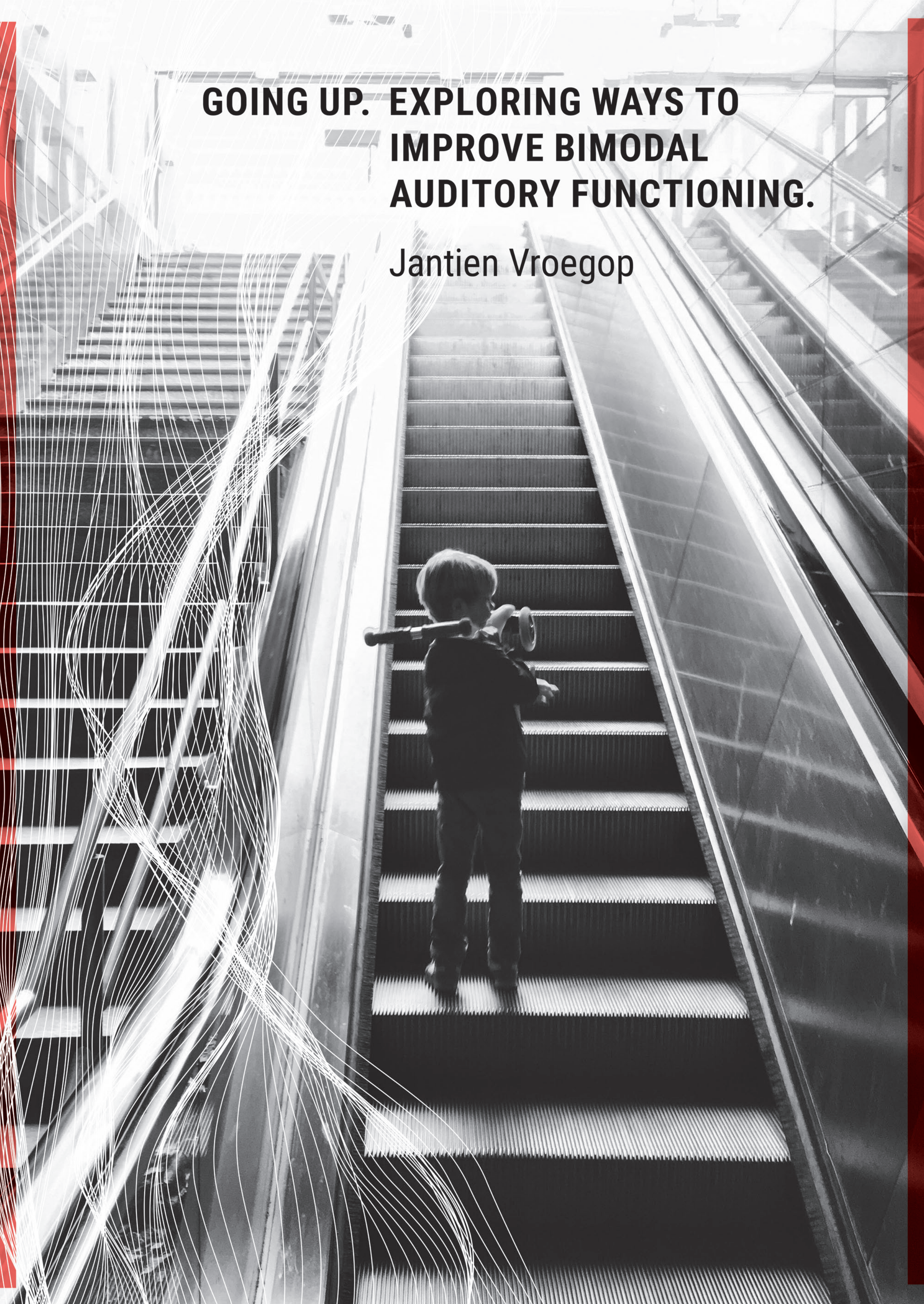

GOING UP. EXPLORING WAYS TO IMPROVE BIMODAL AUDITORY FUNCTIONING.

Jantien Vroegop

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GOING UP. EXPLORING WAYS TO IMPROVE BIMODAL AUDITORY FUNCTIONING.

ONDERZOEK NAAR MANIEREN OM HET BIMODAAL AUDITIEF FUNCTIONEREN TE VERBETEREN

PROEFSCHRIFT

ter verkrijging van de graad van doctor aan de
Erasmus Universiteit Rotterdam op gezag van de
rector magnificus
Prof.dr. R.C.M.E. Engels

en volgens besluit van het College voor Promoties.
De openbare verdediging zal plaatsvinden op

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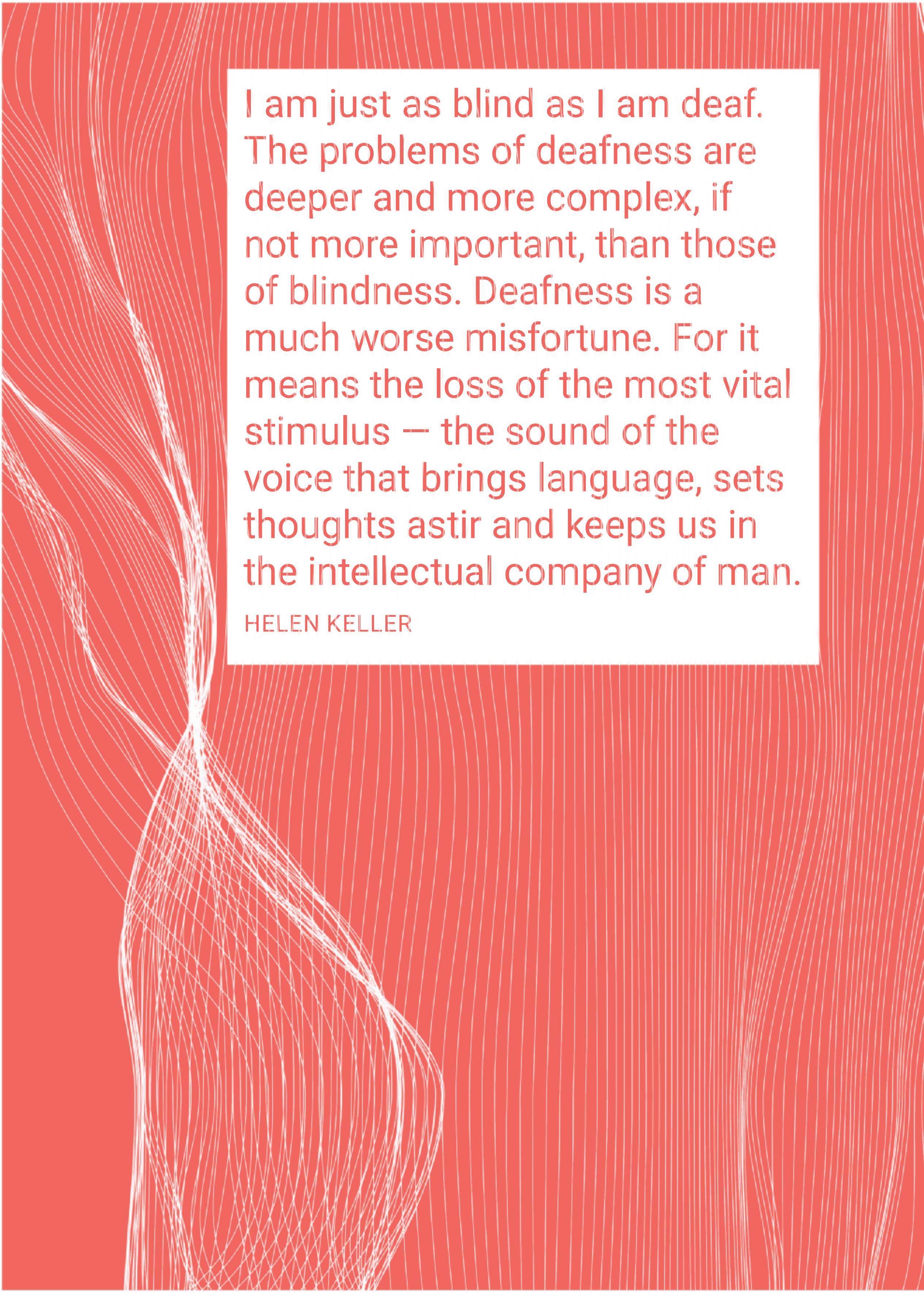

I am just as blind as I am deaf. The problems of deafness are deeper and more complex, if not more important, than those of blindness. Deafness is a much worse misfortune. For it means the loss of the most vital stimulus — the sound of the voice that brings language, sets thoughts astir and keeps us in the intellectual company of man.

HELEN KELLER

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> Hearing loss not only changes lives, it affects every aspect of life, creeping into all the small corners of daily existence that involve communication, one of the fundamentals of life along with air, water and food – and perhaps an occasional glass of wine. GAEL HANNAN

Chapter 1

GENERAL INTRODUCTION

Hearing impairment is a major health problem affecting more than 450 million people worldwide. The burden of hearing loss is higher than ever and continues to expand. Estimates of prevalence increased from 5.7 percent in 2005 to 6.4 percent in 2015 for hearing losses of more than 35 dB HL (Wilson et al. 2017). The prevalence of hearing loss dramatically rises with age. In Rotterdam (the Netherlands) prevalence numbers of 1 percent are reported in the age group 50-54 years old rising up to 77 percent in adults aging 85 years and older (Homans et al. 2017). As people get older and older, due to worldwide higher life-expectancy, prevalence of hearing loss will rise accordingly. The systematic analysis of the Global Burden of Disease Study (Vos et al. 2016) showed that hearing loss was the third leading cause of years lived with disability (YLDs) in 2016. The YLDs due to hearing loss are high, the central estimate of YLDs due to hearing loss is 4.5 percent of the total YLDs due to all causes in 2016.

Hearing loss has a substantial impact on general health and well-being. Hearing loss is associated with a lower quality of life, social isolation, poor self-esteem, cognitive decline and depression (Dalton et al. 2003, Hawkins et al. 2012, Wong et al. 2014, Dupuis et al. 2015, Pronk et al. 2011, Lin et al. 2011). Therefore, adequate and timely treatment is essential in reducing the global burden of hearing loss. As expected, the degree of hearing loss is strongly associated with hearing health care needs in various important ways. Usually hearing loss is classified into different severities varying from mild (26-40 dB HL) to profound (more than 81 dB HL). Whereas mild to moderate hearing losses can generally be adequately revalidated with conventional hearing aids (HA) or middle ear surgery (depending of the cause of the hearing problem), to individuals with severe to profound hearing loss, cochlear implants (CI) have become a viable treatment option.

A CI is a surgically implanted electronic device that allows people with severe hearing loss to access sound and to communicate more effectively with their peers. A CI bypasses the normal hearing mechanics, the spiral ganglion cells of the auditory nerve are directly stimulated by the implant, making use of the cochlear tonotopy. It consists of a sound processor (worn generally behind the ear) which processes incoming sound into an electric signal. This signal is transmitted to the implant. The implant consists of a coil which receives the signals and an array of electrodes. This array of electrodes is placed into the cochlea. The electrodes stimulate the cochlear nerve which allows the patient to hear sounds.

Cochlear implantation has caused a major shift in the treatment of severe to profound sensorineural hearing loss. In less than four decades, CIs have restored hearing of more than six hundred thousand people in the world. For deaf born children, CIs mean access to spoken language. With early bilateral cochlear implantation, prelingual deaf children without additional disabilities are able to achieve near normal language development and verbal cognition (Dettman et al. 2016, Jacobs et al. 2016, deRaeve et al. 2015). For adults who are post lingual deafened, a CI generally provides an good speech perception in quiet environments (Blamey et al. 2013; Kraaijenga et al. 2016). The high performance of most of the CI recipients coupled with the rapid evolution of implant technology lead to a distinct expansion in selection criteria for cochlear implantation. Therefore, more and more candidates with residual hearing are receiving a cochlear implant (Dowell et al., 2016; Leigh et al., 2016).

Despite the enormous added value, CI users do however experience lesser quality of sound compared to normal hearing individuals. Speech comprehension in acoustically complex real-life environments often remains a challenge, due to reverberation and disturbing background noises (Srinivasan et al., 2013, Lenarz et al., 2012). Data logs of CI processors of 1000 adult CI users show that many CI users spent large parts of their day in noisy environments, on average more than four hours a day (Busch et al., 2017). The remaining impairment in difficult listening situations can limit quality of life, professional development and social participation (Ng et al. 2015; Gygi et al. 2016;).

Nowadays, because many recipients of unilateral CIs have usable residual hearing in the non-implanted ear, contralateral hearing aids are worn frequently. The combination of a CI in one ear and a HA in the other ear is called bimodal hearing, aiming to restore binaural hearing as much as possible. It has been shown to be beneficial compared to unilateral CI use alone in several ways. It improves speech recognition in difficult listening situations, improves sound localization abilities and bimodal CI users perceive a better sound quality (Ching et al. 2007; Morera et al. 2012; Illg et al. 2014; Blamey et al. 2015; Dorman et al. 2015; Ching et al. 2004;). However, this improvement in auditory performance is not found for all bimodal CI users. Some CI users do not show bimodal improvements (Ching et al. 2004;Luntz et al.2005;Tyler et al. 2002) and even for some of the CI recipients a degradation in speech perception performance is reported when using a contralateral HA (Armstrong et al. 1997; Mok et al. 2006; Dunn et al. 2005; Veugen et al. 2016).

A possible explanation for the differences in bimodal performance between individual CI users may be the HA fitting. Fitting the CI and HA separately has been described extensively, however HA fitting procedures for bimodal CI users are not well researched or widely accepted. Nevertheless, several CI manufacturers provide HA fitting recommendations for bimodal CI users based on current, but scarce, evidence and clinical practice (Cochlear Corporation, 2012; Oticon, 2016). In recent years, the market saw most CI companies merge with traditional HA companies, in order to improve the interaction between CI processor and the HA. An example of such a partnership is that of the CI manufacturer Advanced Bionics and HA company Phonak. They recently introduced a dedicated bimodal fitting formula, the Adaptive Phonak Digital Bimodal (APDB) fitting formula (Advanced Bionics, 2016). The partnership collaboration of the CI company Cochlear and HA company Resound resulted in a new bimodal fitting flow for the hearing devices of these companies (Cochlear, 2017).

Despite these efforts, international multicenter surveys showed that although almost all clinicians would advise CI recipients to wear a contralateral HA if indicated, no dedicated HA fitting strategies were actually applied clinically to fit the hearing aid (Scherf et al. 2014; Siburt et al. 2015). It can be expected that, in order to achieve optimal bimodal hearing, specific requirements for the HA fitting are needed to reach the full hearing potential of each patient. To achieve this, more evidence about HA fitting in bimodal patients should be collected. Therefore the first part of this thesis focusses on exploring HA fitting methods to optimize bimodal auditory functioning (chapter two to four). For an optimal bimodal auditory performance, it may also be needed to adjust the settings of the CI, however, this is not investigated in the present thesis.

Another way of optimizing bimodal performance is making use of additional technology. Although bimodal listening outperforms using a CI only, speech perception in noise is still far worse compared to that of normal hearing persons. Therefore additional technology, like directional microphones, can possibly provide better speech recognition in difficult listening conditions. Directional microphones aim to improve the signal-to-noise ratio (SNR) by means of enhancing sounds of interest compared to spatially separated interfering sounds (Dillon, 2012). The introduction of directional microphones for CIs has provided a significant improvement in hearing in noise abilities (Hersbach et al., 2012, Spriet et al., 2007). The most recent development of directional HA technology involves wireless communication, which enables the exchange of audio data received by the microphones of both the left and the right HA. The increase in physical separation between the different microphones can be used to achieve narrow beamforming with further SNR improvements (Lotter and Vary, 2006). However, their effect was not evaluated before in bimodal CI users.

The use of directional microphones is often limited, as they require near field situations where the sound source is located close by and directed towards the front. Another way to improve hearing in demanding listening situations is the use of a wireless remote microphone system. Previous research has shown considerable improvement in unilateral CI users' speech recognition in noise (De Ceulaer et al., 2016, Schafer and Thibodeau, 2004, Schafer et al., 2009, Wolfe et al., 2015a, Wolfe et al., 2015b, Razza et al., 2017). Again, their effect was never evaluated in bimodal CI users before.
The second part of the thesis focusses therefore on ways of improving bimodal auditory functioning using wireless technologies (chapter five to seven).

Chapter two describes a systematic review on the effect of different HA fitting strategies on auditory performance in bimodal CI users.
In chapter three, a study to the effect of three different HA fitting approaches in bimodal CI users is described. The effect of the HA fitting method on provided HA gain and bimodal benefit is analyzed. The HA fitting methods differed in initial prescription rule and loudness balancing method.
Chapter four compares the effect of a dedicated bimodal HA fitting formula with a frequently used standard HA fitting formula. The effects of these fitting formulas are evaluated on provided HA gain and on bimodal auditory functioning.
In chapter five, the effect of a binaural beamforming technology on speech recognition in noise in bimodal CI users is investigated. Directional microphones aim to improve the signal-to-noise ratio (SNR) by means of enhancing sounds of interest compared to spatially separated interfering sounds (Dillon, 2012). The most recent development of directional HA technology involves wireless communication, which enables the exchange of audio data received by the microphones of both the left and the right HA. In this chapter, the effect of this binaural beamforming technology is investigated for bimodal auditory functioning.
Another way to improve hearing in demanding listening situations is the use of a wireless remote microphone system. Chapters six and seven describe two studies concerning the benefit of two different wireless remote microphones for speech recognition in noisy environments in bimodal adult CI users.
Chapter eight discusses the results of the studies above taken all together.

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‘Mooi hè, die krekels.’
Ik zeg: ‘Wat?’
‘Die krekels, dat geluid, mooi is
dat hè?’
Ik hoorde niets.
‘Wat hoor je dan?’
‘Dat getsjirp van die krekeltjes,
hoor je dat niet?’

Ik draaide mijn hoofd
beurtelings naar links en naar
rechts, in de hoop toch iets van
het geluid op te vangen: niets,
maar dan ook werkelijk niets.
De volkomen oprechte stomme
verbazing van mijn dochter
vervulde me met schaamte.
Heel gek – ik schaamde me
voor het feit dat ik de krekels
niet hoorde. MIKE BODDÉ

Chapter 2

HOW TO OPTIMALLY FIT A HEARING AID FOR BIMODAL CI USERS: A SYSTEMATIC REVIEW

Vroegop J.L., Goedegebure A.,
Van der Schroeff M.P.
Ear & Hearing, 2018,
39(6):1039–1045

ABSTRACT

Objective Bimodal hearing has shown to improve speech recognition in quiet and in noise and to improve sound localization compared to unilateral cochlear implant use alone. Fitting the cochlear implant (CI) and hearing aid (HA) separately has been described well, but HA fitting procedures for bimodal CI users are not well researched or widely accepted. The aim of the present study was to systematically review the literature on the effect of different hearing aid fitting strategies on auditory performance in bimodal CI users.

Design Original articles, written in English, were identified through systematic searches in Medline (OvidSP), Embase, Web-of-science, Scopus, CINAHL, Cochrane, PubMed publisher and Google Scholar. The quality of the studies was assessed on five aspects: methodological quality (with the MINORS score), number of subjects, quality of the description of contralateral hearing loss, quality of hearing aid verification, and direct comparison of hearing aid fitting procedures based on auditory performance.

Results A total of 1665 records were retrieved of which 17 were included for systematical reviews. Critical appraisal led to three high quality studies, ten medium quality studies and four low quality studies. The results of the studies were structured according to four topics: frequency response, frequency translation/transposition, dynamic range compression and loudness. In general, a bimodal benefit was found in most studies, using various strategies for the HA fitting. Using a standard prescription rule such as NAL-NL1, NAL-NL2 or DSL is a good starting point in children and adults.

Conclusion Although a bimodal benefit was found in most studies, there is no clear evidence how certain choices in hearing aid fitting contribute to optimal bimodal performance. A generally accepted HA prescription rule is an essential part of most fitting procedures used in the studies. Current evidence suggests that frequency lowering or transposition is not beneficial. Individual fine tuning based on loudness or general preference is often applied, but its additional value for auditory performance should be investigated more thoroughly. Good quality comparative studies are needed to further develop evidence-based fitting procedures in case of bimodal listening.

INTRODUCTION

Until recently, only patients with bilateral severe-to-profound hearing loss were considered candidates for a cochlear implant (CI). However, during the last years, CI candidates often have residual hearing in one or both ears as selection criteria have expanded (Dowell et al. 2016; Leigh et al. 2016). These patients are good candidates for the use of a cochlear implant in one ear and a hearing aid (HA) in the contralateral ear, which is referred to as bimodal hearing. Bimodal hearing has shown to improve speech recognition both in quiet and in noise and to improve sound localization compared to unilateral CI use alone (Blamey et al. 2015; Ching et al. 2007; Dorman et al. 2015; Illg et al. 2014; Morera et al. 2012). Although improved localization is found in studies, a considerable part of the studies showed varied results across subjects, ranging from high accuracy to no localization ability (Ching et al. 2004;Dunn et al. 2005; Seeber et al. 2004; Tyler et al. 2002). Also, for speech understanding in quiet and noise, although on average a bimodal benefit is often found, some of the subjects in the studies do not show bimodal improvements (Ching et al. 2004;Luntz et al.2005;Tyler et al. 2002). In fact, even a degradation in speech perception performance is reported for some of the subjects (Armstrong et al. 1997; Mok et al. 2006; Dunn et al. 2005; Veugen et al. 2016a).
Fitting the CI and HA separately has been described extensively, however HA fitting procedures for bimodal CI users are not well researched or widely accepted. Nevertheless, several CI manufacturers provide HA fitting recommendations for bimodal CI users based on current, but scarce, evidence and clinical practice (Cochlear Corporation, 2012; Oticon, 2016). Advanced Bionics recently introduced a dedicated bimodal fitting formula in which gain is reduced in middle to high frequencies when a dead region is suspected (Zhang et al. 2014), compression characteristics are aligned across the CI and HA (Veugen et al. 2016b) and loudness growth is aligned across the CI and HA. Also, some HA fitting protocols for bimodal CI users are proposed in literature (Ching et al. 2004; Ullauri et al. 2007).

Despite these efforts, international multicenter surveys showed that although almost all clinicians would advise CI recipients to wear a contralateral HA if indicated, no dedicated HA fitting strategies were actually applied clinically to fit the hearing aid (Scherf et al. 2014; Siburt et al. 2015). The results of Siburt et al. (2015) showed that different fitting formulas were used across clinicians for programming the hearing aid such as: National Acoustics Laboratory formula (NAL, Byrne et al. 2001), Desired Sensation Level Method (Scollie et al. 2005) or hearing aid manufacturer guidelines. Twelve percent did not reprogram the HA after cochlear implantation. Others used additional methods to reprogram the HA, including loudness balancing and adjusting of the gains based on hearing aid fitting prior to implantation. Sixty percent of clinicians used real ear measurements to verify the HA fitting.

It can be expected, that for optimal bimodal hearing, specific requirements for the HA fitting are needed to reach the full hearing potential of the patients. To achieve this, more evidence about HA fitting for bimodal patients should be collected. Structuring the available evidence is a first and important step in identifying the needs for further research. For an optimal bimodal fitting, it may also be necessary to adjust the settings of the CI, however, this is not investigated in the present study.

The aim of the present study is to systematically analyze the literature about the effect of different fitting strategies of a HA on auditory performance in bimodal CI users, contributing to the development of evidence-based HA fitting strategies to optimize auditory bimodal performance. Auditory performance is categorized in tests of speech understanding in quiet or noise and localization abilities.

METHODS

Protocol

The review was conducted and reported in compliance with the Preferred Reporting Items for Systematic Reviews and Meta-Analyses for Protocols (PRISMA-P) guidelines (Moher et al. 2015). A protocol to conduct the systematic review was specified in advance and documented according to the PRISMA-P guidelines (see supplemental file 1). It consist of a 17-item checklist and details the rational and planned methodological and analytical approach of the review.

Search strategy

Studies were identified by searching electronic databases and scanning reference lists of included articles. This search was applied to Medline (OvidSP), Embase, Web-of-science, Scopus, CINAHL, Cochrane, PubMed publisher and Google Scholar. The last search was performed on 10 February 2017(see supplemental file 2).

Study selection

Eligibility assessment was performed independently in a standardized manner by two reviewers (J.V. and M.S.). For inclusion, studies had to focus on the effect of one or more HA fitting method(s) for bimodal CI users on auditory performance. Secondly, only studies using real HA's and CI's in test sessions were deemed eligible. Studies using HA or CI simulations with normal hearing listeners or studies using insert phones instead of HAs were excluded. No publication date or publication status restrictions were imposed. No restrictions were made in relation to the age of the participants in different studies, neither was any specific type of study methodology required. Only studies written in English were included. All relevant articles were screened by title and abstract. When disagreements regarding the inclusion or exclusion of any given article arose, the two researchers discussed their rationale until agreement was reached or the third researcher (A.G.) was consulted to adjudicate. Afterwards, all eligible articles were read full text and assessed according to the inclusion and exclusion criteria, see figure 1.

Study assessment

We assessed the quality of each study on five aspects: methodological quality, number of subjects, quality of the description of the contralateral hearing loss, quality of the HA verification, and if direct comparison of hearing aid fitting procedures were performed. The methodological quality of the included articles was independently assessed by two reviewers (J.V. and M.S.) using the MINORS scale (Slim et al. 2003). This is a validated scoring tool for non-randomized studies including a 12-item assessment. Each item can be given a score from zero to two with a maximum overall score of 16 for non-comparative studies and a score of 24 for comparative studies. As all studies included in this review use participants as case and control simultaneously without using a comparative control group, the four items concerning comparative studies were omitted. The maximum MINORS score in this review therefore was 16. Criterion 6 of the MI-

NORS scale was specified to meet our research question. Because of the design of the reviewed studies, a score of two was given when the study follow-up period contained a take-home period, see table 3.
The number of subjects was given zero points if it was below 1 SD of the average number of subjects over all studies, one point if the number of subjects was between -1 SD and +1 SD of the mean and two points if the number of subjects was > 1 SD above the mean. Quality of the description of the contralateral hearing loss was given zero points if no information was available, one point if the hearing loss was plotted in a figure and two points if individual hearing losses were written in a table. Quality of HA verification was given zero points if it was not performed, one point if only 2-CC-coupler measurements were performed and two points if real-ear-measurements were done. A study was given two points if two or more different HA fitting methods were compared and zero if auditory performance was assessed with only one HA fitting method. In table 1, the criteria for the quality subscores are displayed.

