Lawson, D. T., et al. (1991). Better speech recognition with cochlear implants. *Nature*, *352*, 236–238.

Yost, W. A. (1974). Discriminations of interaural phase differences. *J Acoust Soc Am*, *55*, 1299–1303.

Zeng, F. G., & Shannon, R. V. (1992). Loudness balance between electric and acoustic stimulation. *Hear Res*, *60*, 231–235.

Zhang, T., Dorman, M. F., Spahr, A. J. (2010). Information from the voice fundamental frequency (F0) region accounts for the majority of the benefit when acoustic stimulation is added to electric stimulation. *Ear Hear*, *31*, 63–69.

Zhang, T., Spahr, A. J., Dorman, M. F., et al. (2013). Relationship between auditory function of nonimplanted ears and bimodal benefit. *Ear Hear*, *34*, 133–141.

Zierhofer, C. (2003). Electrical nerve stimulation based on channel specific sampling sequences. *US Patent 6*,594,525.