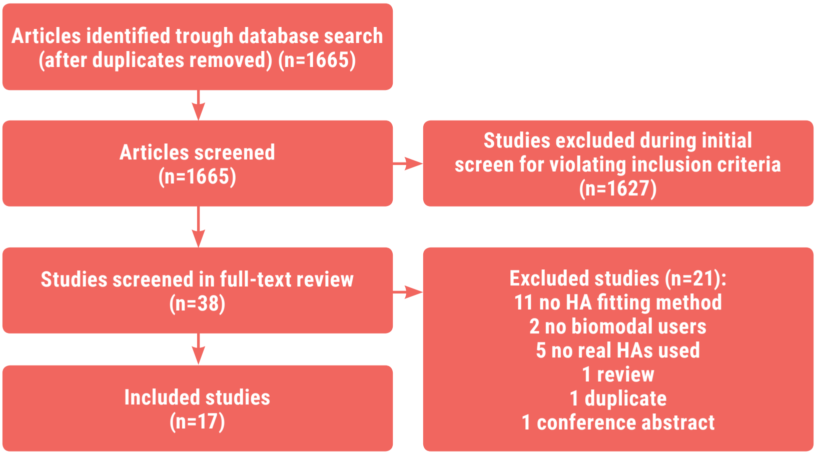

Figure 1. The Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) flow diagram of the study identification, the screening, the eligibility, and the inclusion process within the systematic search to HA fitting in bimodal CI users.

The overall quality was calculated by adding subscores 1 – 5, see table 2. We choose to combine the sub scores, as there is no uniformly accepted hierarchal order and we considered methodological quality, number of subjects, quality of the description of the contralateral hearing loss, quality of the HA verification, and direct comparison of hearing aid fitting procedures to be of equal importance.

Data extraction

From all included studies, we extracted data about the number of subjects included, study design, bimodal experience, contralateral hearing loss, basis of fitting formula and the age of the subjects. Outcome measures were the results of auditory performance tests, such as speech understanding in quiet, speech understanding in noise and sound localisation.

RESULTS

Literature search and selection

Our initial search strategy identified 1665 articles, of which 1627 were excluded because screening by title and abstract concluded that the articles did not meet the inclusion criteria. The assessment of the full-text articles resulted in another 21 excluded articles because in twelve studies no HA fitting method was investigated, in two studies the subjects were not bimodal users, in five studies no HAs were used during test sessions, two studies did not have the required format (one review, one conference abstract), and one duplicate was found, see figure 1. No additional studies were found by screening the references of the included studies. The total number of included articles was 17 (see figure 1).

Study assessment

Table 3 displays the results of the study assessment. The number of subjects in the studies ranged between 6 and 21, with a mean of 13 and a standard deviation of 4.5. Quality of description of contralateral hearing loss was poor in three studies, medium in four studies and high in ten studies. Real ear measurements of the HAs were performed in ten studies, HAs gains were tested with 2-cc coupler measurements in two studies. In five studies, no HA verification was performed. Nine studies compared auditory performance of two or more HA fitting methods. The assessment of the methodological quality of the studies (MINORS) resulted in scores between 7 and 14 (out of the possible 16) with a mean of 11.3 and a standard deviation of 1.5. Only three studies performed a prospective calculation. All, but 1 study, described a stated aim. Twelve studies included a take-home period of the HA. By adding the subscores (1-5) the overall quality assessment resulted in three high-quality studies, ten studies of medium quality and four studies of low quality.

HA fitting methods

The included studies described four different areas of HA fitting: frequency response, frequency transposition or compression, dynamic range compression and loudness. In one study, Keilmann et al. (2009), (volume) adjustments were made to the HA as well as to the CI. For all other studies, only HA adjustments were made.

Frequency response

Ten studies described the effect of HA frequency response on bimodal performance with a total of 144 subjects (Ching et al. (2001,2004 and 2005); Davidson et al. 2015; English et al. 2016; Messersmith et al. 2015; Morera et al. 2012; Neumann et al. 2013;Potts et al. 2009;Ullauri et al. 2013) , see supplemental table 4. Overall two different approaches were used. Some studies investigated the effect of more or less emphasis on high frequencies compared to the initial prescription rule (Ching et al. (2001,2004,2005);English et al. 2016;Morera et al. 2012;Ullauri et al. 2013). Other studies investigated the effect of restricted high frequency amplification (Davidson et al. 2015;Messersmith et al. 2015;Neumann et al. 2013). See supplemental table 4 for more details of the specific designs of the studies. In the studies of Ching et al.(2001, 2004 and 2005), NAL-RP, NAL-NL1 or NAL-NL2 based fittings were used, including variants with respectively more and less emphasis on high frequencies compared to the basic prescription rule. With the preferred frequency response (NAL-based in 60-80% of cases), an average bimodal benefit was shown for all tests (speech understanding in noise, and localization) compared to CI alone. Morera et al. 2012 compared auditory performance with the clinical HA settings of the subjects and the fitting procedure according to Ching et al. 2004. They did not find any difference between both fitting procedures. Davidson et al. (2015), Neuman et al. (2013) and Messersmith et al. (2015) investigated the effect of reducing the high frequency gain. In the study of Davidson et al. (2015) no differences for speech perception (noise and quiet) were found between wideband and restricted high frequency amplification. Localization performance was better with the wideband amplification. For speech perception no differences were found between bimodal and CI alone listening. Neuman et al. (2013) found a bimodal benefit for speech perception in noise and quiet for the wideband amplification and the amplification with a cutoff frequency at 2000 Hz. Lower cut off frequencies resulted in worse performance in the bimodal condition compared with the CI alone condition. In the chart review study of Messersmith et al. (2015), three poor bimodal performers fitted with a wideband amplification, did not show bimodal advantages. After fitting them with an amplification with restriction to gain above 2000 Hz the three subjects showed better performance in bimodal condition compared with the CI alone. The study of Potts et al. (2009) investigated the effect of a wideband amplification fitting formula, fitted within the subject dynamic range, on auditory performance. The results of that study showed a bimodal benefit for speech perception in quiet as well as for localization.

Frequency transposition or compression

Five studies assessed the effect of frequency transposition or compression with a total of 52 subjects (Davidson et al. 2015; Hua et al. 2012; McDermott et al. 2010; Park et al. 2012; Perreau et al. 2013), see supplemental table 4. The basic principle of frequency transposition involves transferring the high-frequency sound to a lower frequency by adding the processed signal (transposed) to the unprocessed signal in the lower frequency (Hua et al. 2012). The frequency compression technique involves decreasing the bandwidth for the output signals. The frequency shifting brings down all energy peaks at high frequencies to lower frequencies by a compression factor (Davidson et al. 2015; McDermott et al. 2010; Park et al. 2012; Perreau et al. 2013). A HA with frequency transposition or compression will increase the range of acoustic frequencies that could by perceived via the HA by CI users, who have mostly low-frequency acoustic hearing. In this way improved audibility of the high-frequency acoustic information can be obtained and possibly improved bimodal auditory functioning.

The study of Hua et al. (2012) investigated the effect of linear frequency transposition; the other studies evaluated nonlinear frequency compression. In all studies, no difference for all outcome measures was found between the frequency transposition or compression HA fitting compared with the HA fitting without frequency transposition or compression, except for the study of Perreau et al. (2013). In that study a better performance was found with the HA fitting without frequency compression.

Dynamic compression

Only one study with 15 subjects (Veugen et al. 2016b) assessed the effect of dynamic compression on auditory performance, see supplemental table 4. They matched the automatic gain control (AGC) of the HA to the AGC of the CI. This compression system was implemented as close as possible for speech signals in the AGC-matched HA as follows: (1) slow (240 and 1500 msec) and fast (3 and 80 msec) time-constants were programmed into the HA. (2) Compression channels in the HA were coupled to mimic the single channel broadband compression as present in the CI processor. They found no effect for speech understanding in quiet. For speech with a single-talker noise they found significant bimodal benefit over the CI alone for the AGC-matched HA. Subjects rated the AGC-matched HA higher than the standard HA for understanding of one person in quiet and in noise, and for the quality of sounds.

Loudness

Six studies (Ching et al. (2001, 2004, 2005);English et al. 2016;Keilmann et al. 2009;Veugen et al. 2016a) with a total of 106 subjects described the effect of loudness balancing or scaling on auditory functioning in bimodal CI users, see supplemental table 4. Five studies investigated the effect of loudness balancing between HA and CI. In almost all studies a benefit was found. Loudness balancing showed to have little effect on the final gain. The required gain differed around 3-5 dB from the gain derived with a standard fitting rule. An exception was that subjects with limited or no HA experience required seven dB less gain compared to a standardized fitting rule (Ching et al. 2005). Veugen et al. (2016a) is the only study comparing two different loudness balancing methods. They found on average no difference in auditory performance between the two balancing methods. The three-band balancing method seems to result in less gain compared with a broadband balancing. However, this data was retrieved from the HA fitting software and no real ear measurements were performed in this study. One study (Keilmann et al. 2009) performed loudness scaling in the CI and HA separately and investigated the bimodal effect. The fitting procedure for the hearing aid was based on the desired sensation level (i/o) method. A loudness scaling was used to adjust the loudness perception monaurally and to balance the volume of both the CI as well as the HA. The scaling method was repeated until a bimodal benefit was found for all subjects compared to CI alone. However, no comparison was made with the situation without applying the loudness scaling.

DISCUSSION

With this systematic review we aim to investigate how a hearing aid can be optimally fitted for bimodal CI users. We identified 17 studies, which we systematically assessed for both quality of study design and predefined outcomes of HA fitting for bimodal CI users. The quality assessment of the studies resulted in a moderate overall quality score (four studies had a total score <5, ten studies a score between 5-7 and, three studies had a score of > 7).

The studies have been systematically analyzed and structured according to the four topics of interest; frequency response, frequency compression/transposition, dynamic range compression and loudness. Although a bimodal benefit was found in most studies, no consistent differences in bimodal benefit between fitting procedures were found.

An important reason for the lack of evidence might be the limited overall quality of included studies with regard to our pre-defined targets. First of all, per topic only a low number of studies were found (1 - 10 per topic) with a limited number of subjects (13 on average). Only three of the selected studies performed a prospective calculation of the study size (English et al. 2016; Hua et al. 2012; Perreau et al. 2013). Therefore, the lack of differences between HA fitting methods in many studies might be caused by a lack of power. However, the studies which performed a sample size calculation needed 8-20 subjects according to their calculations, so it is possible that this holds for the other studies too.

More problematic were the large differences in aim, study design and reported outcome measures that were encountered. In many cases only a comparison was made between bimodal listening with a preferred HA fitting and CI alone (whether or not there was a measured bimodal benefit), instead of comparing the bimodal benefit between different types of HA fittings. Another complicating factor was that in a number of studies, different HA topics were combined in one design (Ching et al. (2001, 2004, 2005);English et al. 2016). The benefit found therefore is difficult to attribute to a specific HA fitting topic. We confined this review to qualitative descriptions only, lacking the possibility to draw strong conclusions.

The description of the contralateral hearing losses in the studies was quite good. The low frequency HL varied from 60-90 dB HL, the Fletcher index (FI) varied from 80-109 dB HL. Bimodal benefit was also found for the more severe hearing losses. This suggests that the opportunity to obtain a bimodal benefit does not depend per definition on the severity of the contralateral hearing loss. The studies which investigated the correlation between contralateral hearing loss and bimodal benefit did not find any significant correlations (Ching et al. (2001, 2004);Davidson et al. 2015;Veugen et al. (2016a, 2016b)). Five studies did not perform any HA verification (Keilmann et al. 2009; McDermott and Henshall 2010; Morera et al. 2012; Veugen et al. (2016a, 2016b)). For studies aiming to investigate HA fitting methods, this is quite remarkable. Due to ear canal anatomy, possible perforated eardrums, earmolds and ventings, real aided gain differs often from what is prescribed with the fitting software.

Frequency response is one of the most crucial elements of hearing aid fitting, so we expected this aspect to be thoroughly analyzed in HA fitting studies for bimodal CI users. Indeed, a relatively large number of studies included this factor in the design of the study (Ching et al. (2004, 2005); Davidson et al. 2015;English et al. 2016; Messersmith et al. 2015;Morera et al. 2012;Neumann et al. 2013;Potts et al. 2009; Ullauri et al. 2013). However, only three studies (Davidson et al. 2015;Neumann et al. 2013;Messersmith et al. 2015) compared relevant outcome measures obtained with different settings of the frequency response, without varying other fitting factors. In general wide-band amplification resulted in equal or better performance compared to band-limited amplification. So, this suggests to only band limit the response in special occasions, such as feedback problems of the hearing aid, user complaints about poor sound quality or the presence of cochlear dead regions (Zhang et al. 2014). Morera et al. (2012) compared auditory performance with their own HA settings of the subjects and the fitting procedure according to Ching et al. (2004). They did not find any difference between both fitting procedures.

The effect of applying shifts or tilts to a predefined frequency response is not well studied. In the included studies (Ching et al. (2001, 2004, 2005); English et al. 2016), user preference for a HF or LF-tilt in frequency response is embedded in the fitting procedure. In all these cases, a majority of the pediatric and adult bimodal users preferred the NAL. The few users that did prefer a deviating frequency response had no general preference for either a high- or a low-frequency emphasis. This suggests that a pre-described fitting based on NAL or a similar prescription rule is a good starting point in bimodal HA fitting, and may even provide a (near)-optimal solution for the majority of bimodal users. Individual fine-tuning may be helpful for a subgroup of bimodal users, although the resulting effect on auditory performance remains unclear. More and better comparative HA fitting studies for bimodal CI users are needed, to show what prescription rule provides optimal bimodal performance.

No differences were found in bimodal auditory performance in studies comparing frequency compression or transposition HA fitting methods (Davidson et al. 2015; Hua et al. 2012; McDermott et al. 2010; Park et al. 2012; Perreau et al. 2013) , except for the study of Perreau et al. (2013). In HA patients, frequency compression or transposition has shown to have the largest effect in patients with steep-slope hearing losses for the high frequencies (Ellis et al. 2015; Glista et al. 2009). In the five studies on this topic in our review, the type of hearing loss was heterogeneous between subjects (steep hearing losses as well as relatively flat hearing losses were included). It is possible, when selecting subjects with relatively good low frequency hearing and steep-slope hearing loss, more benefit can be found. Future research on this topic should focus on the effect of frequency compression for these steep-slope hearing losses. For now, current evidence suggests that frequency lowering or transposition is not beneficial for bimodal CI users.

Only one study assessed the effect of dynamic compression on auditory performance (Veugen et al. 2016b). They found a significant bimodal benefit for the AGC-matched HA in a speech test with single-talker noise , that was not found for the standard AGC setting. It draws attention to dynamic compression as a possibly relevant factor in HA fitting for bimodal CI users that may be easily overlooked. The hypothesis is that matched AGC helps to equalize loudness between HA and CI when the devices are in compression, which is favorable to binaural processing. However, more data is needed to provide clarity on this topic.

Quite a few studies (for example: McDermott and Henshall 2010; Veugen et al. 2016b) investigating one of the topics described above, also performed a broadband loudness balancing. This loudness balancing was performed as standard clinical practice and was not the aim of their study. Therefore, these studies were left out from the analysis at this topic. Only one study (Veugen et al. 2016a) compared two different loudness balancing methods. They did not find any difference in performance between broad-band and three-band loudness balancing. Other studies (Ching et al. (2001, 2004, 2005); English et al. 2016) showed that loudness balancing only had a moderate effect on the provided gain. However individual differences were quite large. More research is needed to provide insight for which patients balancing is needed and maybe provide additional bimodal benefit.

In general, a bimodal benefit was found in most studies, however there is no clear evidence how certain choices in hearing aid fitting contribute to optimal bimodal performance. A generally accepted prescription rule as NAL-NL2, NAL-RP or DSL is an essential part of most fitting procedures used in the studies. Current evidence suggests that frequency lowering or transposition is not beneficial. Individual fine tuning based on loudness or general preference is often applied, but its additional value for auditory performance should be investigated more thoroughly.

Sub score 1: number of subjects		Sub score 2: description of contra-lateral hearing loss		Sub score 3: HA verification		Sub score 4: Comparative HA fittings		Sub score 5: MINORS score	
> 17	2	table	2	REM	2	yes	2	> 13	2
9 –17	1	figure	1	2-CC-coupler	1			10 – 13	1
< 9	0	unknown	0	no	0	no	0	< 10	0

Table 1 Rating system of quality assessment using MINORS and four relevant quality parameters for HA fitting in bimodal CI users

Every left column of a sub score denotes the possible score, every right column denotes the given points. REM = real ear measurement, 2-CC-coupler = HA gain measurement with a 2-CC coupler, MINORS = Methodological index for non-randomized studies.

Overall Quality	
High quality	> 7
Medium quality	5 – 7
Low quality	< 5

Table 2 overall quality of the studies on HA fitting in bimodal CI users

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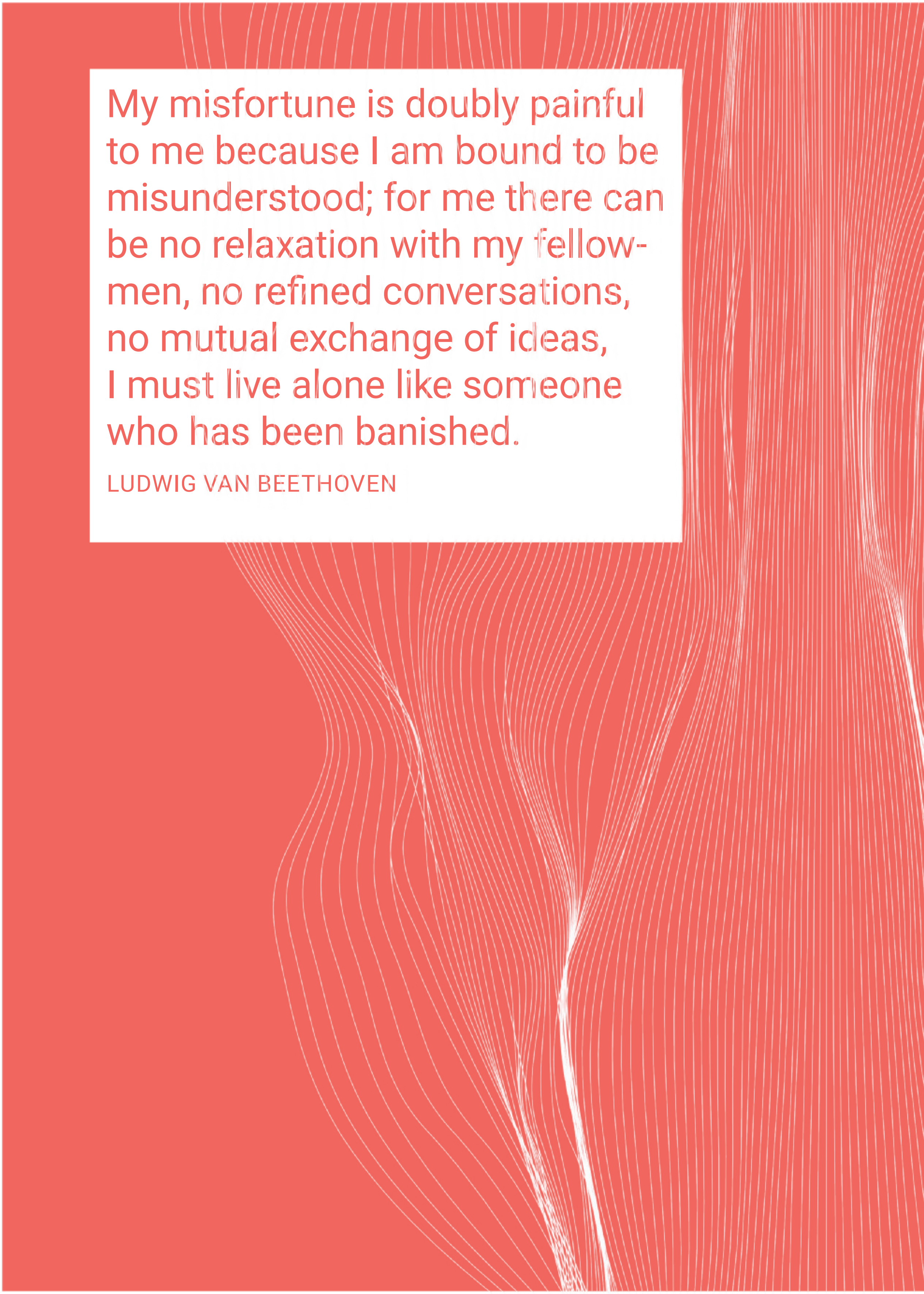
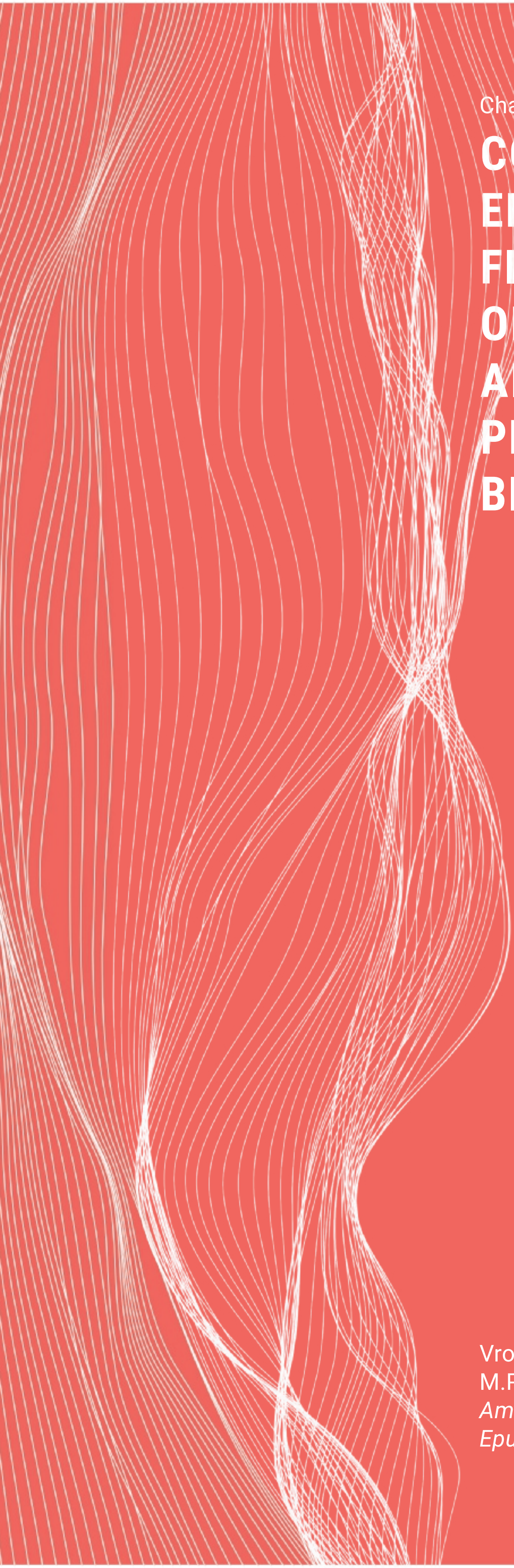


My misfortune is doubly painful to me because I am bound to be misunderstood; for me there can be no relaxation with my fellow-men, no refined conversations, no mutual exchange of ideas, I must live alone like someone who has been banished.

LUDWIG VAN BEETHOVEN



Chapter 3

COMPARING THE EFFECT OF DIFFERENT FITTING METHODS OF THE HEARING AID ON AUDITORY PERFORMANCE IN BIMODAL CI USERS

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ABSTRACT

Purpose To investigate the effect of three hearing aid fitting procedures on provided gain of the hearing aid in bimodal cochlear implant users and their effect on bimodal benefit.

Method This prospective study measured hearing aid gain and auditory performance in a cross-over design in which three hearing aid fitting methods were compared. Hearing aid fitting methods differed in initial gain prescription rule (NAL-NL2 and Audiogram+) and loudness balancing method (broadband versus narrowband loudness balancing). Auditory functioning was evaluated by a speech in quiet test, a speech in noise test and a sound localization test. Fourteen postlingually deafened adult bimodal cochlear implant users participated in the study.

Results No differences in provided gain and in bimodal performance were found for the different hearing aid fittings. For all hearing aid fittings a bimodal benefit was found for speech in noise as well as sound localization.

Conclusion Our results confirm that cochlear implant users with residual hearing in the contralateral ear substantially benefit from bimodal stimulation. However, on average, no differences were found between different types of fitting methods, varying in prescription rule and loudness balancing method.

INTRODUCTION

Until recently, only patients with bilateral severe-to-profound hearing loss were evaluated as suitable candidates for a cochlear implant (CI). Nowadays CI candidates often have residual hearing in one or both ears as selection criteria have expanded (Dowell et al. 2016; Gifford et al. 2010; Leigh et al. 2016; Mudery et al. 2017). These patients are often good candidates for the use of a cochlear implant in one ear and a hearing aid (HA) in the contralateral ear, which is referred to as bimodal hearing. Bimodal hearing has been shown to improve speech recognition in quiet and in noise and improved sound localization compared to unilateral CI use alone (Blamey et al. 2015; Ching et al. 2007, Dorman et al. 2015; Gifford et al. 2013; Illg et al. 2014; Kokkinakis et al. 2014; Morera et al. 2012). Some studies, using international multicenter surveys to study the application of bimodal fitting strategies, indicated that although almost all clinicians would advise to wear a contralateral HA if indicated, however no standard bimodal fitting strategies exist (Scherf et al. 2014, Siburt et al. 2015).

In bimodal hearing the CI will likely be the dominant device in terms of speech intelligibility and access to sound due to limited residual hearing in the contralateral ear. The HA will serve as a ‘helper’ device, which may require a different fitting approach than applying the standard prescription rules developed for hearing aids. In conjunction with expanding CI-criteria, and therefore more residual hearing in the contralateral ear, more and more symmetry in auditory performance between the CI and the HA side is expected. Likely, the degree of residual hearing is an important factor in bimodal hearing performance, but this has not been fully explored.

The aim of bimodal fitting is to optimize combined performance of the two devices for patients’ daily life. To achieve this, some form of loudness balancing across the two ears and devices is generally performed (Ching et al. 2001, 2004; Mok et al. 2006, 2010; Tyler et al. 2002; Sheffield et al. 2014; English et al. 2016; Veugen et al. 2016). Ching et al. (2007) performed a systematic review on bimodal hearing and bilateral implantation. This review includes 7 articles on bimodal localization with a total of 77 subjects and 22 articles on bimodal speech perception with a total of 199 subjects. Based on the results of the review, the authors recommended a loudness balancing procedure based on pairwise comparisons for soft and loud input sounds.

Although loudness balancing across ears is generally recommended and performed, only a few studies investigated the specific effect of balancing methods on auditory performance. Ching et al. 2001 described the effect of a loudness balancing method using three warble tones of different frequencies as stimuli for the balancing. The sixteen children in that study needed 6 dB more gain on average in comparison to a NAL-RP fitting method at 65 dB SPL. The authors hypothesized that a possible explanation for this difference in gain could be that cochlear implant mapping results in speech presented at 65 dB SPL to be perceived as comfortable, whereas NAL-RP aims to provide comfortable listening with the HA for speech presented at 70 dB SPL. The authors found better scores for speech perception and sound localization for this loudness balanced HA setting. In another study of Ching et al. (2004) the effect of a broadband loudness balancing was examined for twenty-one adult bimodal CI users. On average the subjects received 3.7 dB less gain compared to the NAL-NL1 fitting rule due to loudness balancing. Dorman et al. (2014) investigated how speech understanding varies as a function of the difference in loudness between the CI signal and the acoustic signal in five bimodal CI users. The authors showed that acoustic signals that are balanced with the CI signal provide the largest benefit to speech understanding. However, the data seemed to suggest that balancing does not need to be determined with a high degree of precision.

The study of Veugen et al. (2016) is the only published study comparing two different balancing methods. They compared frequency-dependent loudness balancing for three different frequency bands (0-548, 548-1000 and > 1000 Hz) with broadband loudness balancing. The starting point was a fitting formula which reduced gain to zero if the hearing loss exceeded 120 dB HL. The frequency-dependent loudness balancing between devices did not lead to improved speech understanding. In that study, a gain difference between the narrowband and broadband fitting was found for the mid and high frequencies, however, this was not objectified with real ear measurements.

To summarize, the results of studies to the effect of loudness balancing show that loudness balancing have little effect on the final gain settings of the HA device . The required gain after loudness balancing differed around three to five dB from the initial gain derived with a standard fitting rule. However, the effect of these balancing techniques on the final performance is still unclear (Ching et al. 2001a, 2004; English et al. 2016; Veugen et al. 2016a).

As frequency-dependent loudness balancing can be time-consuming, a fitting rule with a frequency response resembling the required gain directly could therefore be efficient. The NAL-NL2 fitting rule (which is a revised version of NAL-NL1, Keidser et al. 2011) may be a suitable option for bimodal fittings, because it is an evidence-based prescription rule optimized for maximizing intelligibility and in which equal loudness across frequencies is obtained. The fitting rule considers the impact of hearing loss in a frequency band on the ability to extract speech information within this frequency band. For severe to profound hearing losses, the fitting rule will lower the prescribed gain for frequencies that do not contribute to speech perception and will focus on more amplification on the frequencies with the better ability to extract speech cues (Johnson and Dillon, 2011; Keidser et al. 2011). English. et al. (2016) conducted a study on the effect of balancing and the use of NAL-NL2 as hearing-aid prescription in bimodal CI users. Just over half (56%) of the participants had an overall gain setting within 5 dB from NAL-NL2 target settings after loudness balancing.

Recently Cochlear Ltd started a partnership collaboration with the HA company Resound Ltd. The standard proprietary fitting algorithm of Resound Ltd is Audiogram+. The Audiogram+ gain prescription is grounded in a loudness normalization rationale, based on the results from Allen et al. (1990). For severe hearing losses, it optimizes the prescription accuracy by providing more gain for the lower frequencies compared to mild to moderately severe hearing losses (Resound, 2009). As this fitting formula is the standard fitting formula of Resound Ltd it is often applied in clinical practice. However, this fitting formula was never investigated in bimodal CI users. As most of bimodal CI users have high frequency hearing loss, audibility of low frequencies will be more important. Audiogram+ provides more low-frequency gain compared to NAL-NL2, therefore, it is possibly a better option for bimodal CI users.

The aim of our study was to compare two different loudness balancing methods (broadband versus narrowband balancing). A second aim of the study was to investigate Audiogram+ as fitting formula compared to a frequently used standard fitting formula, the NAL-NL2, in bimodal CI users. We compared the effect of the fitting procedures on real ear aided gain and on bimodal benefit.

METHODS

Participants

A total of 14 postlingually deafened adults, aged between 20-83 years (group mean age = 57; SD = 19 years) participated in this study. All were bimodal users, unilaterally implanted with the Nucleus CI24RE or CI422 implant by surgeons of the Rotterdam Cochlear Implant team at the Erasmus MC hospital in the Netherlands. Only patients with unaided hearing thresholds in the non-implanted ear better than 75 dB HL at 250 Hz were invited to participate. Figure 1 shows the unaided audiograms of the non-implanted ear for each individual. All study participants were full-time HA users and they had used their CI for at least one year prior to this study (group mean = 3.9 year, SD = 2.0 years), see table 1. All used the Nucleus 6 (CP910) sound processor for at least two months. In addition, all had open-set speech recognition of at least 60% correct phonemes at 65 dB SPL on the clinically used Dutch consonant-vowel-consonant word lists (Bosman & Smoorenburg, 1995) with the CI alone. All participants were native Dutch speakers and signed an informed consent letter before participating. Approval of the Ethics Committee of the Erasmus Medical Centre was obtained (protocol number METC253366).

Participant	**Age** (years)	**Gender**	**Implanted ear**	**Etiology**	**HA experience non-implanted ea** (years)	**CI experience** (years)
1	52	F	L	Ototoxicity	8	4
2	68	F	R	Familiar	17	1
3	34	F	L	Unknown	29	5
4	20	F	R	Genetic	19	8
5	83	M	L	Unknown	29	5
6	80	F	L	Familiar	30	2
7	50	M	L	Congenital	45	5
8	71	M	R	Unknown	26	2
9	26	F	R	Unknown	23	6
10	54	F	L	Unknown	4	2
11	69	M	R	Unknown	26	3
12	58	M	L	Familiar	46	6
13	64	F	L	Unknown	28	4
14	65	F	L	Familiar	30	2

Table 1 . Participant demographics, including details of hearing losses and HA and CI experience.

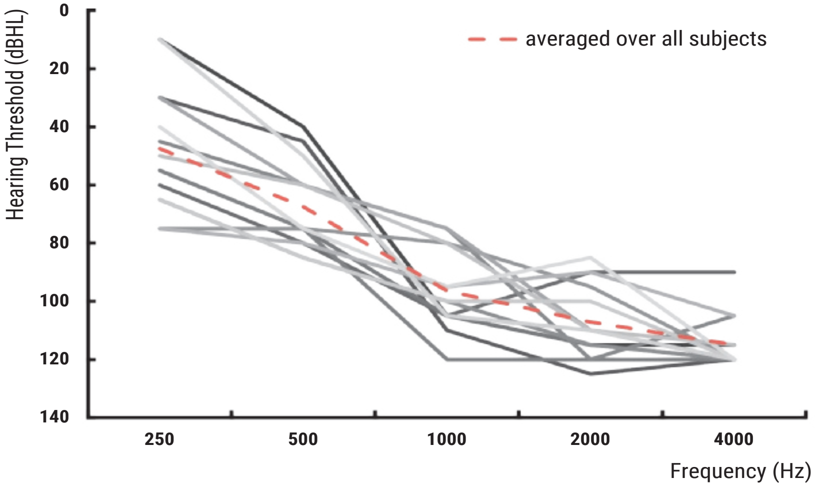
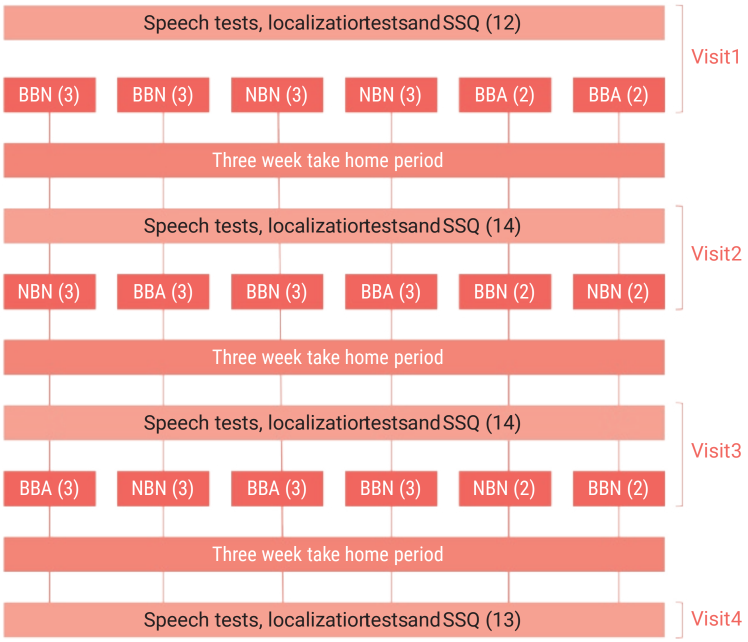


Figure 1 The hearing thresholds of the individual subjects for the ear with the hearing aid. The red line displays the group mean threshold. For calculating the average in case of ‘no response’ 5 dB extra was added to the threshold.

Study design and procedures

This prospective study used a cross-over design in which three fitting methods were compared varying in type of loudness balancing (Narrowband=NB or Broadband=BB) and basic prescription rule (NAL-NL2 or Audiogram+). A more detailed description is given in the section HA and CI fitting. The study consisted of a counter-balanced four-visit cross-over design with 3 weeks between sessions, see Figure 2.

Figure 2 Subjects were randomly distributed over the six arms of the cross-over study. The numbers between parentheses are the number of subjects. BBN denotes broadband balancing of NALNL-2 fitting formula, NBN denotes narrowband balancing of NAL-NL-2 fitting formula, and BBA denotes broadband balancing of Audiogram+ fitting rule.



Test materials

For the speech in quiet and localisation tests three conditions were tested: CI only, HA only and the bimodal condition. For the speech in noise test the CI only and bimodal condition were tested. The order of the conditions was randomized over the participants per test. For the CI only conditions, the HA was removed and a silicone rubber ear plug was used in the non-implanted ear

Speech perception in quiet test

To test speech perception in quiet the clinically used Dutch speech test of the Dutch Society of Audiology was used (Bosman & Smoorenburg 1995), which consists of phonetically balanced monosyllabics (consonant-vowel-consonant). The word lists were presented at 55 and 65 dB SPL. These levels represent the range of speech levels that participants perceive in daily life (Pearsons et al. 1977). For each condition and sound level two lists were presented.

Speech perception in noise test

For testing speech perception in noise, Dutch speech material, female voice, developed at the VU Medical Centre was used (Versfeld et al. 2000). A total of 18 sentences were presented at a fixed level of 70 dB SPL per condition. This level is representative for that of a raised voice in noisy situations (Pearsons et al. 1977). For each sentence the number of words correct was scored and the outcome variable was the percentage of words correctly repeated. The sentences were presented in steady-state, speech-shaped noise which was presented from three loudspeakers from either the front (S0N0), left (S0N-90) or right (S0N90). Prior to all other conditions, all participants started with the CI only and S0N0 condition in which the noise varied in level using an adaptive procedure to estimate the SNR that yielded a target score of 50% correct for this condition (SRT50). An extensive description of the adaptive procedure is given in Dingemanse and Goedegebure, 2015. All following noise configurations and conditions were tested at 1 dB less than this estimated SRT50 for each subject, to prevent for possible ceiling effects of the test scores.

Sound localisation test

For the sound localization test, five loudspeakers were positioned at a distance of 1 m from the subject, at -900, -450, 00, 450 and 900. Two additional loudspeakers (-1350 and 1350) were used to prevent bias in the response at -900 and 900,. Sound was only presented through these additional loudspeakers in the training session, but participants were informed that sounds were coming from all seven loudspeakers during all testing conditions. Participants were instructed to face the loudspeaker positioned at 00 azimuth. The stimuli consisted of 1 second (Verhaert et al. 2012) segments of the International Speech Test Signal (ISTS, Holube et al. 2010) at a fixed presentation level of 70 dB SPL in quiet. The ISTS included all relevant properties of speech and it is based on natural recordings of speech which is non-intelligible due to remixing and segmentation. The signal reflects a female speaker for six different mother tongues reading the same sentence. The 1 second segments were randomly chosen from the ISTS. The order of presentation was randomized across loudspeakers. The stimuli were presented 10 times by each loudspeaker, giving a total of 50 presentations. The test was a source identification task: the subject had to identify the loudspeaker from which he or she thought the stimulus was presented. The participants gave their responses by touching the number on a computer keyboard. No feedback was provided about the correctness of the answers.

Equipment
All testing was performed in a sound-attenuated booth. Participants sat one meter in front of the loudspeakers. For the speech perception in quiet conditions a clinical audiometer (Decos audiology workstation, version 210.2.6) was used. For the speech in noise and localization tests, research equipment was used consisting of a Roland UA-1010 soundcard and a fanless Amplicon PC.

Fitting
Noise reduction algorithms (like adaptive microphones and wind noise reduction) on the CI (SCAN, SNR-NR and WNR) and HA (SoundShaper, WindGuard and Noise Tracker II) were turned off during the balancing procedures and test sessions in the clinic. During the three weeks evaluation at home, the noise reduction algorithms of both the sound processor (SCAN, SNR-NR and WNR) and the HA (SoundShaper, WindGuard and Noise Tracker II on) were activated to provide optimal hearing in daily life situations. No changes in frequency allocation of the CI were made and CI users were used to their CI program for a long time. All CI users were already used to the noise reduction algorithms. The CI was held fixed at the participants' default program and volume setting for the duration of the investigations.

HA fitting
Prior to the study the required gains for the hearing losses of the participants were calculated with the Resound Aventa software for both the NAL-NL2 and the Audiogram+ fitting formula to investigate if these formulas differed from each other. The calculated gains in the Resound Aventa software differed significantly on input levels of 50, 65 and 80 dB SPL at almost all frequencies (250-4000Hz). The Audiogram+ fitting rule provided more gain (3-11 dB) in the lower and mid frequencies (<2000 Hz) and NAL-NL2 provided a higher gain (7-14 dB) for frequencies of 3000 Hz and above.
During the study participants were provided with a Resound Enzo 998 HA. The starting point for the ReSound Enzo HA fitting was the NAL-NL2 fitting rule for the BB-NALNL2 and NB-NALNL2 conditions or Audiogram+ rule for the BB-Audiogram+ condition. Real ear aided gains (REAG) were measured with the Affinity 2.0 of Interacoustics using a ISTS-signal at 55, 65 and 75 dB SPL as input. Subsequent changes in Resound Aventa software were performed during the fitting to obtain the target gain for all frequencies in the real ear and to compensate for the ear mould for the NAL-NL2 fitting. This individual earmould compensation was also used for the Audiogram+ fitting formula.

Loudness balancing procedures
For the broadband fitting a continuous ISTS-signal was presented at 55, 65 and 75 dB SPL and for each of these levels the broadband balancing was performed. A graphic of a head with an arc in the middle was used to indicate where sound was perceived, according to Dorman et al. 2014. Adjustments of the overall gain of the HA were made using an ascending/descending method in steps of 1 dB until the patient reported the signals were perceived at the midline between the ears. After completion of the loudness balancing the REAG was recorded.
The narrowband loudness balancing was performed for three frequency bands (low <500 Hz, middle 500-1000 Hz, and high >1000Hz). The gain of these frequency bands could be adjusted separately in the HA software. The ISTS-signal was band filtered with a 2nd order filter according to the three frequency bands. The band level presenta-

tion levels for the filtered signals were the same as for the broadband signal. The gain of the HA was increased or decreased identical to the broadband balancing. Again, after completion of the loudness balancing the REAG was recorded.

Data analysis

As stated before, the REAG was calculated from real ear measurements. This was done by subtracting the Real Ear Unaided Response from the Real Ear Aided Response. The effect of the loudness balancing on the REAG was calculated as the difference between the REAG before and after loudness balancing.

The effect of fitting on different aspects of binaural hearing was assessed, using the following parameters as used by Buss and colleagues (Buss et al., 2008):

Summation = S0N0bimodal – S0N0CI
Squelch = S0NHAbimodal – S0NHACI
Head-shadow = S0NHACI – S0NCICI

Numbers or abbreviations after S or N indicate the spatial position of the signal (S) and the noise (N). Subscripts indicate the listening condition. The summation effect is the benefit observed for diotic presentation of the stimulus when compared with monotic presentation of the stimulus, when stimulus signal and noise are presented from the front. Squelch refers to the capacity of the central auditory system to process the stimuli received from each ear and to reproduce it with a higher SNR when signal and noise are spatially separated. And the head-shadow effect is the attenuation due to physical placement of the head and leads to an increase in SNR in the ear far from the noise when signal and noise are spatially separated.

Data interpretation and analysis were performed with SPSS (v23). Because of the relatively low number of participants the distribution could not be determined. Therefore non parametric statistical methods were used for the analysis. For the speech recognition in quiet and in noise the Friedman test was used to compare the scores over all listening conditions. Afterwards post hoc comparisons with the Wilcoxon Signed Rank test were performed. The Benjamini-Hochberg method was used to control the false discovery rate for multiple comparisons (Benjamini and Hochberg, 1995).

RESULTS

Complete datasets are shown for 13 of 14 participants, one person did not attend the BB- Audiogram+ test session. The auditory performance results are therefore shown for 13 participants.

HA fittings

HA REAG

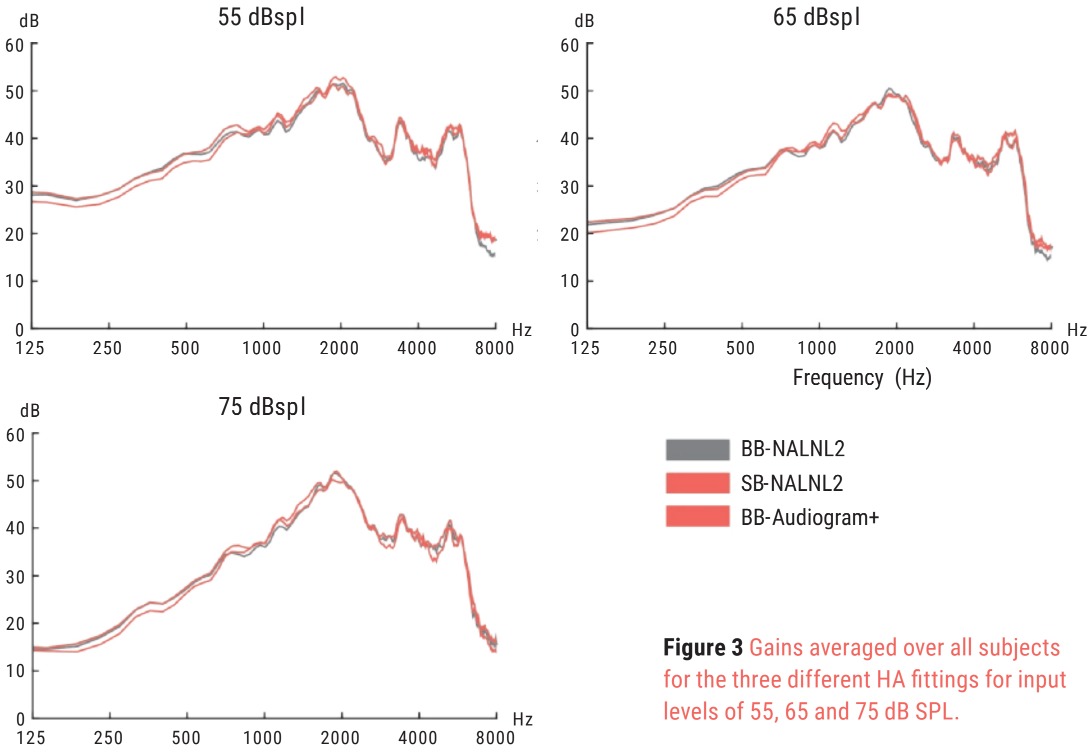


Figure 3 Gains averaged over all subjects for the three different HA fittings for input levels of 55, 65 and 75 dB SPL.

The average REAG for the three HA fittings after loudness balancing at different sound levels are presented in Figure 3. These REAGs were analyzed for significant differences for each presentation level in three different frequency bands, corresponding with the frequency bands of the narrowband loudness balancing method (< 500 Hz, 500-1000 Hz and > 1000 Hz). No statistically significant differences were found between the three HA fittings of the study (Friedman test: $p > .05$). However for three individual participants large differences (more than 8 dB) were found between the three fitting methods.

Effect of balancing on the REAG

To determine the effect of balancing on the REAG, the gain adjustments after balancing were analyzed for each of the three fitting methods. For BB-NAL-NL2 and BB-Audiogram+ there was only one gain adjustment per stimulation level as it was equally adjusted over the whole frequency band (125-8000 Hz). For NB-NAL-NL2, the gain adjustments obtained at each of the three NB frequency bands were averaged. As no significant differences were found between the adjusted gains for the different presentation levels, averaging took place over all presentation levels. No significant

Figure 4 Box-whisker plot of the absolute difference in gain after balancing for all input levels for the three HA fittings. BB-NALNL2 and NB-NALNL2 are compared with NALNL2, BB-Audiogram+ is compared with Audiogram+. Boxes represent the median (thick horizontal line), lower and upper quartiles (end of boxes), minimum and maximum values (ends of whiskers).

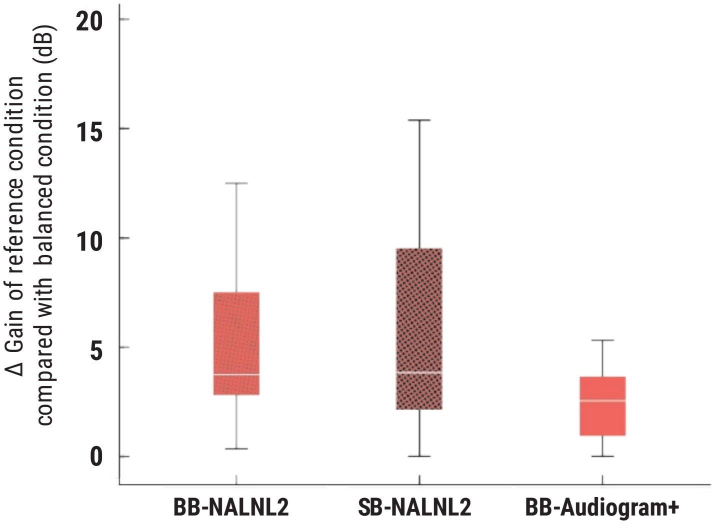
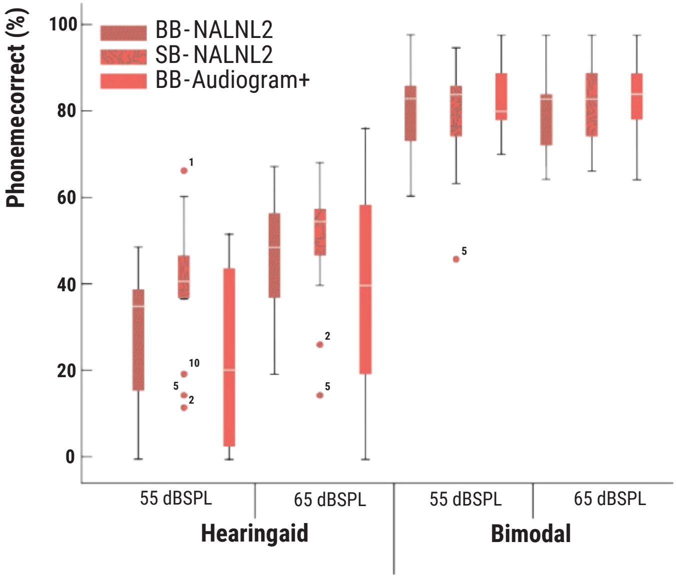


Figure 5 Box-wisker plots of speech perception in quiet at 55 dB SPL and 65 dB SPL for the HA only and bimodal condition. Boxes represent the median (thick horizontal line), lower and upper quartiles (end of boxes), minimum and maximum values (ends of whiskers), outliers (values between 1.5 and 3 times the interquartile range below the first quartile or above the third quartile – circles).



Differences in average gain were found between the conditions with and without balancing for the three fitting methods. As negative and positive gain adjustments of different subjects canceled each other, the absolute sizes of the gain adjustments after loudness balancing were analyzed in addition. Although the median size of gain adjustment is between 2 and 4 dB for all methods, the variance is remarkably larger for BB-NALNL2 and NB-NALNL2 compared to BB-Audiogram+ (see Figure 4).

Therefore, for most of the subjects NAL-NL2 requires more adjustments than Audiogram+ to obtain equal loudness balancing between HA and CI. The mean adjustment for broadband and narrowband loudness balancing for NAL-NL2 was 5 dB and for Audiogram+ was 2 dB.

Bimodal benefit

Speech perception of words in quiet

In figure 5 the results for speech perception in quiet are shown for 55 dB SPL as well as 65 dB SPL. No significant differences were found between the three HA fittings BB-NALNL2, NB-NALNL2, and BB-Audiogram+ for speech perception in quiet for the HA-only as well as the bimodal condition at 55 dB SPL and 65 dB SPL (Friedman test, p > .05). For the HA only condition an average speech perception was found of 28% (SD=20) and 41% (SD=21) for 55 dB SPL and 65 dB SPL respectively. For the bimodal condition an average speech perception was found of 82% (SD=13) and 85% (SD=9) for 55 dB SPL and 65 dB SPL respectively.

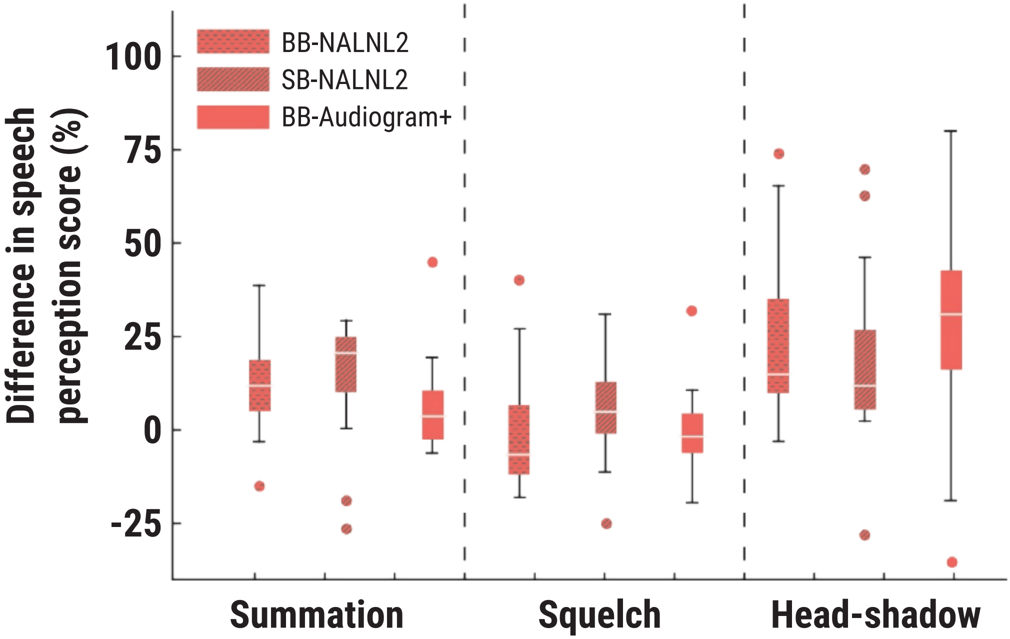
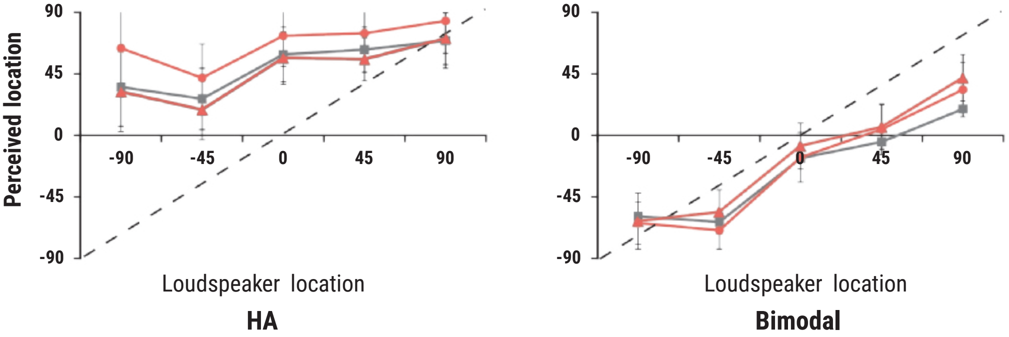


Figure 6 Box-whisker plots of summation, squelch and head-shadow effects for speech perception of sentences in noise the three different HA fittings expressed in percentage points difference. For summation and squelch, positive differences refer to bimodal advantage relative to CI alone. Boxes represent the median (thick horizontal line), lower and upper quartiles (end of boxes), minimum and maximum values (ends of whiskers), outliers (values between 1.5 and 3 times the interquartile range below the first quartile or above the third quartile – circles). The asterisk denotes an extreme value (more than 3 times the interquartile range above the third quartile.

Speech perception of sentences in noise

Figure 6 shows the results for the speech perception in noise tests for the bimodal conditions. Significant positive effects of bimodal stimulation on speech perception were found in most conditions. The average summation effect for the BB-NALNL2 setting was 13 ± 14%, for the SB-NALNL2 setting 15 ± 17%, for the BB-Audiogram+ setting 9 ± 15%. These effects were significantly different from zero (one sample Wilcoxon Signed Rank test, Z=84, p=0.008, Z=93, p=0.01, and Z=65, p=0.04 respectively). The average squelch effects were not significantly different from zero, except for the SB-NALNL2 condition. For this condition the squelch effect was 9 ± 15% (one sample Wilcoxon Signed Rank test, Z=84, p = .048). The average head shadow effects were not significantly different from zero, except for the BB-NALNL2 condition. For this condition the head shadow decreased with 12 ± 15% (one sample Wilcoxon Signed Rank test, Z=90, p = .02). The median of the head shadow effect was 14%. However, no significant differences in bimodal benefit were found between the three HA fittings for any of the measures summation, squelch or head shadow (Friedman test: p < .05).

Localization



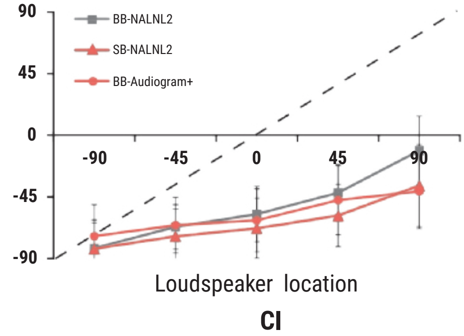


Figure 7 Localization for the CI only, the bimodal and the HA condition for the three different HA fittings. Error bars denote the standard error of the mean. The dashed line represents normal localization.

The results for the localization test are presented in Figure 7. No significant differences in localization were found between the three HA fittings for all conditions tested (bimodal, HA only and CI only, Friedman test: p > .05). Highly significant differences were found between the bimodal, HA only and CI only conditions (Friedman test: x2(13)=22.6, p = .000). Post hoc comparisons using the Wilcoxon

Signed Rank test indicated that localization improved in the bimodal condition compared to the CI only (p = .003) and the HA only condition (p = .001). P-values were not corrected for multiple comparisons. After correcting for multiple comparisons with the Benjamini-Hochberg method these differences remained significant.

Correlating HA fitting with bimodal benefit

Averaged over all subjects no significant differences were found in REAG and in bimodal benefit between the different HA fittings. However, REAG and performance levels vary substantially at an individual level. We therefore calculated the Spearman correlations of the difference in REAG between the HA fitting procedures with the difference in speech in noise performance for each individual. No significant correlations (p >.05) were found for any of the conditions.

Preferences

The majority of participants had a preference for the NB-NALNL2-fitting (62%). BB-Audiogram+ was preferred by 31% of the participants and one participant chose for BB-NALNL2 as his preferred fitting. The majority (62%) of the participants described an equal auditory performance compared to their own HA, 31% of the participants described a better auditory performance compared to their own HA, and one participant described his auditory functioning as worse compared to his own HA.

DISCUSSION

HA fitting

In this study the effect of different fitting methods on HA amplification and auditory functioning in bimodal CI users was investigated. First, the influence of the different fitting methods on the REAG of the HA was investigated. Although the simulated gains of the two fitting formulas in the Resound Aventa software were different, no significant difference in REAG was found between the loudness balanced NAL-NL2 and Audiogram+ . This was not what was expected, but some factors may be explanatory for this finding. Firstly, due to the loudness balancing procedure the differences between the two fitting rules may be reduced. Loudness seemed to be the main determinant of the preferred HA gain when balancing with the CI side. Secondly, the ear canal and earmould will affect the REAG. Therefore, the simulated gain of the fitting software was possibly different compared to the REAG, especially for the low frequencies. A third reason may be that due to feedback restrictions or limited maximum power output the gain for the higher frequencies was restricted and therefore an identical REAG for the two fitting formulas was found. In contrast, Veugen et al. (2016) described a gain difference between the narrowband and broadband fitting, that was found for the mid and high frequencies. However, in that study the simulated gain values given by the HA software were used, whereas in the current study REAGs were used. Moreover, Veugen et al. used a fitting formula which reduced gain to zero if the hearing loss exceeded 120 dB HL. When they excluded those hearing losses in their analysis, only a gain difference remained between the narrowband and broadband fitting for loud input sounds of 80 dBSPL. That makes their results more comparable with the results of our study. Loudness balancing did not result in large deviations from the prescribed gain by the initial fitting rule. The mean deviation of 2-4 dB in this study is less than the 6-dB deviation stated by Ching et al. (2001), using NAL-RP as prescription procedure. In another study of Ching et al. (2004) a deviation from the NAL-NL1 of 3.7 dB was found, which is comparable with our findings. In the study of English et al. (2016) more than half of the patients deviated less than 5 dB from the NAL-NL2 fitting rule. However, in both Ching et al. (2004) and English. et al. (2016) the fitting procedure combined the effect of different frequency responses with a loudness balancing procedure. So, the precise effect of balancing on the provided gain remained unclear. Interestingly, in some individuals in our study larger deviations were found. So, the prescription rules does not provide equal loudness balance between HA and CI for all bimodal users. This shows that fine-tuning may still be relevant to individual users. When using Audiogram+ as prescription rule, the provided gain was hardly changed after loudness balancing by any of the individuals, which means that Audiogram+ predicts the gain needed for equally balanced loudness between CI and hearing aid. This is probably because the Audiogram+ prescription aims to restore loudness as it is grounded in a loudness normalization rationale and therefore less correction is needed by loudness balancing, considering that the CI fitting is also primarily loudness based.

The repeatability of the gain differences from the loudness balancing procedure was not tested in our study. To incorporate this in future research will add valuable information for the interpretation of the findings of the study.

Bimodal benefit

Secondly, the effect of the different fitting methods on auditory functioning was investigated. No differences were found on all auditory functioning tests and the SSQ between the three HA fittings. The main reason is most probably the small differences in gain provided by the different fittings of the HA. For speech perception in quiet, no bimodal benefit was found. This is comparable with the study of Veugen et al. 2016, but in contrast to the studies of Dorman et al. 2015 and Illg et al. 2014. The subjects of our study were relatively good performers, which may have led to a ceiling effect for speech perception in quiet that did not allow for much additional improvement due to bimodal stimulation.

In noise, a consistent summation effect was found, confirming the important additional value of bimodal stimulation, regardless of the fitting method used. On average, we did not find a squelch effect. Possibly due to unsynchronized and uncoordinated modes of stimulation and independent fitting protocols. A median better ear advantage of 14% was found. These findings are in accordance with the literature (e.g. Schafer et al. 2007). While no significant mean differences were found for the summation and squelch effect, there is a tendency towards a greater variance for the BB-NALNL2 and SB-NALNL2 conditions. This suggests that some subjects could potentially get more benefit from the NAL-NL2 fittings as for the Audiogram+ fitting. For other patients this will be the other way around. This is possible due to individual differences in low frequency gain between the fitting formulas. Future studies with a larger sample size could possibly determine which specific patients will need which fitting strategies.

For all HA fittings an improvement in sound localization was found for the bimodal condition. However a bias to the CI side was found, possibly because the better auditory functioning of the CI side in comparison with the HA side. No roving was used in the design of the current study, which may have led to an overestimation of the effects found (Francart et al. 2012).

Although on average no significant differences were found between the HA fittings, individual differences were present. It could be that for specific patients more fine tuning of the HA settings improves their auditory functioning. Moreover our study population was relatively small, so small differences between settings or in auditory performance can be missed. The sample size limits the statistical power of the analysis, therefore future studies should include more subjects to enhance the statistical power of the analyses and reveal possibly more differences between methods. In our study, also subjects with a wide-range of residual hearing are included, which has possibly resulted in the non-uniform results. Future studies should also assess bimodal performance for the fitting algorithms without balancing, so the effect of balancing can be determined more thorough.

Another aspect that may have had impact on our results is the use of stationary noise for the speech perception in noise test. Some studies suggest (Veugen et al. 2016b, Illg et al. 2014) that bimodal benefit is larger when multi talker babbles are chosen instead of stationary noises as noise stimuli. Future research should therefore also incorporate this type of stimulus.

On average, loudness balancing did not result in large deviations from the prescribed HA gain. However, for some individuals larger deviations were found. For clinical practice, we recommend therefore to perform a simple broadband loudness balancing procedure to check if equal loudness between HA and CI is established. More evidence is needed for more precise/extended procedures.

CONCLUSION

The results of this study showed that different hearing aid fitting methods applied in bimodal CI users resulted in comparable real ear aided gains. Our results confirm that cochlear implant users with residual hearing in the contralateral ear substantially benefit from bimodal stimulation. However, on average, no differences were found between different types of fitting methods, varying in prescription rule and loudness balancing method. Apparently, more research is needed, using larger sample sizes, to reveal possible overall differences of fitting in auditory performance.

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> I’m always trying to figure out what’s really going on. Always having to fill in the gaps, but never getting all the details. It’s like trying to do a jigsaw when I don’t even know what the picture is, and I’m missing one of the vital middle pieces.
>
> CHRISSIE PERRY

Chapter 4

COMPARING TWO HEARING AID FITTING ALGORITHMS FOR BIMODAL COCHLEAR IMPLANT USERS

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ABSTRACT

Objectives To investigate the possible advantage of the use of a dedicated bimodal hearing aid fitting formula, the Adaptive Phonak Digital Bimodal (APDB), compared to a frequently used standard hearing aid fitting formula, the NAL-NL2. We evaluated the effects of bimodal hearing aid fitting on provided hearing aid gain and on bimodal auditory functioning in a group of experienced bimodal cochlear implant users. A second aim of our study was to determine the effect of broadband loudness balancing on the prescribed gain of those two fitting formulas.

Design This prospective study used a cross-over design in which two fitting methods were compared varying in basic prescription formula (NAL-NL2 or APDB fitting formula). The study consisted of a three-visit cross-over design with three weeks between sessions. Nineteen postlingually deafened experienced bimodal cochlear implant users participated in this study. Auditory functioning was evaluated by a speech in quiet test, a speech in noise test, and a questionnaire on auditory performance.

Results Significant differences between the two fitting formulas were found for frequencies of 2000 Hz and above. For these frequencies less gain was provided by the APDB fitting formula compared to NAL-NL2. For the APDB fitting formula a higher compression ratio for frequencies of 1000 Hz and above was found compared with the NAL-NL2 fitting formula. Loudness balancing did not result in large deviations from the prescribed gain by the initial fitting formula. Bimodal benefit was found for speech perception in quiet as well as for speech perception in noise. No differences in auditory performance were found between the two fitting formulas for any of the auditory performance tests.

Conclusion The results of this study show that cochlear implant users with residual hearing at the contralateral ear can benefit from bimodal stimulation, regardless of the fitting method which was applied. Although significant differences between the output and compression ratio of the NAL-NL2 and the APDB fitting formula existed, no differences in bimodal auditory performance were observed. Therefore, NAL-NL2 or the APDB fitting prescription both seem suited for bimodal fitting purposes. Additional loudness balancing has a marginal effect on the provided hearing aid output.

INTRODUCTION

One of the main challenges for patients with cochlear implants (CI) is speech comprehension in acoustically complex real-life environments due to reverberation and disturbing background noises ((Lenarz et al. 2012; Srinivasan et al. 2013). Data logs of CI processors of 1000 adult CI users show that many CI users spent large parts of their day in noisy environments, on average more than four hours a day (Busch et al. 2017). Although auditory performance in quiet environments is generally acceptable, the remaining impairment in difficult listening situations can limit quality of life, professional development and social participation (Ng et al. 2015; Gygi et al. 2016;).

Wearing a hearing aid (HA) in the contralateral ear, referred to as bimodal hearing, has been shown to improve auditory functioning (Blamey et al. 2015; Ching et al. 2007). Speech recognition in quiet (English et al. 2016), in steady state noise (Morera et al. 2012; Illg et al. 2014; Vroegop et al. 2017), in competing talker noise (Illg et al. 2014), and in babble noise (Dorman et al. 2015) improved compared to unilateral CI use alone. HA fitting has been described extensively, however HA fitting procedures for bimodal CI users are not well researched or widely accepted. Nevertheless, several CI manufacturers provide HA fitting recommendations for bimodal CI users based on current, but scarce, evidence and clinical practice (Cochlear Corporation, 2012; Oticon, 2016). Despite these efforts, international multicenter surveys showed that although almost all clinicians would advise to wear a contralateral HA if indicated, no dedicated HA fitting strategies were actually applied clinically to fit the hearing aid (Scherf et al. 2014; Siburt et al. 2015).

The general aim of the HA fitting in bimodal CI users is to optimize the additional auditory input provided by the HA in the contralateral ear in various daily life conditions. In most studies on this topic, some form of loudness balancing between the HA and the CI is performed (Ching et al. 2001a, 2004; Tyler et al. 2002; Mok et al. 2006; English et al. 2016) . But still, the effect of these balancing techniques on the final performance is still unclear (Ching et al. 2001a, 2004; English et al. 2016; Veugen et al. 2016a). Loudness balancing showed to have little effect on the final gain. The required gain after loudness balancing differed around three to five dB from the initial gain derived with a standard fitting rule. An exception was that listeners with limited or no HA experience required seven dB less gain compared to a standardized fitting rule (Ching et al. 2004). Veugen et al. (2016a) is the only study comparing two different loudness balancing methods. They found on average no difference in auditory performance between the two balancing methods.

Usually, the starting point for normal HA fitting is a pre-calculated gain target that is derived from a pure-tone audiogram. A commonly used fitting formula , the NAL-NL2 (which is a revised version of NAL-NL1, (Keidser et al. 2012)), is an evidence-based prescription formula that is optimized for preferred overall loudness for a variety of hearing aid users and hearing losses. It considers the impact of raised hearing thresholds on the ability to extract speech information within separate frequency bands. For severe to profound hearing losses, the fitting formula will lower the prescribed gain for frequencies that are not expected to contribute to speech perception and provide more amplification for the frequencies with the better ability to extract speech cues (Johnson et al. 2011; Keidser et al. 2012). NAL-based fitting formulas have also been used frequently in bimodal HA fitting (Ching et al. 2001a; Perreau et al. 2013; Messersmith et al. 2015; English et al. 2016).

Recently Phonak Ltd (Stäfa, Swiss) introduced the Phonak Naída Link hearing aid. A special prescriptive fitting formula, the Adaptive Phonak Digital Bimodal (APDB) fitting formula, for bimodal hearing is used for this hearing aid (Advanced Bionics, 2016). For this fitting formula additional loudness balancing seems not be required. This formula differs from more standard fitting formulas on three aspects: the frequency response, the loudness growth and the dynamic compression. Firstly, the APDB fitting formula aims to align the frequency response by optimizing low-frequency gain and bandwidth. Low-frequency gain is increased using the model of effective audibility to ensure audibility of speech recognition in quiet (55 dB SPL) environments (Ching et al. 2001b). Frequency bandwidth is optimized by assuring that bandwidth is as wide as possible, based on a study of Neuman et al. 2013. In that study the effect of frequency bandwidth on bimodal auditory performance was investigated. They found that smaller frequency bandwidths resulted in worse performance in the bimodal condition compared with the CI alone condition. Besides, the fitting formula ensures frequencies between 250 and 750 Hz are audible (Sheffield et al. 2014), and that amplification does not extend into presumed dead regions (Zhang et al. 2014). To obtain the latter, a reduced frequency bandwidth of the gain is applied if the slope of the hearing loss is more than 35 dB per octave and the high frequency hearing loss exceeds 85 dB HL or if the hearing loss is more than 110 dB HL.

Secondly, the loudness growth is aligned by implementing the input-output function of the cochlear implant in the hearing aid. Thirdly, the dynamic compression behavior is aligned by porting the CI dual-loop AGC into the hearing aid. The CI processor has a single-channel dual-loop AGC system, incorporating both slow and fast attack and release time-constant circuits. This compression system was implemented as close as possible for speech signals in the AGC-match HA as follows: (1) Slow and fast time-constants were programmed in the HA, (2) Compression channels in the HA were coupled to mimic the single channel broadband compression as present in the CI processor. In both devices the compression knee point was fixed at 63 dB SPL, considering the long-term average speech spectrum. A more detailed description of this AGC alignment is given in Veugen et al. 2016b. However, although all different sub parts of this fitting formula are based on scientific research as described above, the effect of combining these algorithms into one prescriptive fitting formula on bimodal performance has not been subject to study yet.

Therefore the aim of our study is to investigate the possible advantage of using a dedicated bimodal HA fitting formula, the APDB fitting formula, compared to a frequently used standard fitting formula, the NAL-NL2. We evaluated the effects of bimodal HA fitting on provided HA gain and on bimodal auditory functioning in a group of experienced bimodal CI users.

Because the effect of balancing techniques for bimodal hearing is still unclear, a secondary aim of our study is to determine the effect of broadband loudness balancing on the prescribed gain of those two fitting formulas.

MATERIALS AND METHODS

Participants

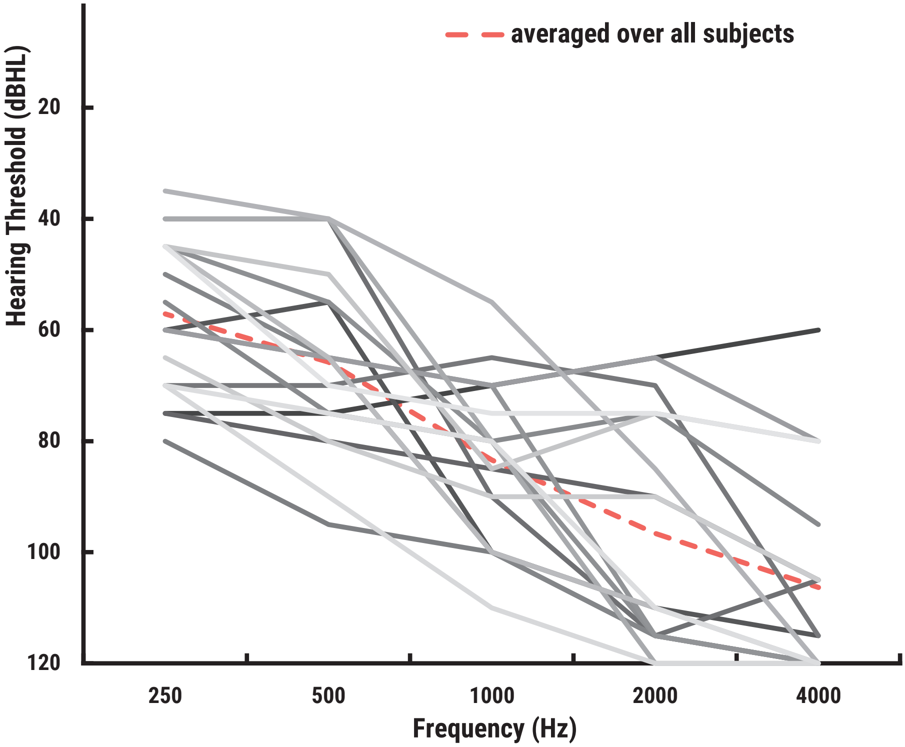


Figure 1 The hearing thresholds of the individual subjects for the ear with the hearing aid. The red line displays the group mean threshold. For calculating the average in case of 'no response' 5 dB extra was added to the threshold.

A total of 19 postlingually deafened adults participated in this study, see table 1 for demographics. Participants ranged in age from 32 to 81 years old (group mean age = 62; SD = 15 years). All were experienced bimodal users, unilaterally implanted with an AB HiRes 90K implant by surgeons of four different cochlear implant teams in the Netherlands. All participants had used their cochlear implant for at least six months prior to this study (average = 4 years, SD = 3,5 years). All participants used the AB Naída Q70 or AB Naída Q90 sound processor in daily life. During the test sessions those with a Naída Q70 were given the Naída Q90 sound processor. In addition, all had open-set speech recognition of at least 70% correct phonemes at 65 dB SPL on the clinically used Dutch consonant-vowel-consonant word lists (Bosman et al. 1995) with the cochlear implant alone. Only participants with unaided hearing thresholds in the non-implanted ear of 80 dB HL or better at 250 Hz were included. Figure 1 shows the unaided audiograms of the non-implanted ear of the individual participants. All participants used a hearing aid prior to the study, which was replaced by the Phonak Naída Link UP HA for the duration of this study. All participants were native Dutch speakers who signed an informed consent letter before participating in the study. Approval of the Ethics Committee of the Erasmus Medical Centre was obtained (protocol number METC306849).

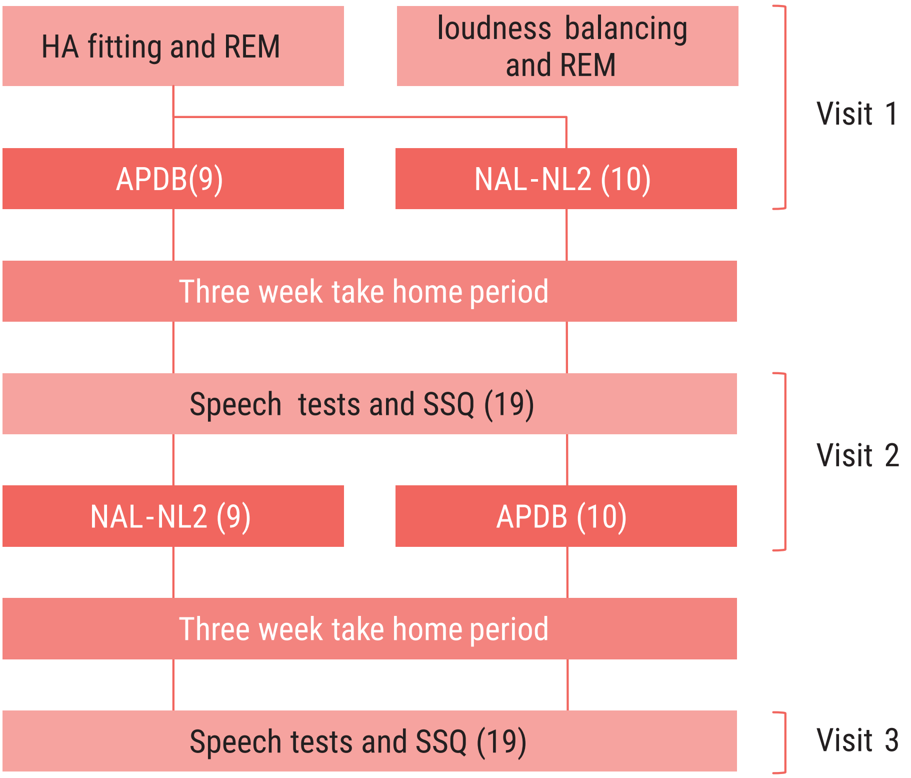


Figure 2 Subjects were randomly distributed over the two arms of the cross-over study. The numbers between parentheses are the number of subjects. REM denotes real ear measurement, APDB denotes adaptive Phonak digital bimodal fitting formula.

Study design and procedures

This prospective study used a cross-over design in which two fitting methods were compared varying in basic prescription formula (NAL-NL2 or APDB fitting formula). The study consisted of a three-visit cross-over design with 3 weeks between sessions, see Figure 2.

Test materials

For all auditory functioning tests two conditions were tested: CI only and the bimodal condition. The order of the conditions was randomized over the participants per test.
To test speech perception in quiet we opted for the clinically used Dutch speech test of the Dutch Society of Audiology (NVA-list, Bosman and Smoorenburg 1995), which consists of phonetically balanced monosyllabics (consonant-vowel-consonant). The word lists were presented at 45 and 55 dB SPL. These levels represent the range of speech levels that participants perceive in relatively quiet speech environments (Pearsons et al. 1977). For each condition and sound level two lists were presented.
For testing speech perception in noise, Dutch speech material developed at the VU Medical Centre was used (Versfeld et al. 2000). The speech material (female voice) was presented at a fixed level of 70 dB SPL per condition. This level is representative for that of a raised voice in noisy situations (Pearsons et al. 1977). The sentences

were presented in steady-state, speech-shaped noise which was presented from either the front (S0N0), left (S0N-90) or right (S0N90) side. We scored the correct words per sentence per list. An adaptive procedure was used to calculate the signal-to-noise ratio targeting at a score of 50% correct words (Speech Reception Threshold or SRT). An extensive description of the adaptive procedure is given in Dingemanse et al. 2015. For each condition and for each subject a list with 20 sentences was randomly selected from a total of 25 lists.

All testing was performed in a sound-attenuated booth. For the speech perception in quiet conditions a clinical audiometer (Decos audiology workstation, version 210.2.6) was used. For the speech in noise tests, research equipment was used consisting of a Roland UA-1010 soundcard and a fanless Amplicon personal computer. Participants were placed in front of the loudspeakers at a distance of one meter.

The Speech Spatial and Qualities of Hearing questionnaire (Gatehouse et al. 2004) was completed by the participants after each of the 3-week home-trial periods at the second and third visit. The SSQ questionnaire (48 questions) reflects perceived hearing abilities in everyday life and consists of three domains: speech comprehension (14 questions), spatial hearing (17 questions) and quality of sound (18 questions). The questionnaire was rated on a 0 (not good) to 10 (perfect) scale per question. The score for each domain was calculated by averaging the sub scores per question.

HA and CI fitting

For the duration of the study participants were provided with a Phonak Naída Link UP hearing aid. Loudness balancing procedures were conducted to measure the effect on the HA output, not for the experimental testing. Real ear measurements (REM) were performed with an Interacoustic Affinity 2.0 system. The International Speech Test Signal (ISTS, Holube et al. 2010) with a duration of 25 seconds was used as input signal for input levels of 55, 65 and 75 dB SPL. The NAL-NL2 output target was shown on the monitor. The gain of the HA was iteratively adjusted using (Phonak) Target software to obtain the required in-situ output for at least 500 Hz until 1000Hz. In this way we could compensate for the individual shape of the ear canal and the ear mould. For the APDB fitting formula, no gain changes were made. As the APDB fitting formula is a proprietary fitting algorithm of Phonak, no fitting targets are available in the real ear measurement software. Also, the manufacturer recommends to use this fitting formula as predicted by the fitting software. At the end of the fitting the Real Ear aided Response (REAR) were recorded for the NAL-NL2 fitting formula and for the APDB fitting formula. Additionally, broadband balancing was performed to assess the gain adjustments needed for the HA to obtain an equal loudness percept with the CI. A continuous ISTS-signal was presented at 55, 65 and 75 dB SPL and for each of these levels the balancing was performed. Adjustments of the overall gain of the HA were made using an ascending/descending method until the patient reported the signals were perceived at the midline between the ears (comparable with the procedure used in Dorman et al. 2014). After completion of the loudness balancing a second REAR was recorded. The REAR was performed to measure the effect of balancing on the provided gain of the HA. Evaluation of the HA at home and bimodal auditory functioning testing were performed with NAL-NL2 or the APDB fitting formula without balancing.

Noise reduction algorithms on the cochlear implant (Clearvoice, Windblock, Soundrelax) and HA (Noiseblock, Soundrelax, Windblock) were turned off during the balancing procedures and test sessions in the clinic. During the three weeks evaluation at home, the noise reduction algorithms of both the sound processor and the HA were activated to provide optimal hearing and comfort in daily life situations. All CI users were already familiar with the use of noise reduction algorithms in their daily life CI and HA programs. The settings of the CI processor were held fixed at the participants' default program and volume setting during the whole test period. During the test sessions in the clinic the same AB Naída Q90 sound processor and the same Phonak Naída Link HA were used for all participants.

Data analysis

The compression ratio was calculated with the REAR data per octave band by the following formula:

Compression ratio = Δinput/Δoutput

where Δinput =10 and Δoutput=the change in REAR output between inputs at 55 and 65 dB SPL. Δoutput is retrieved from the REAR data.

An 'a priori Power Analysis' is performed with a required power of 0.8 and a significance criterion of 0.05, using a Wilcoxon singed-rank test model with G*Power. For speech perception a clinical significant difference is a difference ≥ 15%. With a slope of the psychometric function of 6.4% / dB on average (Dingemanse et al. 2015), the difference between two test conditions must be ≥ 2 dB to be clinically significant. We planned paired comparisons between several test conditions. With a minimum of 2dB between groups the effect size is 0.71. With these input parameters, the required number of participants is 15.
Data interpretation and analysis were performed with SPSS (v23). Because of the low number of participants, non-parametric statistical methods were used. Firstly, the Friedman test was used to compare differences over conditions. Afterwards, post hoc comparisons with the Wilcoxon Signed Rank test were performed.

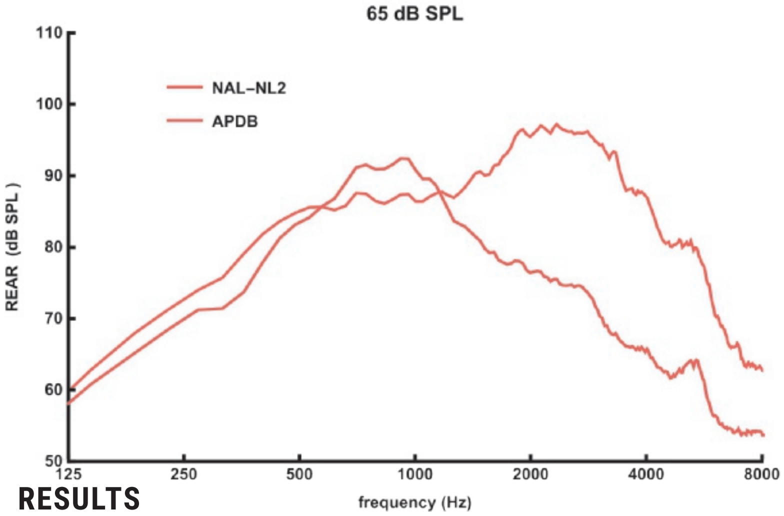


Figure 3 REAR averaged over all subjects for the two different hearing aid fittings for an input level of 65 SPL.

RESULTS

Real Ear Aided Response

Effect of fitting formula

Figure 3 shows the average REAR for the APDB fitting formula as well as the NAL-NL2 fitting formula at an input level of 65 dB SPL. We calculated the HA output levels per frequency band for each of the individual REARs to allow for quantitative comparisons. We found significant differences between the REAR of the two fitting formulas (Friedman test: x2(2) = 178, p <0.0001). Post hoc comparisons using the Wilcoxon Signed Rank test showed significant differences in the REAR for the octave bands of respectively 2000, 4000 and 8000 Hz (Wilcoxon Signed Rank test, Z=-3.334, p=0.001, Z=-3.703, p<0.0001 and Z=-3.282, p=0.001). For the lower frequencies (250, 500 and 1000 Hz) no significant differences in HA output were found between the fitting formulas. Input levels of 55 and 75 dB SPL

showed similar differences between the two fitting formulas. Table 2 displays the output levels for the APDB and NAL-NL2 fitting formulas. In table 3 the differences between the two fitting formulas per octave band are given.

To compensate for the individual shape of the ear canal and the ear mould, the first fit with NAL-NL2 was adapted until the target curves in the REAR were reached. This resulted in an average adjustment over all participants of 0 to +4 dB for the different frequency bands, see table 3 for more details. The REAR of the first fit is also compared with the REAR of the bimodal fitting formula. Similar significant differences between the two fitting formulas were found.

Effect of hearing loss on REAR

Participants with different types of hearing loss were included in this study. To investigate the effect of the different types of hearing loss on the provided gain, the participants were divided in two groups. The hearing loss was classified as (relatively) flat or steep. A flat hearing loss was defined as a difference in hearing loss of less than 40 dB between 250 Hz and 2 kHz, a steep hearing loss was defined as a difference in hearing loss of 40 dB or more between 250 Hz and 2 kHz. Figure 4 shows the average REAR for each of the two groups. For the flat hearing losses, the REAR of the APDB fitting formula was not significantly different from the REAR of the NAL-NL2 fitting formula. For the group participants with a steep hearing loss, the REAR of the NAL-NL2 was higher compared with the APDB fitting formula for the frequencies of 2000, 4000 and 8000 Hz respectively (Wilcoxon Signed Rank test, Z=-2.905, p=0.005, Z=-2.805, p<0.0005 and Z=-2.805, p<0.0005).

Compression

Figure 5 shows the REAR for the three input levels (55, 65 and 75 dB SPL) for the APDB and the NAL-NL2 fitting formula. The compression ratio was calculated by the formula described in the methods. For the compression ratio between input levels of 55 and 65 dB SPL no significant differences between the two fitting formulas were found. For the compression ratios between input levels of 65 and 75 dB SPL, a significant higher compression for the APDB fitting formula compared with the NAL-NL2 fitting formula was found for frequencies of 1000 Hz, 2000 Hz and 4000 Hz respectively (Wilcoxon Signed Rank test, Z=-2.993, p=0.003, Z=-2.924, p=0.003 and Z=-2.782, p=0.005). Table 4 shows the different compression ratios averaged over patients per octave band for the input levels of 65 to 75 dB SPL.

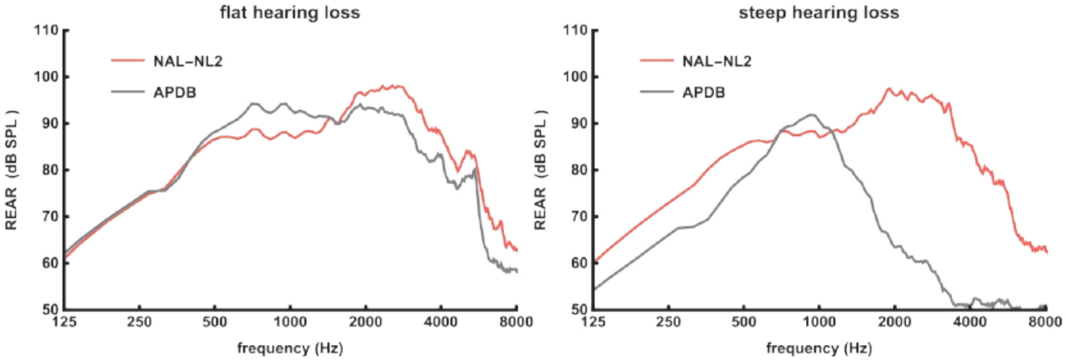
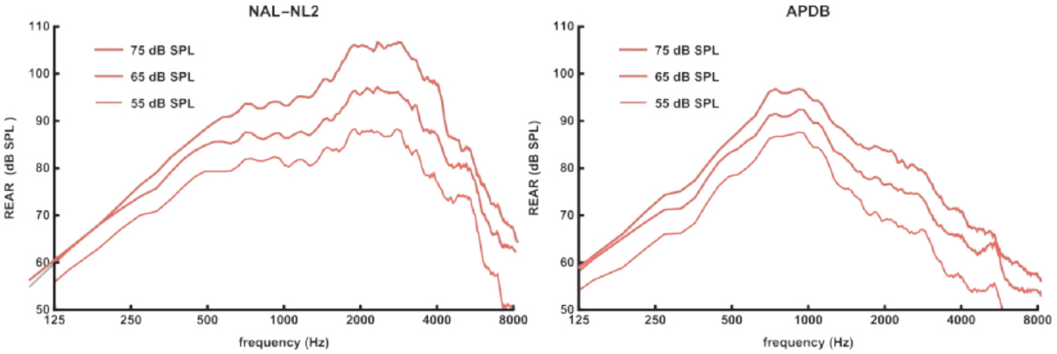


Figure 4 REAR averaged over two groups of subjects for the two different HA fittings for an input level of 65 SPL. On the left the REAR for the group with the flat hearing losses is shown, on the right, the REAR for the group with the steep hearing losses is shown.

Figure 5 REAR for the NAL-NL2 fitting formula (left) and the APDB fitting formula (right) for input levels of 55, 65 and 75 dB SPL.



Effect of broadband loudness balancing

For the balancing analysis the HA output levels were calculated per frequency band for each of the individual REARs. No significant differences were found in overall output level between the conditions with and without balancing for each of the two fitting formulas. As negative and positive gain adjustments of different subjects may cancel each other, the absolute gain adjustments after loudness balancing were analyzed in addition (see figure 6). The median size of gain adjustment is relatively small and comparable for the two fitting formulas (between 0.6 and 1.6 dB), although the variance is larger for the APDB fitting formula.

To investigate the effect of hearing loss of the contralateral ear on the direction and amount of gain adjustments due to balancing, the correlations (Spearman) between the gain adjustments per fitting formula and the contralateral hearing loss per octave band frequency were obtained. A significant correlation of -0.480 (p=0.038) was found between the gain adjustments due to balancing with the APDB fitting formula and the hearing loss at 250 Hz. Subjects with more severe loss at 250 Hz tend to require less gain than prescribed to obtain equal loudness balance between HA and CI. No other correlations were found.

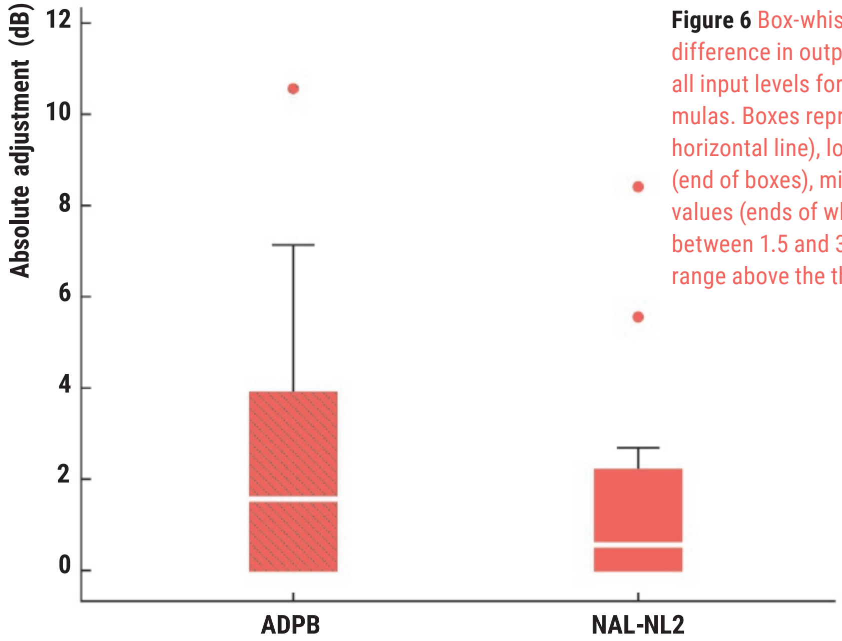


Figure 6 Box-whisker plot of the absolute difference in output after balancing for all input levels for the two HA fitting formulas. Boxes represent the median (thick horizontal line), lower and upper quartiles (end of boxes), minimum and maximum values (ends of whiskers), outliers (values between 1.5 and 3 times the interquartile range above the third quartile – circles).

Bimodal auditory functioning

We tested the CI only condition in quiet as well as in noise in every test session, to be able to determine possible learning effects. No significant differences in CI only scores were observed between the two test sessions on any of the test conditions. For the analysis of bimodal benefit, the CI only scores of session one and session two were averaged.

Speech perception in quiet

No significant differences between the bimodal performance for the two fitting formulas were found for 45 dB SPL as well as 55 dB SPL (Wilcoxon Signed Rank test, Z=-2.18, p=0.827, Z=-1.946, p=0.052). The average speech perception score for the NAL-NL2 fitting formula was 69% (SD = ±16) for 45 dB SPL and 80% (SD = ±9) for 55 dB SPL and for the APDB fitting formula this was 70% (SD = ±15) at 45 dB SPL and 85% (SD = ±10) at 55 dB SPL.
CI only scores were 65% (SD = ±12) at 45 dB SPL and 80% (SD = ±12) at 55 dB SPL.

Speech perception in noise

Figure 7 shows the results for the speech perception in noise test. No significant differences between the bimodal performance for the two fitting formulas were found for S0NHA, S0N0, and S0NCI (Wilcoxon Signed Rank test, Z=-0.724, p=0.469, Z=-1.046, p=0.295, Z=-0.322, p=0.748). For the S0N0 condition the results showed an average signal-to-noise ratio (SNR) of 3.2 dB, for the S0NHA an average SNR of 2.4 dB, and for the S0NCI condition an average SNR of 5.0 dB.
For the CI only, the S0N0 condition resulted in an average SNR of 4.6 dB, for the S0NHA this was 3.1 dB and for the S0NCI condition an average SNR of 7.6 dB was found.

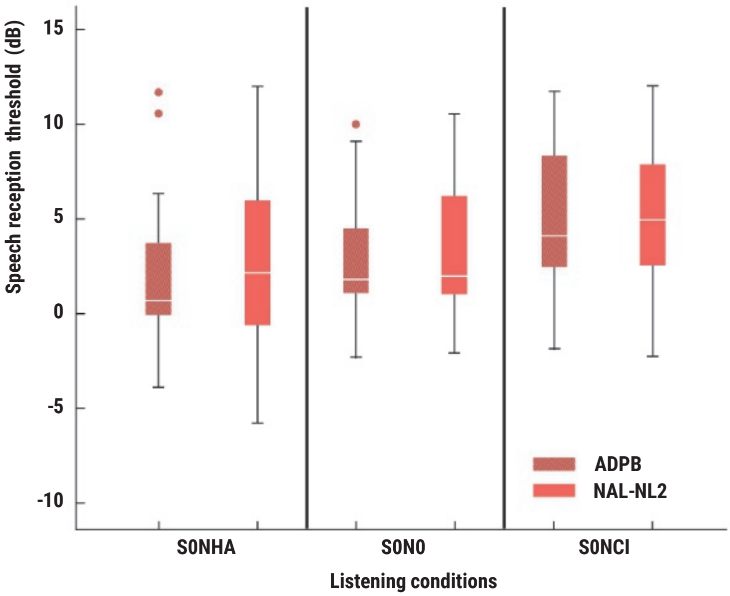


Figure 7 Box-whisker plot of the speech perception in noise performance for the three listening conditions for the two HA fitting formulas. Boxes represent the median (thick horizontal line), lower and upper quartiles (end of boxes), minimum and maximum values (ends of whiskers), outliers (values between 1.5 and 3 times the interquartile range above the third quartile – circles).

SSQ

No significant differences between the two fitting formulas for the different sub parts of the SSQ existed. The average total score for the NAL-NL2 fitting formula was 5.4 and for the APDB fitting formula this was 5.2. Also, when analysing the SSQ data per question, no significant differences were found. The average score for the NAL-NL2 fitting formula for the speech part was 5.1, for the spatial part this was 4.9 and for the quality of sound was 6.1. For the APDB fitting formula the results for the speech domain was 5.1, for the spatial part this was 4.6 and for the quality of sound 5.9 was found.

Bimodal benefit

The results for speech perception in quiet at 45 dB SPL showed a bimodal benefit of 5% for the APDB fitting formula as well as the NAL-NL2 fitting formula (Wilcoxon Signed Rank test, Z=-2.060, p=0.039, Z=-2.287, p=0.02). For speech perception in quiet at 55 dB SPL a significant bimodal benefit of 5% was found for only the APDB fitting formula (Wilcoxon Signed Rank test, Z=-2.354, p=0.02).
For speech perception in noise a bimodal benefit of 1.1 and 1.6 dB was found for the S0N0 condition for the NAL-NL2 and APDB fitting formula respectively (Wilcoxon Signed Rank test, Z=-3.541, p<0.0001, Z=-2.254, p=0.024). The bimodal benefit for the S0NCI condition was 2.6 dB for both fitting formulas (Wilcoxon Signed Rank test, Z=-2.038, p=0.003, Z=-3.139, p=0.002). For the S0NHA condition no bimodal benefit was found.

Auditory functioning and contralateral hearing loss

No consistent significant correlations were found between the contralateral hearing loss and the bimodal auditory performance. For the two groups of participants (with flat or steep hearing losses) bimodal auditory performance was compared for the two fitting formulas. For both groups no significant differences in auditory performance between the two fitting formulas were found (Wilcoxon Signed Rank test).

DISCUSSION

The development of a dedicated fitting formula for bimodal HA fitting, the APDB fitting formula, is potentially promising for bimodal CI users. Although different sub parts of this fitting formula were based on scientific research (Ching et al. 2001b; Neuman et al. 2013; Sheffield et al. 2014; Zhang et al. 2014; Veugen et al. 2016b), the effect of combining these algorithms into one prescriptive fitting formula on bimodal performance had not been subject to study yet. Therefore we investigated the possible benefits of this dedicated bimodal fitting formula on auditory performance. We compared this fitting formula with a standard HA fitting formula, NAL-NL2.

Real Ear Aided Response

The two fitting formulas differed in prescribed gain for frequencies of 2 kHz and above. This is what was expected, because NAL-NL2 provides gain for the whole frequency range whereas the APDB fitting formula applies a reduced frequency bandwidth of the gain if the slope of the hearing loss is more than 35 dB per octave and the high frequency hearing loss exceeds 85 dB HL or if the hearing loss is more than 110 dB HL. Figure 5 confirms that this gain difference is most pronounced for the group of participants with steep hearing losses. The APDB fitting formula is also designed to increase gain at low frequencies to map soft speech (of 55 dB SPL) within the effective audibility range. However, contrary to expectation, we did not find substantial differences at the lower frequencies. A possible explanation could be that for the APDB fitting formula no compensation for the individual ear canal and ear mould was performed. As for the NAL-NL2 fitting formula this compensation resulted in an average gain adjustment of +0 to +5 dB for the different frequency ranges. It can be hypothesized that when taking this compensation into account for the APDB fitting formula, slightly larger differences with the NAL-NL2 would be found. However, as no substantial differences were found when analyzing the data before compensation took place, this cannot be the main explanation for the limited differences. Another characteristic of the APDB fitting formula is to keep the output for speech at almost the same level above input levels of 65 dB SPL, e.g. applying a high amount of compression. From our REAR results we indeed obtained a higher compression ratio for input levels above 65 dB SPL for frequencies of 1000 Hz and above for the APDB fitting formula, although the differences are again a bit smaller than expected.

Loudness balancing did not result in large deviations from the prescribed gain by the initial fitting formula. The mean deviation of 1-2 dB in this study is less than what is generally reported in literature. In a study of Ching et al. (2001a) a 6-dB deviation was found, using NAL-RP as prescription procedure. However, this study was performed with children, for which gain and output requirements are not the same as for adults. In another study of Ching et al. (2004) a deviation from the NAL-NL1 of 3.7 dB was found, which is more comparable with our findings. There is however a substantial difference between NAL-NL2 and NAL-NL1/RP, which may explain the differences. NAL-RP is a linear prescription formula and NAL-NL1 is the first non-linear amplification formula from the National Acoustics Laboratories of Australia. NAL-NL2 is a revision of NAL-NL1 and they differ in a number of ways. NAL-NL1 is a purely theoretically derived formula aimed at maximizing speech intelligibility for any input level of speech while keeping the overall loudness of speech at or below normal loudness. NAL-NL2 is a revision of NAL-NL1 and they differ in a number of ways. The main differences are the use of a more recent loudness model resulting in a less overall loudness compared to NAL-NL1.

And for subjects with profound hearing loss, optimized gains at high and low levels are adjusted so that excessively high compression ratios are not prescribed (Dillon et al. 2012, Keidser et al. 2012). More recently, English et al. (2016) reported more than half of the patients deviating less than 5 dB from the NAL-NL2 fitting formula with a procedure including loudness balancing, which is in line with our results. This means that fitting based on NAL-NL2 generally results in a good loudness balance with the CI. This holds for the APDB fitting formula as well, although the variation in gain adjustments after balancing was slightly larger. Interestingly, in some individuals in our study larger deviations (8-10 dB) were found. In other words, the prescription formulas does not provide equal loudness balance between HA and CI for all bimodal users. This shows that fine-tuning may still be relevant for the individual user. The repeatability of the balanced settings was not tested in our study. To incorporate this in future research will add valuable information for the interpretation of the findings of this study. Unfortunately, we do not know which type of HA and HA fitting formula each participant was used to prior to this study. It can be hypothesized that participants who had been using some type of NAL formula have more difficulty adjusting to the APDB fitting formula. However, in the current study almost every participant adapted easily to the APDB fitting strategy.

Bimodal auditory functioning

For speech in quiet a bimodal benefit of 5-percentage point difference was found for most of the test conditions, regardless of the fitting formula used, consistent with most previous findings (Illg et al. 2014; Dorman et al. 2015). We found a bimodal benefit of 1-1.5 dB for speech in noise in the S0N0 condition (summation effect) and 2.6 dB in the S0NCI condition, again in line with other studies (Ching et al., 2007; Ilgg et al. 2014; Schafer et al. 2007). For the S0NHA condition (squelch effect) we found no bimodal benefit, consistent with what is reported in a meta-analysis by Schafer et al. (2007) who evaluated the squelch effect for 3 studies in a meta-analysis. The authors calculated a mean squelch estimate of 10.1 percentage points, Morera et al. (2012) reported a squelch estimate of 2.6-3.6 dB but questioned generalizability, because these estimates seemed to be driven by the results for two bimodal participants who demonstrated much better speech recognition performance with the HA over the implanted ear. No significant differences were found between the two fitting formulas on bimodal auditory functioning, for speech in quiet as well as for speech in noise. This is not surprising, regarding the relatively small differences in REAR. As they both provide bimodal benefit, there seems no clear general advantage to apply this dedicated bimodal fitting formula compared to a more standard formula as the NAL-NL2. However, individual differences and preferences exist. It could be that for specific patients more fine tuning of the HA settings improves their auditory functioning. However, this is difficult to measure in a controlled research setting. Moreover, our study population was relatively small, so small differences between settings or in auditory performance can be missed. Another aspect that may have had impact on our results is the use of stationary noise for the speech perception in noise test. Some studies suggest (Veugen et al. 2016b, Illg et al. 2014) that bimodal benefit is larger when multi talker babbles are chosen instead of stationary noises as noise stimuli, especially regarding matching of compression between the hearing aid and CI. Future research to this fitting formula should therefore also incorporate this type of stimulus.

CONCLUSION

The results of this study show that CI users with residual hearing at the other ear can benefit from bimodal stimulation, regardless of the fitting method which was applied. Although significant differences between the output and compression ratio of the APDB fitting formula and NAL-NL2 existed, no differences in bimodal auditory performance were observed in quiet as well as in stationary background noise. Therefore, NAL-NL2 or the APDB fitting prescription both seem suited for bimodal fitting purposes. Additional loudness balancing has only a marginal effect on the amount of amplification provided by the hearing aid.

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Participant	Age	Gender	Etiology	HA experience non-implanted ear (years)	CI experience (years)
1	59	M	Unknown	21	5
2	49	F	Unknown	16	6
3	34	F	Familiar	9	4
4	71	M	Familiar	17	1
5	62	F	DFNA9	26	4
6	64	F	Unknown	20	2
7	69	M	Unknown	13	2
8	72	F	Unknown	38	12
9	79	M	Unknown	25	1
10	80	F	Menière	23	2
11	48	M	Familiar	20	0.5
12	76	F	Unknown	16	1
13	48	M	Unknown	18	1
14	74	M	Menière	25	9
15	49	M	Unknown	27	11
16	68	F	Familiar	28	0.5
17	32	M	Unknown	31	4
18	57	M	Unknown	2	1
19	81	M	Unknown	20	2

Table 1 Participant demographics, including details of hearing losses and hearing aid (HA) and cochlear implant (CI) experience

Octave band frequency	125 Hz	250 Hz	500 Hz	1000 Hz	2000 Hz	4000 Hz	8000 Hz
Average adjustment [min max]	2.8 dB [-7 – 15]	3.8 dB [-5 –19]	3.5 dB [-3 – 15]	1.9 dB [-5 – 12]	1.3 dB [-6 – 11]	0.2 dB [-4 – 7]	1.5 dB [-5 – 21]

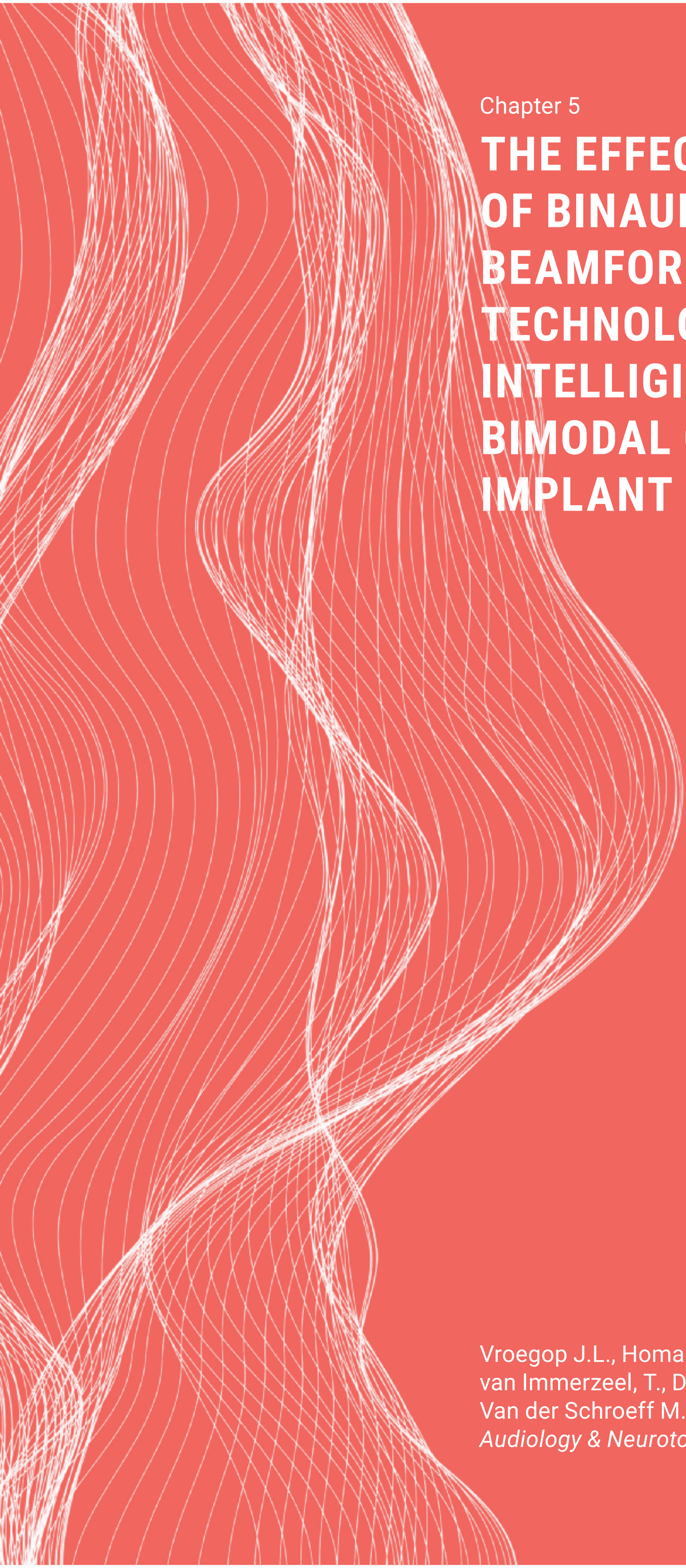
Table 2 Difference between first fit to NAL-NL2 target output values averaged over all patients

Octave band frequency	125 Hz	250 Hz	500 Hz	1000 Hz	2000 Hz	4000 Hz
APBD NAL-NL2	14.3 20.0	3.2 4.0	2.8 2.2	2.0 1.4	1.6 1.0	2.2 1.3

Table 3 Compression ratios for the two fitting formulas and different octave bands.

Als de schemer valt op een warme zomeravond en de kolen nog nagloeien in de barbecue, komen de gezellige kaarslichtgesprekken op gang. Maar als de zon ondergaat, valt het doek voor een slechthorende. Spraakmakende gezichten worden silhouetten en het volgen van een gesprek is tasten in het duister…
Gisteravond deed zich weer zo'n typisch zomers tafereel voor, met als enige verschil: ik verstond ook in het donker wat er werd gezegd! Ik hoefde niet te vragen of die oogverblindende TL-verlichting weer van zolder gehaald kon worden, of dat het hele gezelschap in die broeierige hitte voor mij naar binnen wilde gaan.

FRANCES GALLIMORE,
enkele maanden nadat haar CI was aangesloten



Chapter 5

THE EFFECT OF BINAURAL BEAMFORMING TECHNOLOGY ON SPEECH INTELLIGIBILITY IN BIMODAL COCHLEAR IMPLANT RECIPIENTS

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ABSTRACT

Although the benefit of bimodal listening in cochlear implant users has been agreed on, speech comprehension remains a challenge in acoustically complex real-life environments due to reverberation and disturbing background noises. One way to additionally improve bimodal auditory performance is the use of directional microphones. The objective of this study was to investigate the effect of a binaural beamformer for bimodal CI users. This prospective study measured Speech Reception Thresholds (SRT) in noise in a repeated measures design varying in listening modality for a static and a dynamic listening condition. A significant improvement in SRT of 4.7 dB was found with the binaural beamformer switched on in the bimodal static listening condition. No significant improvement was found in the dynamic listening condition. We conclude that there is a clear additional advantage of the binaural beamformer in bimodal CI users for predictable/static listening conditions with frontal target speech and spatially separated noise sources.

INTRODUCTION

Cochlear implant (CI) selection criteria have expanded (Dowell et al., 2016; Leigh et al., 2016) over the last few years. The use of a cochlear implant in one ear and a hearing aid (HA) in the contralateral ear, which is referred to as bimodal hearing, has become standard care. Bimodal hearing has been shown to improve speech recognition and sound localization compared to unilateral CI use alone (Blamey et al., 2015; Ching et al., 2007; Dorman et al., 2015; Illg et al., 2014; Morera et al., 2012). However, speech comprehension remains a challenge in acoustically complex real-life environments due to reverberation and disturbing background noises (Lenarz et al., 2012; Srinivasan et al., 2013).

Directional microphones aim to improve the signal-to-noise ratio (SNR) by means of enhancing sounds of interest compared to spatially separated interfering sounds (Dillon, 2012). The most recent development of directional HA technology involves wireless communication, which enables the exchange of audio data received by the microphones of both the left and the right HA. The increase in physical separation between the different microphones can be used to achieve narrow beamforming with further SNR improvements (Lotter and Vary, 2006).

However, binaural information is distorted by using this technology. Hearing aid studies investigating binaural beamforming have shown a trade-off between improvement in SNR on one hand, and a deterioration of binaural cues on the other hand (Kidd et al., 2015; Picou et al., 2014). The acoustical conditions play a critical role, as more static and/or predictable listening conditions result in more effect of binaural beamforming compared to more dynamic set-ups (Best et al., 2015; Neher et al., 2017).

Until now, there are no studies evaluating the effect of bilateral beamforming for bimodal CI users. Recently a HA enabling wireless communication was introduced, offering possibilities for a beamforming algorithm for bimodal hearing. As bilateral directional processing for HAs tends to be a tradeoff between SNR improvement and binaural cue preservation, the aim of this study was to investigate íf and in what conditions, usage of a binaural directional microphone algorithm would improve the auditory functioning of bimodal CI users. Two settings were used for testing, reflecting daily life in a static and a more dynamic setting. We hypothesized that an optimal benefit of the binaural beamformer will be found for the static condition and that sub-optimal orientation under dynamic conditions would reduce the benefit obtained from the binaural beamformer.

METHODS

Participants

A total of 18 postlingually deafened adults participated in this study, see table 1 for patient demographics. Participants ranged in age from 32 to 81 years old (group mean age = 62; SD = 15 years). All were experienced bimodal users, unilaterally implanted with AB HiRes 90K implant by surgeons of four different cochlear implant teams in the Netherlands. All participants had used their cochlear implant for at least six months prior to this study (average = 4 years, SD = 3,5 years). All participants used the AB Naida Q70 or AB Naida Q90 sound processor in daily life. In the study all participants used the AB Naida Q90 sound processor to gain access to the bimodal beamforming function ('Stereozoom'). In addition, all had open-set speech recognition of at least 70% correct phonemes at 65 dB SPL on the clinically used Dutch consonant-vowel-con-

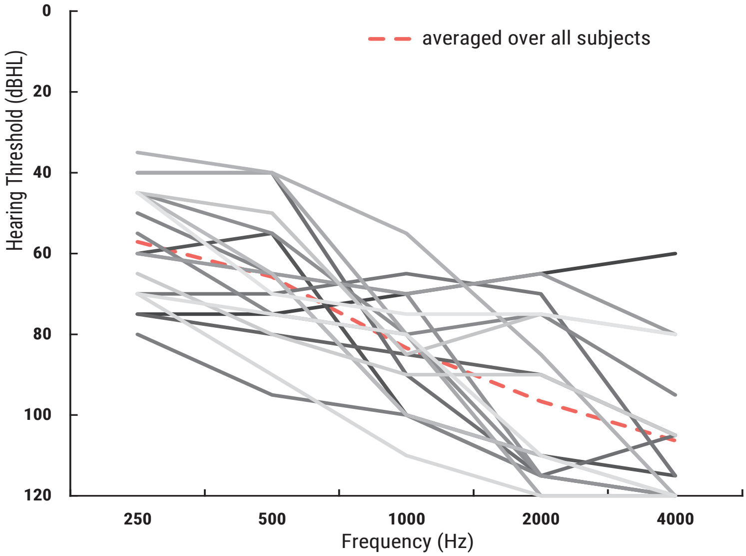

Figure 1 The hearing thresholds of the individual participants for the ear with the hearing aid. The dashed line displays the mean hearing loss.

sonant word lists (Bosman and Smoorenburg, 1995) with the cochlear implant alone. Only participants with unaided hearing thresholds in the non-implanted ear of 80 dB HL or better at 250 Hz were included. Figure 1 shows the unaided audiograms of the non-implanted ear of the individual participants. All participants used a hearing aid prior to the study, which was replaced by the Phonak Naida Link UP HA for the tests in the study. All participants were native Dutch speakers. All participants signed an informed consent letter before participating in the study. Approval of the Ethics Committee of the Erasmus Medical Centre was obtained (protocol number METC306849).

HA and CI fitting

The HA was fitted with the Phonak bimodal fitting formula, a special prescriptive fitting formula for bimodal hearing which was developed for this hearing aid. This formula differs from more standard fitting formulas on three aspects; the frequency response, the loudness growth and the dynamic compression. Firstly, this formula aims to align the frequency response by optimizing low-frequency gain and bandwidth. Low-frequency gain optimization uses the model of effective audibility to ensure audibility of speech recognition in quiet environments (Ching et al., 2001). Frequency bandwidth is optimized, making frequencies between 250 and 750 audible (Sheffield and Gifford, 2014), to maximal width (Neuman and Svirsky, 2013), and amplification does not extend into presumed dead regions (Zhang et al., 2014). Secondly, the loudness growth is aligned by implementing the input-output function of the cochlear implant in the hearing aid. Thirdly, the dynamic compression behavior is aligned by porting the Naída CI dual-loop AGC into the hearing aid (Veugen et al., 2016). The Naida Link HA is able to wireless communicate with the AB Naida CI Q90 and the Q70. With the Q90 the communication is extended to obtain a narrow binaural beamformer, called 'Stereozoom' (Phonak, 2013). This Phonak binaural beamformer combines the four omnidirectional microphones from the Phonak Naida Link HA and the AB Naida CI Q90. First, on each side, the two microphones are processed to obtain a standard dual microphone system. Then these directional signals are exchanged over the wireless link between the HA and the CI. Utilizing a frequency-dependent weighting function, the HA and the CI then linearly combines the ipsilateral and contralateral directional signals to create a bin-

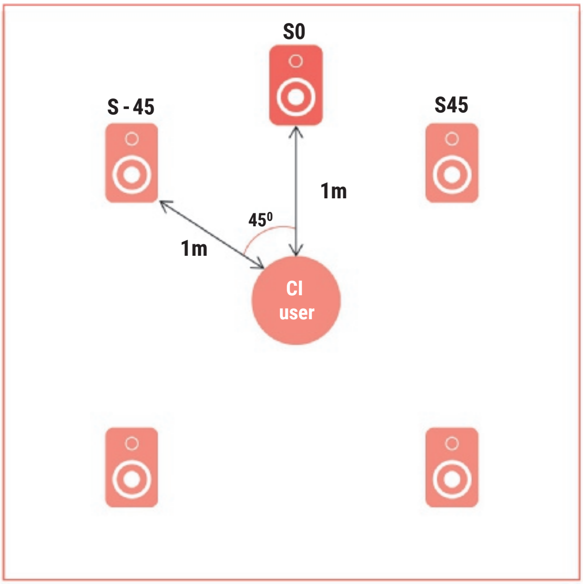

Figure 2 A schematic representation of the test environment. The CI user is in the middle of five loudspeakers, all at a distance of 1 m. The target signal is coming from S0 for the static listening condition and randomly from the loudspeaker at -45^0 or 45^0 for the dynamic listening condition.

aural directivity. The binaural beamwidth is controlled by the weighting function and is typically narrower than what a simple monaural two-microphone beamformer is able to achieve. No fine tuning of the HA or volume adjustments were performed.
For the test session the participant's current 'daily' CI program was used, which was made during clinical programming. The participants were using their current CI program for ten months (SD = 6 months) on average before the start of the study. The methods of CI programming, completed clinically before study participation, was as follows. The upper electrical current levels (M-levels) were set to a most comfortable level for each individual electrode through an ascending loudness judgment procedure. Subsequently, electrodes were checked for equal loudness between them. The minimum current levels (T-levels) were set to threshold levels measured for 0% detection on each individual electrode. Threshold levels were obtained using an ascending presentation, followed by a standard bracketing procedure. After that, the overall level of the M-level profile was adjusted to make live speech sound comfortable and easily understandable. Additional fine tuning of the T- and M-level profiles were applied based on the feedback of the CI user and the professional judgement of the clinical audiologist. Noise reduction algorithms on the cochlear implant (Clearvoice, Windblock, Soundrelax) and HA (Noiseblock, Soundrelax, Windblock) were turned off during the test sessions. Omnidirectional microphone modes were used for condition one to four.

Study design and procedures

This prospective study used a 'within-subjects repeated measures' design. Two factors were used: Listening modality (CI only, bimodal, binaural beamformer), and Speaker location (S0 or S-45/45). The study consisted of one visit in which speech-in-noise tests were performed for six different combinations of factors mentioned above: 1) CI only, S0, 2) CI only, S-45/45, 3) bimodal, S0, 4) bimodal, S-45/45, 5) binaural beamformer, S0, and 6) binaural beamformer, S-45/45), see table 2. The order of the six conditions was randomized to prevent any order effects.

Test environment and materials

Dutch speech material developed at the VU Medical Centre (Versfeld et al., 2000) was used for testing speech recognition in noise. From this speech material, unrelated sentences were selected. A list with 20 sentences were presented at a fixed level of 70 dB SPL for each test condition. This level is representative for a raised voice (Pearsons et al. 1977) in background noise. The sentences were presented in a reception babble noise. We scored the correct words per sentence per list. An adaptive procedure was used to find the signal-to-noise ratio targeting at a score of 50% correct words (Speech Reception Threshold or SRT). For each condition and for each participant a list with 20 sentences was randomly selected from a total of 25 lists. An extensive description of the speech reception in noise test is given in (Dingemanse and Goedegebure, 2015).

For the static condition, sentences were presented from a loudspeaker that was located at 1m at 00 azimuth for conditions 1, 3 and 5. For the dynamic condition, sentences were presented randomly from a loudspeaker at -450 or 450 for conditions 2, 4 and 6 reflecting frequently occurring social situations in which a listener has to understand speech coming from more than one location. Four uncorrelated reception babble noises were presented with four loudspeakers located at -450, 450, -1350 and 1350 azimuth. The rationale for this loudspeaker set-up was to simulate a diffuse, uncorrelated noise that exists in typical noisy daily life situations. Figure 2 displays a schematic of the test environment.

All testing was performed in a sound-attenuated booth. Participants were seated one meter in front of a loudspeaker. For the speech in noise tests, research equipment was used consisting of a Roland UA-1010 soundcard and a fanless Amplicon pc.

Statistical analysis

An ‘a priori Power Analysis’ was performed with a required power of 0.8 and a significance criterion of 0.05, using a Wilcoxon singed-rank test model with G*Power.

For speech perception, we decided to choose a difference of ≥ 15% as clinical significant. With a slope of the psychometric function of 7.5% / dB on average, the difference between two test conditions must be ≥ 2 dB to be clinically significant. We planned paired comparisons between several test conditions. With a minimum of 2dB between groups the effect size dz is 0.71. With these input parameters, the required number of participants is 15.

Data interpretation and analysis were performed with SPSS (v23). Because the low number of participants, non-parametric statistical methods were used. For the speech recognition in noise, the Friedman test was used to compare SRTs over all listening conditions. Afterwards, post hoc comparisons with the Wilcoxon Signed Rank test were performed. We used the Benjamini-Hochberg method to control the false discovery rate for multiple comparisons (Benjamini and Hochberg, 1995).

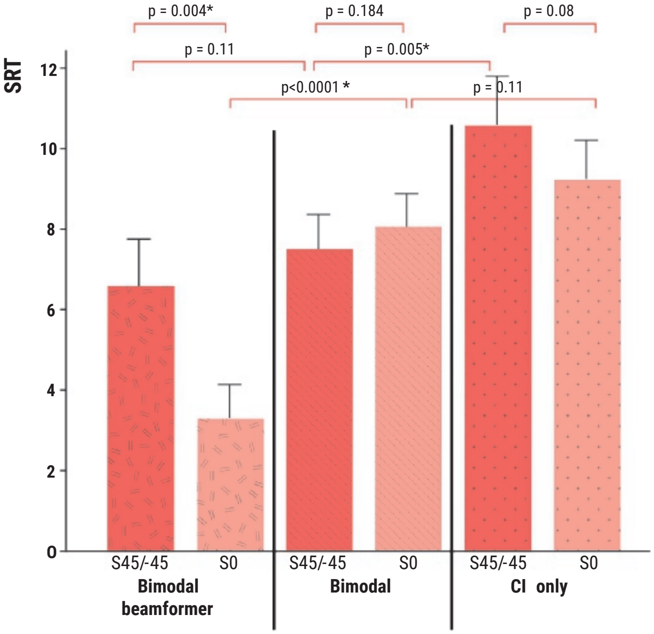

Figure 3 The results of the speech perception in noise test for the six listening conditions. SRT is the speech reception threshold in dB. P-values are corrected for multiple comparisons of Wilcoxon Signed-Rank tests. Asterisks denote significant differences. The error bars represent the standard errors of the mean.

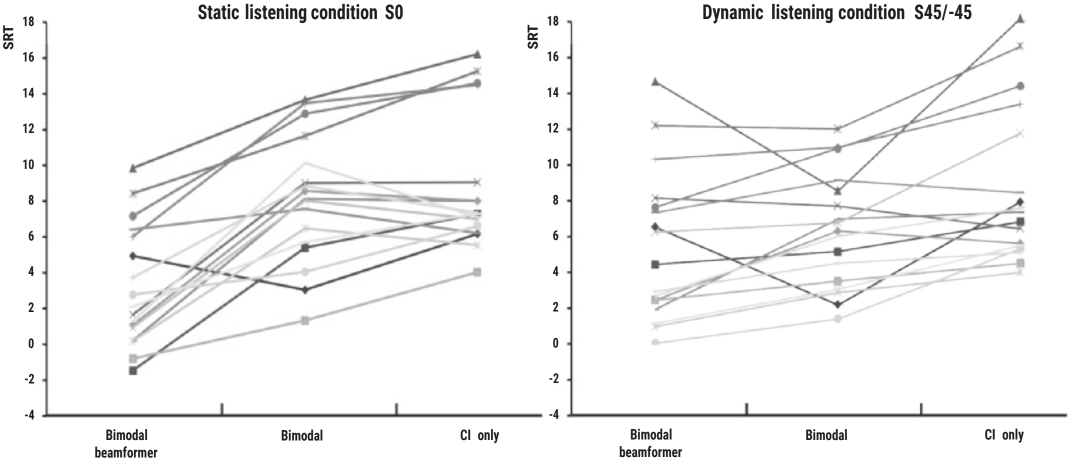

Figure 4 The results of the speech perception in noise test for individual participants for the dynamic as well as the static listening condition. SRT is the speech reception threshold in dB.

RESULTS

The results for the speech recognition in noise test are presented in Figure 3. Significantly different speech reception thresholds were found across the listening conditions (Friedman test: x2(5) = 42.9, p < 0.0001). Post hoc comparisons using the Wilcoxon Signed Rank test showed for the S0 condition no significant difference between the bimodal and CI only condition (Z=-1.76, p=0.11), but a significant improvement was found for the binaural beamformer condition compared with the bimodal condition (4.7 dB, Z=-3.55, p<0.0001). For the S45/-45 condition a significant improvement of the SRT was found for the bimodal condition compared with the CI only condition

(3.1 dB, Z = -3.11, p=0.005), while no significant difference was found between the bimodal condition compared to the binaural beamformer (Z=-1.67, p=0.11). Comparing the results of the two different loudspeaker set-ups (S0 and S45/-45), the binaural beamformer provided a significantly better SRT for the frontal target speech compared to the dynamic speech condition (3.3 dB, Z=-3.20, p=0.004). For the bimodal hearing and CI only condition no difference between the two loudspeaker conditions was found (Z=-1.33, p=0.184 and Z=-1.98, p=0.08 respectively). Reported p-values were corrected for multiple comparisons with the Benjamini-Hochberg method.

Figure 4 shows the SRT scores for the individual participants for the static and the dynamic listening condition. SRT-scores vary largely among participants from 0 to 20 dB, however almost all participants show the same pattern between the listening conditions. Only few participants did not show a benefit for the binaural beamformer condition and for two participants the binaural beamformer deteriorated the SRT for the dynamic and/or static condition.

DISCUSSION

This study showed a statistically significant and clinically relevant benefit of a binaural directional beamforming algorithm for bimodal CI users in term of better speech perception thresholds for the frontal speech target signal. This is in accordance with our hypothesis. Speech was within the spot of the beamformer and the noise sources, coming from other directions, were attenuated. Our results are comparable with the HA only studies investigating the effect of binaural beamformers, were also improvements in SNR were found together with large variability between participants (Best et al., 2015; Kidd et al., 2015; Neher et al., 2017; Picou et al., 2014).

Our results suggest that directionality reduces the localization performance of the participants as no improvement is found in a more dynamic listening condition which is a more demanding task in terms of sound localization. Most probably, the listeners could not localize the sound source optimally as their face was not turned towards the sound source, leaving the target source outside the spot of the beamformer. These results are comparable with the study of Best et al. 2015, who also found reduced SNRs for dynamic speech targets. Also in the study of Picou et al. 2014 a deterioration in localization ability was found.

The bimodal hearing test condition was tested with omnidirectional microphone mode to maximise the localization ability for the dynamic speech target. However, it is possible that with a conventional directional microphone mode in both the CI and the HA separately, a better SNR would have been found, especially for the frontal target signal. Future research with comparisons of more different directional microphone algorithms is needed to provide more information for which situations which algorithm provides the largest benefit for bimodal CI users.

We chose to evaluate the effect of this binaural beamformer with the settings of the HA according to the clinical recommendations of the manufacturer, in order to be able to mimic the daily clinical practice as much as possible. One of these recommendations is the use of the special developed bimodal fitting rule, which we used in this study. However, although all different sub parts of this fitting rule are based on scientific research (Ching et al., 2001; Neuman and Svirsky, 2013; Sheffield and Gifford, 2014; Veugen et al., 2016; Zhang et al., 2014), the effect of the bimodal fitting formula as a whole has not been tested before on auditory functioning. We found a relatively small effect of bimodal hearing compared to the CI only condition. A possible explanation could be that this fitting formula is not the optimal one for all participants. Further investigations to these special developed HA fitting formula and the effect on bimodal hearing are needed. Another limitation of the study is that we only tested the effect of the binaural beamformer in experimental conditions. Future studies should contain field studies to evaluate if the found effect of the beamformer is consistent with the experiences of participants in normal daily life.

CONCLUSION

To conclude, the use of a binaural beamformer for bimodal CI users significantly improves the SNR for frontal target speech. Therefore, application of this binaural beamformer for bimodal users is an effective way to deal with challenging listening conditions, as it optimally uses hearing capacities while enhancing the SNR. However, counseling of the CI users in the function of this binaural beamformer is very important as the user needs to know where the target signal is coming from to be able to obtain the optimal benefit.

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Participant	Age	Gender	Etiology	HA experience non-implanted ear (years)	CI experience (years)
1	59	M	Unknown	21	5
2	49	F	Unknown	16	6
3	34	F	Familiar	9	4
4	71	M	Familiar	17	1
5	62	F	DFNA9	26	4
6	64	F	Unknown	20	2
7	69	M	Unknown	13	2
8	72	F	Unknown	38	12
9	79	M	Unknown	25	1
10	48	M	Familiar	20	0.5
11	76	F	Unknown	16	1
12	48	M	Unknown	18	1
13	74	M	Menière	25	9
14	49	M	Unknown	27	11
15	68	F	Familiar	28	0.5
16	32	M	Unknown	31	4
17	57	M	Unknown	2	1
18	81	M	Unknown	20	2

Table 1 Participant demographics, including HA and CI experience.

Condition number	Listening condition	Speaker location
1	CI only	S0
2	CI only	S-45/45
3	Bimodal	S0
4	Bimodal	S-45/45
5	Bimodal beamformer	S0
6	Bimodal Beamformer	S-45/45

Table 2 Different test conditions

Does sound have rhythm? Does it rise and fall like the ocean? Does sound come and go like wind? MYRON UHLBERG

Chapter 6

EVALUATION OF A WIRELESS REMOTE-MICROPHONE IN BIMODAL COCHLEAR IMPLANT RECIPIENTS

Vroegop J.L., Dingemanse, J.G., Homans, N.C., Goedegebure A.

International Journal of Audiology, 2017, 56(9):643-649

ABSTRACT

Objective To evaluate the benefit of a wireless remote microphone (MM) for speech recognition in noise in bimodal adult cochlear implant (CI) users both in a test setting and in daily life.

Design This prospective study measured Speech Reception Thresholds in noise in a repeated measures design with factors including bimodal hearing and MM use. The participants also had a 3-week trial period at home with the MM.

Study sample Thirteen postlingually deafened adult bimodal CI users.

Results A significant improvement in SRT of 5.4 dB was found between the use of the CI with the MM and the use of the CI without the MM. By also pairing the MM to the hearing aid (HA) another improvement in SRT of 2.2 dB was found compared to the situation with the MM paired to the CI alone. In daily life, participants reported better speech perception for various challenging listening situations, when using the MM in the bimodal condition.

Conclusion There is a clear advantage of bimodal listening (CI and HA) compared to CI alone when applying advanced wireless remote microphone techniques to improve speech understanding in adult bimodal CI users.

INTRODUCTION

Over the past few years, more patients with residual hearing are receiving a cochlear implant. These patients are good candidates for the use of a cochlear implant (CI) in one ear and a hearing aid (HA) in the other ear, which is referred to as bimodal hearing. Bimodal hearing has shown improved speech recognition in quiet and in noise and sound localization compared to unilateral CI use alone (Blamey et al. 2015; Dorman et al. 2015; Illg et al. 2014; Morera et al. 2012). However, in acoustically complex, real-life environments, speech comprehension remains a challenge. In these situations, the presence of reverberation and background noise causes deterioration of understanding a conversation (Lenarz et al. 2012; Srinivasan et al. 2013).

The introduction of directional microphones for CIs has provided a significant improvement in hearing in noise (Hersbach et al. 2012; Spriet et al. 2007). Directional microphones work optimally in near field situations when the sound source is located closely, directed towards the front while the background noise is behind the listener. However, in daily life, full benefit of directional microphones is often not reached, because most listening conditions do not match with the requirements of the directional microphones. The speech source can be at a distance from the CI microphone whereas the background noise more nearby makes the signal-to-noise ratio (SNR) too low for speech understanding, despite the effect of the directional microphone. Furthermore, reverberation can compromise the benefit of the directional microphones. Thirdly, background noise does not only come from behind the listener, but can also be located next to or in front of the listener, which will cause a diminished effect of the directional microphones. Adaptive beamforming has been introduced to address this last limitation, as the direction and shape of the beam can be adjusted dependent of the location of speakers and the background noise (Kreikemeier et al., 2013, Picou et al., 2014).

Another way to improve hearing in demanding listening situations is the use of a wireless remote microphone system. Typically these systems consist of a microphone placed near the speaker’s mouth, which picks up the speech, converts it to an electrical waveform and transmits the signal directly to a receiver worn by the listener with a digital radio frequency (RF) transmission. By acquiring the signal at or near the source, the SNR at the listener’s ear is improved and consequently the negative effects of ambient noise, as well as those of distance and reverberation, are reduced. Previous research has shown considerable improvement in unilateral CI users’ speech recognition in noise using RF systems (de Ceulaer et al. 2016; Schafer et al. 2004, 2009). These studies are laboratory studies with a multiple loudspeaker set up. Noise is coming from behind or next to the subjects. In the study of de Ceulaer et al., a diffuse noise field is created with four loudspeakers with speech coming from three loudspeakers. Bimodal users were instructed to take their hearing aid off during the testing. Improvements of 6 – 14 dB in SNR has been reported in these studies, depending on the test setup.

A new technology for wireless remote microphones based on the 2.4 GHz wireless frequency band has been developed. With this technology, wireless assistive listening devices, like a remote microphone or a streamer for sound from the TV, has been developed by several manufacturers of hearing aids. Cochlear Ltd and Resound Ltd developed the Cochlear Wireless Mini Microphone or Resound Mini Microphone, which is a small personal streaming device microphone for transmitting sound from the microphone or the output from any external audio source directly to a Cochlear sound processor and to a Resound hearing aid. The microphone can be clipped onto the speaker's clothing and provides a wireless link between the speaker and the listener that will potentially improve the signal-to-noise ratio. In a study of Wolfe et al. (2015) a significant improvement in speech recognition in quiet and noise was found for unilateral as well as bilateral CI users when using this wireless microphone. Bimodal users were instructed to take their hearing aid off during the testing.

Recently, Cochlear Ltd. and Resound Ltd. introduced an upgrade of the system, called the Wireless Mini Microphone 2^{+} (Cochlear) or Wireless Multi Microphone (Resound), further on abbreviated as MM. Directional microphones are added to the design and the working range of the MM is extended to 25 meters.

In all adult studies describing the effect of RF systems or the Cochlear Mini Microphone in cochlear implant users, these devices were only connected to the cochlear implants and not to the contralateral hearing aid. The potential extra benefit of enhancing contralateral acoustical hearing by using a remote microphone system was not yet investigated for adult CI users. The objective of this study is, therefore, to evaluate the potential benefit of an advanced remote wireless microphone system with a fixed omnidirectional microphone mode in the bimodal situation with a CI in one ear, hearing aid in the contralateral ear, and the signal of the remote microphone coupled to the CI and hearing aid. We investigated the effect on speech recognition in noise in bimodal adult cochlear implant users both in a test setting and in daily life.

METHODS

Participants

A total of 13 postlingually deafened adults participated in this study. Participants ranged in age from 19 to 83 years old (group mean age = 56; SD = 20 years). All were experienced bimodal users, unilaterally implanted with the Nucleus CI24RE or CI422 implant by surgeons of the Rotterdam Cochlear Implant team at the Erasmus MC hospital in the Netherlands. Only subjects with unaided hearing thresholds in the non-implanted ear better than 75 dB HL at 250 Hz were included. Figure 1 shows the unaided audiograms of the non-implanted ear of the individual participants. All subjects used a hearing aid (Phonak Naida SP or UP) prior to the study, which was replaced with a Resound Enzo 998 hearing aid during the study. This HA was fitted with the NAL-NL2 or Audiogram^{+} (depending on the subjects preference) fitting rule as a first fit. Real ear measurements were used to verify the fitting of the HA. For the real ear measurements an ISTS-signal (Holube et al., 2010) was presented at 55, 65 and 75 dBSPL and gains were adjusted to the fitting rule if needed. The fitting was adjusted afterwards with a loudness balancing procedure to balance the perceived loudness with the CI and the HA. All subjects had used their cochlear implant for at least one year prior to this study (a 1-8 years range, group mean = 4.1 year, SD = 2.1 years), see table 1. All subjects used the Nucleus 6 (CP910) sound processor for at least two months. In addition, all had open-set speech recognition of at least 60% correct phonemes at 65 dB SPL on the clinically used Dutch consonant-vowel-consonant word lists (Bosman & Smoorenburg 1995) with the cochlear implant alone. All participants were native Dutch speakers. All participants signed an informed consent letter before participating in the study. Approval of the Ethics Committee of the Erasmus Medical Centre was obtained (protocol number METC253366).

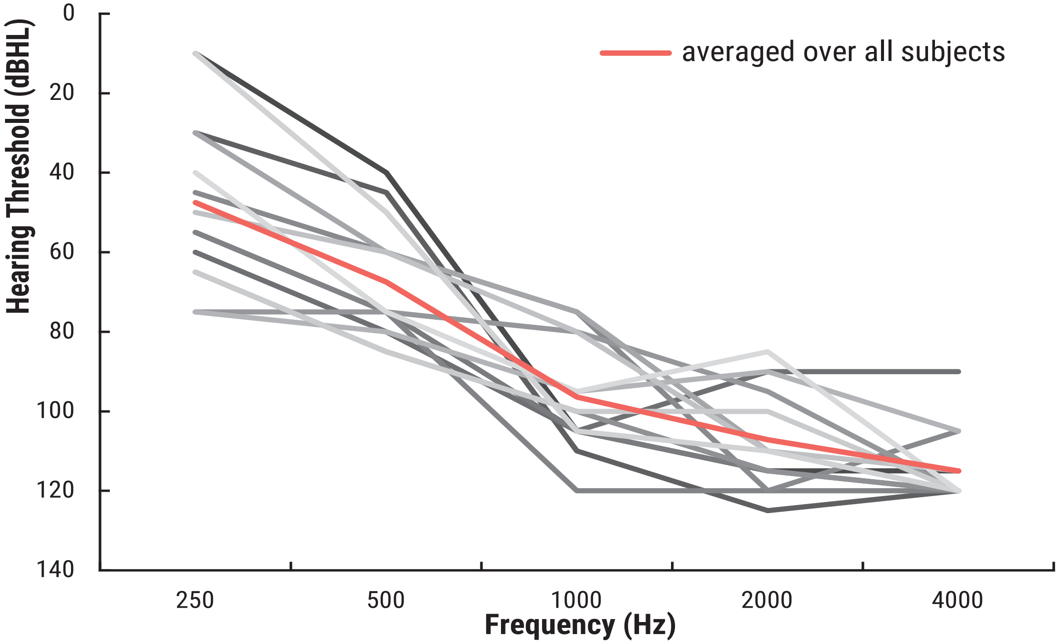

Figure 1 The hearing thresholds of the individual subjects for the ear with the hearing aid. The solid line displays the mean hearing loss.

Study design and procedures

This prospective study used a within-subjects repeated measures design with two factors: Bimodal (yes/no) and MM (yes/no). The study consisted of one visit in which speech-in-noise tests were performed for four different combinations of the two factors: 1) CI only (other ear blocked), no MM, 2) CI and HA, no MM, 3) CI only (other ear blocked), with MM and 4) CI and HA both paired to the MM. The order of the four conditions was randomized to prevent any order effects. Noise reduction algorithms on the cochlear implant (SCAN, SNR-NR and WNR) and HA (SoundShaper, WindGuard and Noise Tracker II) were turned off during the test session in the clinic.
At the end of the test session, subjects received a diary to evaluate the effect of the MM for three weeks in daily life with the MM paired to both CI and HA. During the three weeks evaluation at home, the noise reduction algorithms of both the sound processor (SCAN, SNR-NR and WNR) and the HA (SoundShaper, WindGuard and Noise Tracker II on) were activated to provide optimal hearing in daily life situations of the subjects.

Test environment and materials

Dutch speech material developed at the VU Medical Centre (Versfeld et al. 2000) was used for testing speech recognition in noise. From this speech material, unrelated sentences were selected. A list with 18 sentences were presented at a fixed level of 70 dB SPL for each test condition. This level is representative for a raised voice (Pearsons et al. 1977) in background noise. The sentences were presented in steady state, speech-shaped noise. We scored the correct words per sentence per list. An adaptive procedure was used to find the signal-to-noise ratio targeting at a score of 50% correct words (Speech Reception Threshold or SRT). For each condition and for each subject a list with 18 sentences was randomly selected from a total of 28 lists. An extensive description of the speech reception in noise test is given in Dingemanse and Goedegebure, 2015.

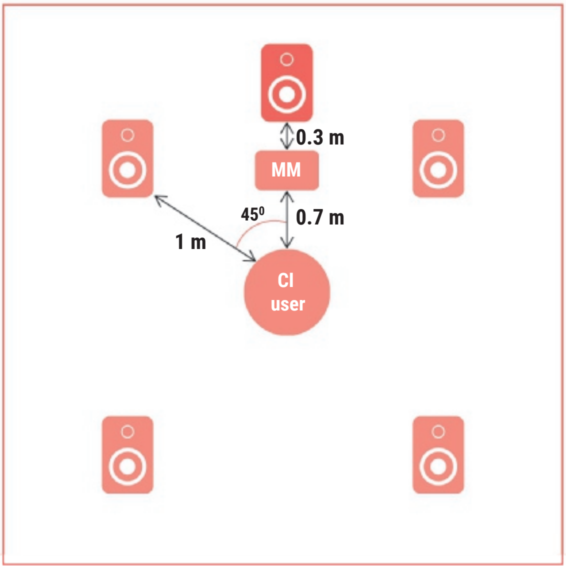

Figure 2 A schematic representation of the test environment. The CI user is in the middle of five loudspeakers, all at a distance of 1 m. The MM is placed at 0.3 m from the loudspeaker with the speech material, the other four loudspeakers presented uncorrelated speech-shaped noises.

Sentences were presented from a loudspeaker that was located at 1m at 00 azimuth. Four uncorrelated speech-shaped noises were presented with four loudspeakers located at -450, 450, -1350 and 1350 azimuth. The rationale for this loudspeaker arrangement was to simulate a diffuse, uncorrelated noise that exists in typical noisy daily life situations. During testing, the MM was positioned in horizontal direction (in omnidirectional mode) at 30 cm from the centre of the cone of the loudspeaker used to present the sentences. Because of the radiation pattern of the loudspeaker in the vertical plane we decided to place the MM no closer than 30 cm to the loudspeaker. At this distance from the loudspeaker the sound level was 77.5 dBSPL, meaning a better SNR of 7.5 dB compared to the place of the subject.. Figure 2 displays a schematic of the test environment.

All testing was performed in a sound-attenuated booth. Participants sat one meter in front of a loudspeaker. For the speech in noise tests, research equipment was used consisting of a Madsen OB822 audiometer, a Behringer UCA202 soundcard, and a Macbook pro notebook.

The subjects received a diary to evaluate the effect of the MM in daily use for different listening situations. Subjects were asked to indicate on a visual analog scale (VAS) if the MM reduced or enhanced their speech recognition in a particular situation. The scale ranges from -5 to +5, where -5 indicates ‘much worse’ and +5 indicates ‘much better’, comparing the condition with the MM to the condition without the MM. The midpoint of the scale (0) indicates that the participant experienced no changes. Subjects were asked to evaluate this for six listening situations, including: a conversation with one person with and without background noise, a group conversation with and without background noise, speech from over a distance and listening to a smartphone or tablet.

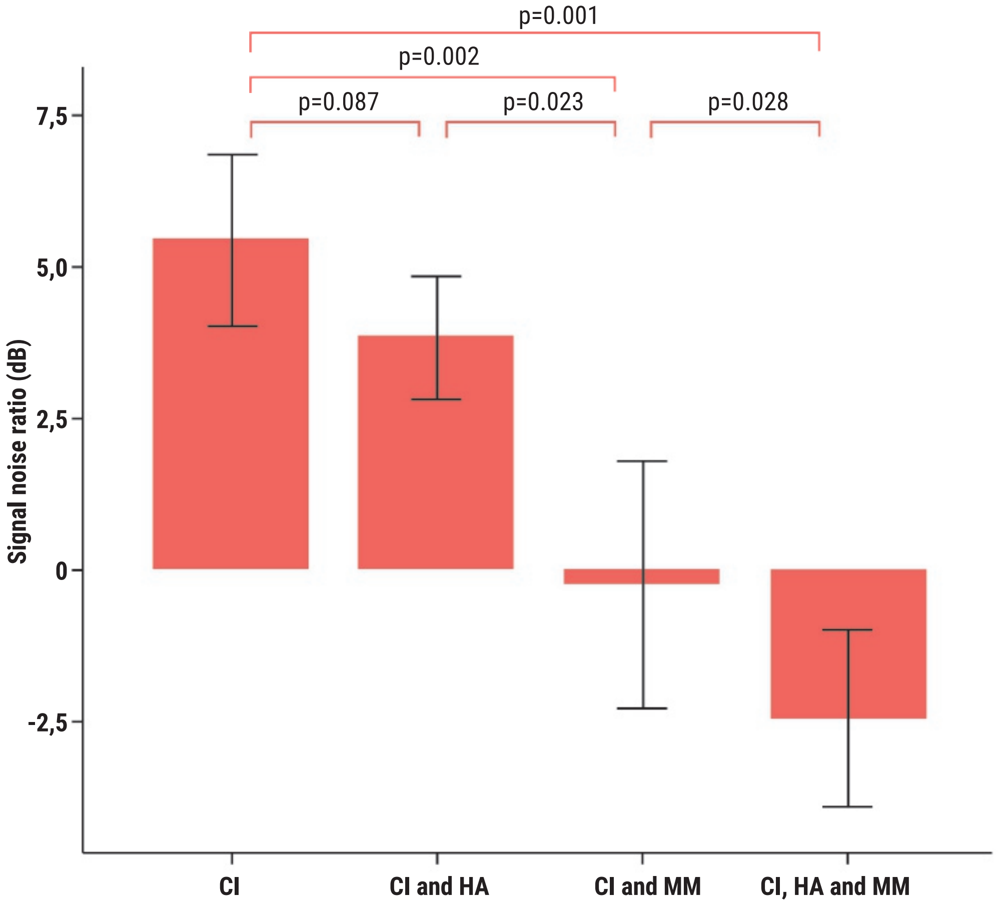

Figure 3 The results of the speech perception in noise test for the four listening conditions. P-values are uncorrected p-values of Wilcoxon Signed-Rank tests. Asterisks denote significant differences after correction for multiple comparisons. The error bars represent the standard errors of the mean.

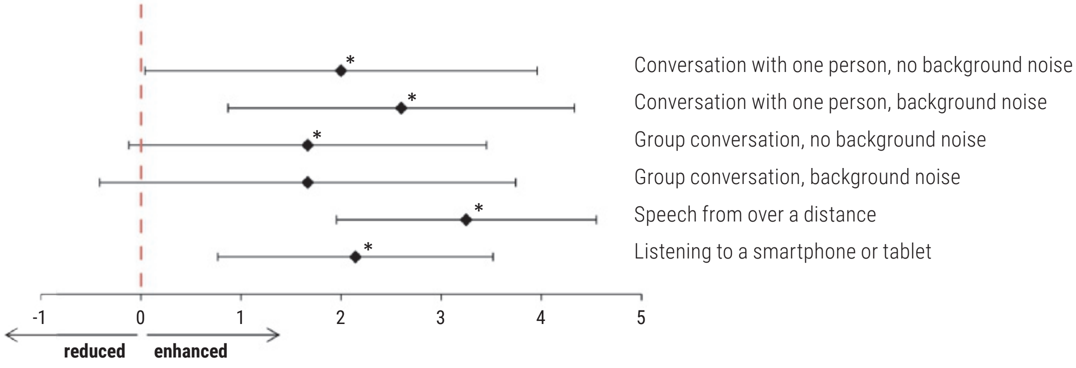

Figure 4 The results for the diary for the six different listening situations. Asterisks denote significant differences. The error bars represent the standard errors of the mean.

Statistical analysis

Data interpretation and analysis were performed with SPSS (v23). Because the low number of subjects, non-parametric statistical methods were used. For the speech recognition in noise, the Friedman test was used to compare SRTs over all listening conditions. Afterwards, post hoc comparisons with the Wilcoxon Signed Rank test were performed. We used the Benjamini-Hochberg method to control the false discovery rate for multiple comparisons (Benjamini and Hochberg, 1995). To analyse the diary data, the one-sample Wilcoxon Signed Rank test was used.

RESULTS

Speech recognition in noise

The results for the speech recognition in noise test are presented in Figure 3. Significantly different speech reception thresholds were found across the listening devices (Friedman test: x2(3) = 27.4, $p < 0.001$). Post hoc comparisons using the Wilcoxon Signed Rank test indicated that a significant difference in SRT of 5.4 dB was found between the use of the MM with the CI and the use of the CI without the MM (Z = -3.11, p=0.002, Y=-0.86). By also pairing the MM to the HA, another improvement in SRT of 2.2 dB was found (Wilcoxon Signed Rank test, Z=-2.20, p=0.028, Y=-0.61). Reported p-values were not corrected for multiple comparisons. After correcting for multiple comparisons with the Benjamini-Hochberg method, these differences remained significant. No correlation was found between the amount of hearing loss and the benefit of the MM.

Additionally, we compared the benefit of bimodal hearing between the two conditions without MM and with MM ($\text{SRT}_{\text{CI and H}}$A - SRT_{CI} versus $\text{SRT}_{\text{CI and HA and MM}} - \text{SRT}_{\text{CI and MM}}$). No significant difference was found, so the benefit of bimodal hearing remains intact when using MM.

Results of the diary

Ten subjects completed the diary to evaluate the use of the MM. The results of the MM diary are presented in Figure 4. A significant improvement of the use of the MM was found for the conversation with one person (both with and without background noise), the group conversation without background noise, the speech from over a distance and listening to a smartphone or tablet (one-sample Wilcoxon Signed Rank test, p=0.02, p=0.01, p=0.02, p=0.01, p=0.03 respectively). For the group conversation with background noise, an improvement was found of borderline statistical significance (one-sample Wilcoxon Signed Rank test, p=0.05). Examples of different places and situations where participants used the MM are shown in table 2. To examine if there was a difference in SRT score between the ten subjects who used the diary and the three subjects who did not use the diary, we used a Mann-Whitney test on the SRT for CI and MM. This test indicated that the groups did not have a significantly different SRT score (U=9, p=0.31, r=0.28).

DISCUSSION

This study showed a large statistically significant and clinically relevant benefit of an advanced remote wireless microphone system that is connected to a CI in one ear and a hearing aid in the contralateral ear. This large improvement in performance for speech perception in noise is the combined effect of the two factors that we investigated: the effect of the MM, and the effect of the bimodal connection of the MM. The effect of the MM explained the largest part of the improvement and is a known effect. At the location of the MM the speech had a higher level giving a better speech-to-noise ratio of the signal that is transmitted to the CI and HA. In our setup, the SNR at the position of the MM was 7.5 dB better compared to the position of the listener. The SNR improvement due to the MM is 5.7 dB for the CI only condition and 6.3 dB for the bimodal condition , which is relatively close to this maximum value.

In our study, we found an improvement of 1.6 – 2.2 dB due to bimodal hearing. This is comparable to what was reported by Ching et al. (2007) in a review about bimodal hearing. They described an improvement which ranges from 1-2 dB across all reviewed studies.

An interesting finding is that the bimodal connection of the MM gave an additional improvement over the connection to the CI alone. With this MM connected to both hearing devices, both devices received the same input signal. This input signal was processed independently by the hearing aid (acoustical and cochlear processing) and the CI (purely electrical processing), resulting into two different patterns of auditory nerve stimulation at each ear, providing both similar and complementary information to the central auditory system. In central auditory processing these differences and similarities in auditory information were used for better speech intelligibility in noise, giving the complementarity effect and the binaural redundancy (Ching et al. 2007).

This is the first study to evaluate the performance of the MM for speech recognition in bimodal adult CI users. The only previous study with the previous version of the MM, the Cochlear Mini Microphone focused on the use of this microphone connected to the CI alone (Wolfe et al. 2015) for unilateral as well as bilateral CI users. Wolfe et al. also found a significant improvement in speech recognition in noise, but they measured improvement of word scores for different fixed SNRs, making a comparison between our results and their findings difficult. Possible differences between the effect of the Mini Microphone for unilateral or bimodal CI users were not investigated in that study.

The average SRT with the MM is –2.5 dB for the bimodal condition in our study sample. With the used speech material, the SRT for normal-hearing subjects is -7 dB. Even with the use of a MM, the CI users preformed less than normal hearing subjects. However, in our study the distance of the MM to the speech source was 30 cm. To improve the SNR further it is important to make this distance shorter by using the MM in daily life. A distance of 15 cm is the clinical recommendation, and this may give an additional improvement up to 6 dB compared to our setup, bringing the SRT close to that of normal hearing subjects. In this study the MM is used in omnidirectional mode. It is expected that by using the directional mode of the MM, even a better SRT could be obtained. All testing was completed in a sound-attenuated booth, which is a limitation of the study. Performance and benefits with the MM will probably be greater when tested in a sound booth rather than a real world environment, because of the greater reverberation in the latter.

The customized diaries of the subjects showed perceived improvement due to the MM for all reported listening situations but one. For group conversations with background noise, no significant benefit of the MM was found. This is probably because in such situations the microphone is placed in the middle of the group. Because of the increased distance of the speakers to the MM the SNR will decrease, whereby speech perception, even with the MM, will become difficult. This is comparable with the results of de Ceulaer et al. (2016) who used a multiple talker network test set up with three speech sources to simulate a group conversation. They found only a limited improvement in SRT when using one Phonak Roger Pen, but a considerable improvement by using three Roger Pens.

The results of the diary also showed that the MM can be used easily in a lot of different places. Only ten out of thirteen subjects used the diary. It can be hypothesized that mainly the participants who perceived benefit from the MM used the diary. However, in the speech test situation in the booth no difference between the subjects who used the diary and the subjects who did not was found.

This study has its limitations. First, the study sample is relatively small. Subsequently, all participants were evaluated while using one model of sound processor, hearing aid and wireless remote microphone. These results may differ for other types of sound processors, hearing aids or remote microphones.

CONCLUSION

To conclude, the use of the MM in combination with the Nucleus 6 sound processor and the Resound Enzo 998 hearing aid provided significantly better sentence recognition in noise than what was obtained without the use of the MM. Furthermore, the use of the MM in bimodal situation provides additional benefit compared to MM use with the CI alone. Also participants reported significantly better speech perception in daily life for different listening situations. Therefore, application of advanced wireless remote systems in bimodal users is an effective way to deal with challenging listening conditions, as it optimally uses bimodal hearing capacities while enhancing the SNR.

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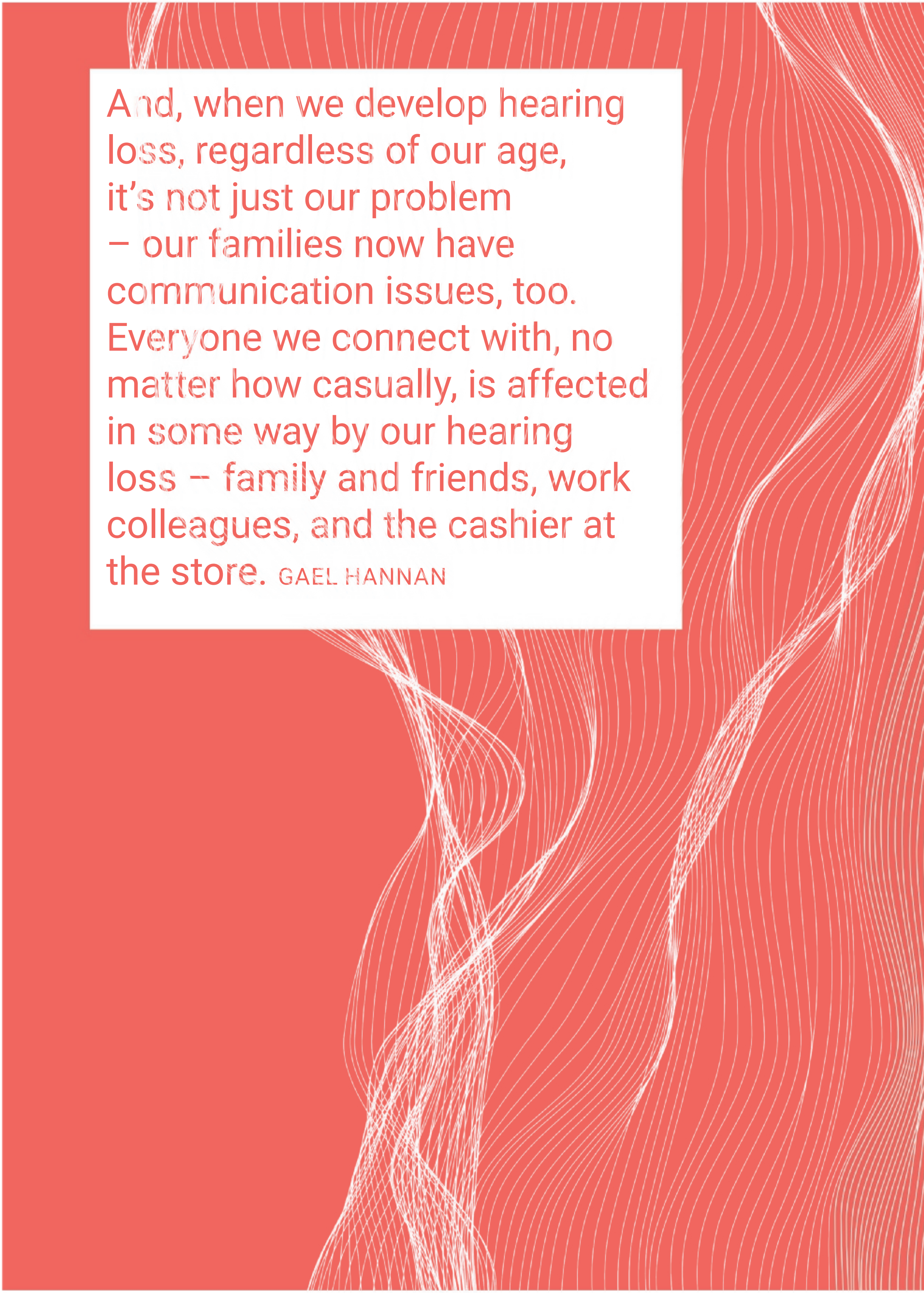
Participant	Age	Gender	Implanted ear	Etiology	HA experience non-implanted ear (years)	CI experience (years)
1	52	F	L	Ototoxicity	8	4
2	68	F	R	Familiar	17	1
3	34	F	L	Unknown	29	5
4	20	F	R	Genetic	19	8
5	83	M	L	Unknown	29	5
6	80	F	L	Familiar	30	2
7	50	M	L	Congenital	45	5
8	71	M	R	Unknown	26	2
9	26	F	R	Unknown	23	6
10	54	F	L	Unknown	4	2
11	69	M	R	Unknown	26	3
12	58	M	L	Familiar	46	6
13	64	F	L	Unknown	28	4

Table 1 Participant demographics, including details of hearing losses and HA and CI experience.

Listening situation	Place
Conversation with one person, no background noise	Dinner at home
	Next to each other at the sofa (without speech reading)
	Conversation in two separate rooms
Conversation with one person, with background noise	Conversation with television as background noise
	Conversation in the car
	Conversation in the train
Group conversation, no background noise	Dinner with more than four persons
	Chatting with friends
	Meeting with ten persons
Group conversation, with background noise	Restaurant with more than four persons
	Conversation in a car with more than three persons
	Group conversation with eight persons
Speech from over a distance	Television
	During lectures
	Conversation in a garden
	Supermarket

Table 2 Different places were the MM is used during the evaluation at home for the different listening situations.



And, when we develop hearing loss, regardless of our age, it’s not just our problem – our families now have communication issues, too. Everyone we connect with, no matter how casually, is affected in some way by our hearing loss – family and friends, work colleagues, and the cashier at the store. GAEL HANNAN

Chapter 7

A DIRECTIONAL REMOTE-MICROPHONE FOR BIMODAL COCHLEAR IMPLANT RECIPIENTS

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ABSTRACT

Objective To evaluate whether speech recognition in noise differs according to whether a wireless remote microphone is connected to just the cochlear implant (CI) or to both the CI and to the hearing aid (HA) in bimodal CI users. The second aim was to evaluate the additional benefit of the directional microphone mode compared with the omnidirectional microphone mode of the wireless microphone.

Design This prospective study measured Speech Recognition Thresholds (SRT) in babble noise in a 'within-subjects repeated measures design' for different listening conditions. Study sample Eighteen postlingually deafened adult bimodal CI users.

Results No difference in speech recognition in noise in the bimodal listening condition was found between the wireless microphone connected to the CI only and to both the CI and the HA. An improvement of 4.1 dB was found for switching from the omnidirectional microphone mode to the directional mode in the CI only condition.

Conclusions The use of a wireless microphone improved speech recognition in noise for bimodal CI users. The use of the directional microphone mode led to a substantial additional improvement of speech perception in noise for situations with one target signal.

INTRODUCTION

One of the main challenges for patients with cochlear implants (CI) is speech comprehension in acoustically complex real-life environments due to reverberation and disturbing background noises (Srinivasan et al., 2013, Lenarz et al., 2012). Data logs of CI processors of 1000 adult CI users showed that many CI users spent large parts of their day in noisy environments, on average more than four hours a day (Busch et al., 2017). Although speech perception in quiet is generally good, the remaining impairment in difficult listening situations can limit quality of life, professional development and social participation (Gygi and Hall, 2016, Ng and Loke, 2015).

The introduction of directional microphones for CIs has provided a significant improvement in hearing in noise abilities (Hersbach et al., 2012, Spriet et al., 2007), however their use is often limited as they require near field situations where the sound source is located close by, directed towards the front while the background noise is behind the listener.

Another way to improve hearing in demanding listening situations is the use of a wireless remote microphone system. Typically these systems consist of a microphone placed near the speaker's mouth, which picks up the speech, converts it to an electrical waveform and transmits the signal directly to a receiver worn by the listener. By acquiring the signal at or near the source, the signal-to-noise ratio (SNR) at the listener's ear is improved and consequently the negative effects of ambient noise, as well as those of distance and reverberation, are reduced. Previous research has shown considerable improvement in unilateral CI users' speech recognition in noise (De Ceulaer et al., 2016, Schafer and Thibodeau, 2004, Schafer et al., 2009, Wolfe et al., 2015a, Wolfe et al., 2015b, Vroegop et al., 2017, Razza et al., 2017). A directional mode of the wireless microphone would possibly improve speech recognition even more, but specific, comparative data is lacking. The previously mentioned studies used either an omnidirectional mode of the wireless microphone (Wolfe et al., 2015b, Vroegop et al., 2017, Razza et al., 2017), an adaptive mode changing between omnidirectional and directional depending on the amount of the background noise (De Ceulaer et al., 2016), or a directional mode (Schafer et al., 2009).

Due to expanding CI selection criteria (Leigh et al., 2016, Dowell et al., 2016) the use of a cochlear implant in one ear and a hearing aid (HA) in the contralateral ear, which is referred to as bimodal hearing, has become standard care. Bimodal hearing has been shown to improve speech recognition in noise compared to unilateral CI use alone (Ching et al., 2007b, Illg et al., 2014, Morera et al., 2012, Blamey et al., 2015, Dorman et al., 2015). Only one study (Vroegop et al., 2017) described the combined effect of a wireless microphone and bimodal hearing. Their results showed that the use of the wireless microphone in the bimodal situation, connected to both the CI and the HA, provided additional benefit compared to the use of the wireless microphone with the CI alone. However, the study failed to differentiate the benefit found between the result of wireless microphone use or just the addition of the HA. Therefore in our current study we investigate whether speech recognition in noise differs according to whether the wireless microphone is connected to just the CI or to both the CI and HA. The second aim in our study was to evaluate the effect of a directional microphone mode of the wireless microphone.

METHODS

Participants

A total of 18 postlingually deafened adults participated in this study, see table 1 for participant demographics. Participants ranged in age from 32 to 81 years old (group mean age = 62; SD = 15 years). All were experienced bimodal users, unilaterally implanted with AB HiRes 90K implant by surgeons of five different cochlear implant teams in the Netherlands and Belgium. All participants had used their cochlear implant for at least

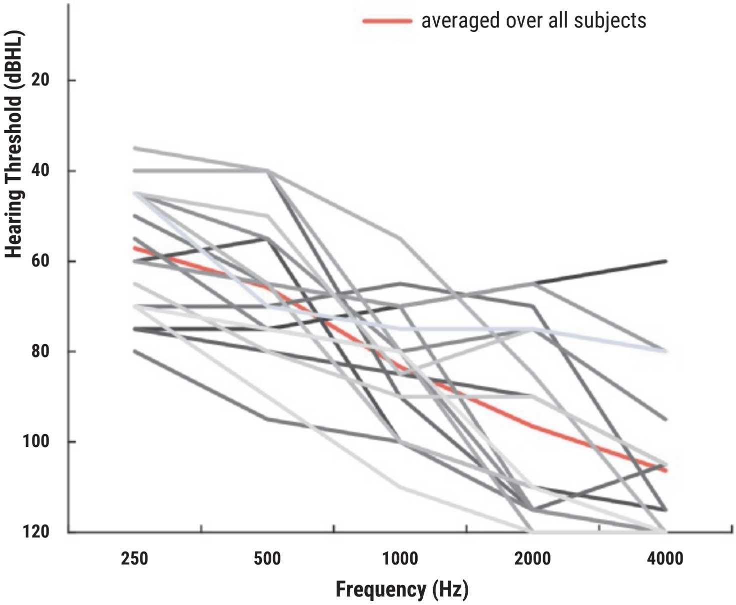

Figure 1 The hearing thresholds of the individual participants for the ear with the hearing aid. The dashed line displays the mean hearing loss averaged over all participants.

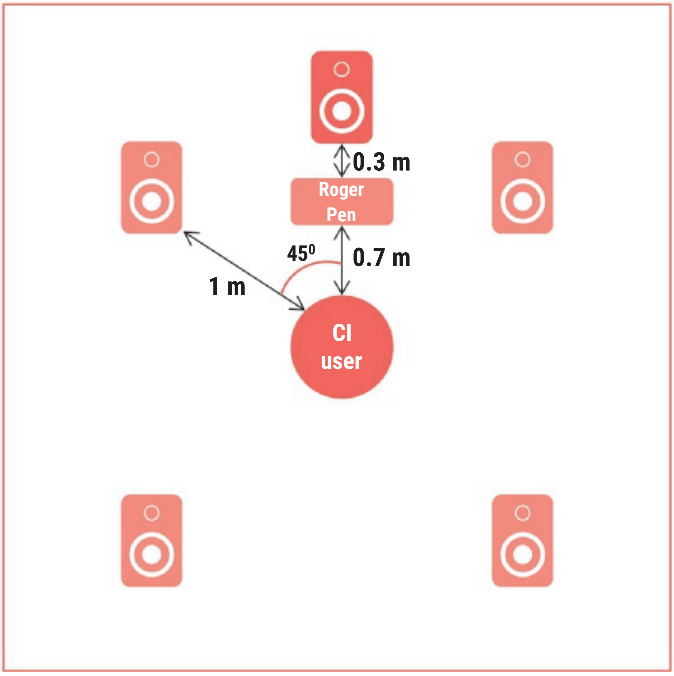

Figure 2 A schematic representation of the test environment. The CI user is in the middle of five loudspeakers, all at a distance of 1 m. The target signal is coming from S0. The Roger Pen is placed at 30 cm from the target signal.

six months prior to this study (average = 4 years, SD = 3.6 years). All participants used either the AB Naida Q70 or AB Naida Q90 sound processor. In the study all participants were provided with a new AB Naida Q90 sound processor. In addition, all had open-set speech recognition of at least 70% correct phonemes at 65 dB SPL on the clinically used Dutch consonant-vowel-consonant word lists (Bosman and Smoorenburg, 1995) with the cochlear implant alone. Only participants with unaided hearing thresholds in the non-implanted ear of 80 dB HL or better at 250 Hz were included. Figure 1 shows the unaided audiograms of the non-implanted ear of the individual participants. All participants used a hearing aid prior to the study, which was replaced by a new Phonak Naida Link UP HA for the tests in the study. For the test conditions with the wireless microphone, the Phonak Roger Pen was used. Integrated Roger 10 and Roger 17 receivers were used for connection with the HA and CI respectively. All participants were native Dutch speakers. All participants signed an informed consent letter before participating in the study. Approval of the Ethics Committee of the Erasmus Medical Centre was obtained (protocol number METC306849).

HA, CI and wireless microphone settings

The HA was fitted with the Phonak bimodal fitting formula, a special prescriptive fitting formula for bimodal hearing which was developed for the Phonak Naida Link hearing aid. This formula differs from more standard fitting formulas in three respects: the frequency response, the loudness growth and the dynamic compression. Firstly, this formula focusses on the frequency response by optimizing low-frequency gain and optimizing frequency bandwidth. Low-frequency gain optimization uses the model of effective audibility to ensure audibility of speech recognition in quiet environments (Ching et al., 2001). Frequency bandwidth is optimized by assuring that bandwidth is as wide as possible, based on a study of Neuman and Svirsky, 2013. In that study the effect of frequency bandwidth on bimodal auditory performance was investigated. They found that smaller frequency bandwidths of the HA resulted in a worse performance of subjects in the bimodal condition compared with a wider frequency bandwidth of the HA. Besides, the fitting formula ensures frequencies between 250 and 750 Hz are audible (Sheffield and Gifford, 2014), and that amplification does not extend into presumed dead regions (Zhang et al., 2014). To obtain the latter, a reduced frequency bandwidth of the gain is applied if the slope of the hearing loss is more than 35 dB per octave and the high frequency hearing loss exceeds 85 dB HL or if the hearing loss is more than 110 dB HL. Secondly, the input-output function of the CI is implemented in the hearing aid (compression kneepoint = 63 dB SPL, compression ratio = 12:1) with the aim of improving loudness balance between HA and CI. Thirdly, the dynamic compression behavior is aligned by integrating the Naída CI dual-loop AGC into the hearing aid (Veugen et al., 2016). No fine tuning of the HA or volume adjustments was performed.

For the test session the participant’s current ‘daily’ CI program was used. The Phonak Roger Pen is part of the Phonak Roger system. It uses digital signal transmission and digital signal processing to feature an adaptive gain adjustment. For the condition with the Roger Pen the CI program was modified by changing the signal input from 100% microphone input to a 70:30 mix of the Roger 17 aux input and the participant’s microphone respectively. Wolfe and Schafer (2008) advised a mixing ratio of 50:50 mainly based on soft speech in quiet surroundings. In the study of De Ceulaer et al. (2016) also a mixing ratio of 50:50 was used. However, they did not find benefit of using one Roger Pen in a diffuse noise field. In our study we investigated the effect of the Roger Pen

for different listening conditions in background noises. We wanted a condition for the Roger Pen in which the effect of the Roger Pen was expected to be found, otherwise we would not be able to distinguish between listening conditions. A 100% Roger condition would presumably result in the largest effect, however CI users generally do not prefer this in daily life, because they are disconnected from the surrounding sounds. Therefore we choose to use the 70:30 condition, in which 70% of the signal comes from the Roger Pen and 30% comes from the participant's CI processor microphone.
Noise reduction algorithms on the cochlear implant (Clearvoice, Windblock, Soundrelax) and HA (Noiseblock, Soundrelax, Windblock) were turned off during the test sessions. In the listening conditions omnidirectional microphone modes of both the CI and the HA were used.

Study design

This prospective study used a 'within-subjects repeated measures' design. The study consisted of one visit in which speech-in-noise tests were performed for six different listening conditions: 1) CI and HA, 2) CI and HA, Roger Pen paired to CI 3) CI and HA, Roger Pen paired to both, 4) CI only, 5) CI and Roger Pen Omni Directional, 6) CI and Roger Pen Directional, see table 2. The order of the six conditions was randomized to prevent any order effects.

Test environment and materials

Dutch speech material, single talker, female voice, developed at the VU Medical Centre (Versfeld et al., 2000) was used for testing speech recognition in noise. From this speech material, unrelated sentences were selected. A list of 20 sentences was presented at a fixed level of 70 dB SPL for each test condition. This level is representative for a raised voice (Pearsons et al., 1977) in background noise. The first list of sentences was used for exercise and adaptation to the test. The sentences were presented in a reception babble noise with an average spectrum similar to the international long-term average speech spectrum (ILTASS). We scored the correct words per sentence. An adaptive procedure was used to find the signal-to-noise ratio targeting a score of 50% correct words (Speech Recognition Threshold or SRT). For each condition and for each participant a list with 20 sentences was randomly selected from a total of 25 lists without replacement. The adaptive procedure used was a stochastic approximation method with step size 4 ⌧ (Pc(n − 1) − target_Pc) (Robbins & Monro, 1951), with Pc(n − 1) being the percent correct score of the previous trial. An extensive description of the speech recogniton in noise test is given in (Dingemanse and Goedegebure, 2015). Sentences were presented from a loudspeaker that was located at 1m at 00 azimuth. Four uncorrelated reception babble noises were presented with four loudspeakers located at -450, 450, -1350 and 1350 azimuth, placed at 1m from the listener as well. The rationale for this loudspeaker set-up was to simulate a diffuse, uncorrelated noise that exists in typical noisy daily life situations. During testing, the Roger Pen was positioned in horizontal direction (in omnidirectional mode) at 30 cm from the centre of the cone of the loudspeaker used to present the sentences. Because of the radiation pattern of the loudspeaker in the vertical plane we decided to place the Roger Pen no closer than 30 cm to the loudspeaker. At this distance from the loudspeaker the sound level was 77.5 dBSPL, meaning a better SNR of 7.5 dB compared to the place of the participant. Figure 2 displays a schematic of the test environment.

All testing was performed in a sound-attenuated booth. For the speech in noise tests, research equipment was used consisting of a Roland UA-1010 soundcard and a fanless Amplicon pc.

Statistical analysis

A Power Analysis was computed using the G*Power software, with a required power of 0.8 and an alpha-error level of 0.05. For speech perception we choose a difference of ≥ 15% as relevant, which is the least possible difference which is measurable with this test in one person. With a slope of the psychometric function of 6.4% / dB on average (Dingemanse and Goedegebure, 2015), the difference between two test conditions must be ≥ 3 dB to be significant. We planned a repeated measures ANOVA. The sample size calculation indicated that a sample of 12 subjects would be needed to detect a significant difference. Because we found 18 suitable participants willing to participate, we added them to the sample.
Data analysis were performed with SPSS (v23). For the speech recognition in noise, the repeated measures ANOVA was used. Afterwards, post hoc testing using Bonferroni correction was performed. We used the Benjamini-Hochberg method to control the false discovery rate for multiple comparisons. This method controls the expected proportion of falsely rejected hypotheses and is described in the study of Benjamini and Hochberg, 1995.

RESULTS

In figure 3 the results for the speech recognition in noise test are displayed. A normality check of the SRTs revealed normally distributed data for all listening conditions. A repeated measures ANOVA determined that the speech recognition threshold differed significantly between listening conditions (F(5, 85) = 17.923, P < 0.0001).
Paired wise comparisons between the bimodal listening condition in which he Roger Pen was paired to the CI only (SRT = 6.5 dB) and the condition with the Roger

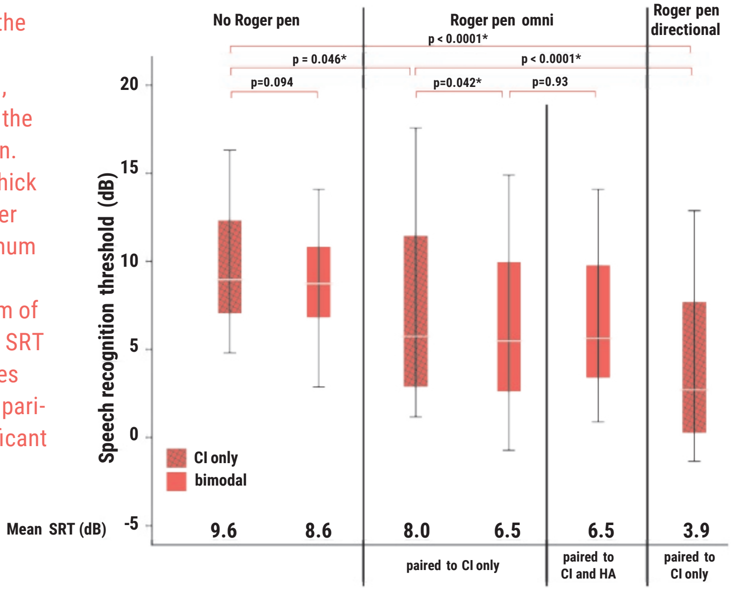

Figure 3 Box-whisker plots of the speech recognition threshold for the no Roger Pen condition, Roger pen omni condition and the Roger pen directional condition. Boxes represent the median (thick horizontal line), lower and upper quartiles (end of boxes), minimum and maximum values (ends of whiskers). Values at the bottom of the figure denotes the average SRT per listening condition. P-values are corrected for multiple comparisons. Asterisks denotes significant differences.

Pen paired to both the CI and HA (SRT = 6.5 dB) revealed no significant differences (p=0.93). The average SRT in the bimodal conditions with the Roger Pen was 1.4 dB better compared with the Roger Pen paired to the CI only. The best SRT was found for the directional mode of the Roger Pen, which improved the SRT with 4.1 dB compared with the CI only condition in which the Roger Pen was paired in omnidirectional mode (p<0.0001).

Paired wise comparison between bimodal listening (SRT = 8.6 dB) and CI only conditions (SRT = 9.6 dB) without the Roger Pen revealed no significant bimodal benefit (p=0.094), although the performance tends to increase slightly.

DISCUSSION

In most previous studies on the effect of the Roger Pen, the device was used with the CI only (De Ceulaer et al., 2016, Wolfe et al., 2015a, Razza et al., 2017). Another wireless microphone, the Cochlear Minimic 2+, was also tested in bimodal CI users. Vroegop et al. (2017) showed that the use of the wireless microphone in the bimodal situation provides additional benefit compared to the use of the wireless microphone with the CI alone. However, they did not include a bimodal condition where the wireless microphone was connected to the CI only. Therefore it is possible that the benefit found was not due to the connection of the wireless microphone to the HA, but just to the addition of the HA. Our current study setup acknowledged this flaw but nevertheless showed no difference for the bimodal condition between connecting the Roger Pen to the CI alone and connecting the Roger Pen to both the CI and HA. However, improved SRT's are found when study participants added the HA while already using the Roger pen with the CI. This is probably the result of central bimodal processes, which consists of the complementarity effect and the binaural redundancy (Ching et al., 2007a). As no additional benefit is shown for also connecting the Roger Pen to the HA, apparently these bimodal processes were not influenced by this in our study set-up. However, in our set-up relatively small SNR improvements were found by use of the wireless microphone. It is possible that these central processes might be influenced if better SNRs at the HA side would be found.

In our setup, the SNR at the position of the Roger Pen was 7.5 dB better compared to the position of the listener. With a mix ratio of 70:30, theoretically, a 5 dB improvement would be expected for using the Roger Pen in the CI only condition. However, we did find a relatively small improvement of 1.6 dB for this condition compared with the CI only condition without the Roger Pen. The exact factors accounting for this difference are not known. However, the theoretical calculation assumes a 100% equally distributed omnidirectional pattern, no loss of signal quality due to transmission and mixing of the Roger signal with the CI-microphone signal. Also, there is a possible interfering effect due to the high compression ratio of the CI.

As we used a 70:30 mixed ratio, we were at least able to find a small benefit in contrast to the study of De Ceulaer et al. (2016) who used a less favourable mix-ratio of 50:50. They also used a higher distance between the Roger Pen and the loudspeaker.

The average SRT in the best condition (Roger Pen in directional microphone mode) is 3.9 dB in our study sample, an improvement in SNR of 4.1 compared with the Roger Pen in omnidirectional microphone with the CI only. In our study the distance of the Roger Pen to the speech source was 30 cm. In daily life it will likely be important to further improve the SNR by making this distance shorter. A distance of 15 cm is the clinical recommendation, which will give an additional improvement in SNR.

We chose to evaluate the effect of the Roger Pen with the settings of the Phonak Naída Link HA according to the clinical recommendations of the manufacturer, in order to be able to mimic the daily clinical practice as much as possible. One of these recommendations is the use of the special developed bimodal fitting rule, which we used in this study. However, although all different sub parts of this fitting rule are based on scientific research (Ching et al., 2001, Neuman and Svirsky, 2013, Sheffield and Gifford, 2014, Veugen et al., 2016, Zhang et al., 2014), the effect of the bimodal fitting formula as a whole has not been tested before on auditory functioning. We found no effect of bimodal hearing compared to the CI only condition in the condition without the Roger Pen. A possible explanation could be that this fitting formula is not the optimal one for all participants. Another constraint is that in our study the subjects were not used to this fitting formula. Further investigations to these special developed HA fitting formula and the effect on bimodal hearing are needed.

In the study we found for some listening conditions only small differences. It is questionable if these differences are really clinically significant. Another limitation of the study is that testing was completed in a sound-attenuated booth. Performance and benefits with the Roger Pen will probably be greater when tested in a sound booth rather than a real world environment, because of the greater influence of reverberation in the latter. Another limitation is that all participants were evaluated while using one model of sound processor, hearing aid and wireless remote microphone. These results may differ for other types of sound processors, hearing aids or remote microphones.

CONCLUSION

To conclude, the use of a wireless microphone improved speech recognition in noise for bimodal CI users. In this study it seemed sufficient to connect the wireless microphone to the CI only in the bimodal condition. The use of the directional microphone mode of the Roger Pen led to a substantial additional improvement of speech perception in noise. Therefore, application of the Roger Pen is advised for bimodal CI users to optimize their hearing in difficult listening conditions and the directional microphone mode is advised for situations with one target signal.

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Participant	Age	Gender	Etiology	HA experience non-implanted ear (years)	CI experience (years)
1	59	M	Unknown	21	5
2	34	F	Familiar	9	4
3	71	M	Familiar	17	1
4	62	F	DFNA9	26	4
5	64	F	Unknown	20	2
6	69	M	Unknown	13	2
7	72	F	Unknown	38	12
8	79	M	Unknown	25	1
9	80	F	Menière	23	2
10	48	M	Familiar	20	0.5
11	76	F	Unknown	16	1
12	48	M	Unknown	18	1
13	74	M	Menière	25	9
14	49	M	Unknown	27	11
15	68	F	Familiar	28	0.5
16	32	M	Unknown	31	4
17	57	M	Unknown	2	1
18	81	M	Unknown	20	2

Table 1 Participant demographics, including details of hearing losses and HA and CI experience

Condition number	Listening condition	Roger Pen
1	CI and HA	Off
2	CI and HA	Omnidirectional (paired to CI)
3	CI and HA	Omnidirectional (paired to CI and HA)
4	CI only	Off
5	CI only	Omnidirectional
6	CI only	Directional

Table 2 Different test conditions

Er is wel eens een vlinder op m'n vinger neergestreken
Het was er net zo een als op de kaft van Papillon
Ik heb het kleine wonder bijna ademloos bekeken
Dat op m'n eigen hand zichzelf ontvouwde in de zon

Wat lomp was alles bij die broze vlinder vergeleken
En net zat ik me af te vragen of je meten kon
Hoe breekbaar iets kan worden zonder werkelijk te breken
Toen hij weer aan zijn wilde dansje met de wind begon

KEES TORN

Chapter 8

GENERAL DISCUSSION

Sometimes, one can wonder if measuring can possibly also have a negative impact. Last year, the Council for Public Health and Society of the Netherlands published a paper about evidence-based practice in healthcare. It states that, although evidence-based healthcare has really improved healthcare, there are also pitfalls to be aware of. One of them being the ignoration of the patients context. Also the importance of the social and interpersonal aspects of the clinician and patient interaction are neglected. Ultimately, evidence-based medicine is by design based on standardized situations and average patients.

Despite these pitfalls, evidence based medicine has brought us great insights and clinical progress. I am confident that the results of this clinical study are sufficiently robust to withstand academic trial. They, however, are also pliable enough for the context of each individual patient.

HA FITTING

Bimodal hearing has shown to be beneficial compared to unilateral CI use alone in several ways; in general it improves speech recognition in difficult listening situations, sound localization abilities are better and bimodal CI users generally perceive a better sound quality. Sadly, this improvement in auditory performance is not for all bimodal CI users to be enjoyed and for some even a degradation in speech perception performance is reported.

A potential reason for this is that conventional HA fitting is mainly focused on restoring audibility. However, it can be hypothesized that for HA fitting in bimodal CI users specific characteristics of the HA fitting are needed, different to the standard HA fitting used in non-CI users. For example, it can be expected that as most CI users only have low frequency contralateral hearing, the HA fitting formula should have an accent on the low frequencies. Another difference between normal unilateral HA fitting is the loudness balance between HA and CI. For an optimal binaural hearing it could be beneficial to optimize loudness balance between HA and CI. This may be achieved by adapting overall volume or the dynamic range compression settings of the hearing aid. Other possible characteristics of the HA fitting formula which are important for bimodal hearing are frequency lowering and the frequency bandwidth. While all these aspects are probably highly individual and thus, may have to be addressed by individual fine-tuning, others might be accounted for, at least partially, by a prescriptive fitting formula.

One of the aims of this thesis was to explore how HA fitting could optimize bimodal auditory functioning. This was investigated in a systematic review of the literature and experimental studies with experienced bimodal adult CI users. We assessed the effect of different fitting strategies on bimodal auditory performance.

From the systematic review we learned that, although bimodal benefit was found in most of the reviewed studies, no clear evidence was found how certain choices in HA fitting contribute to optimal bimodal performance. A generally accepted HA prescription rule was an essential part of most fitting procedures used in the included studies. However, the effect of applying different HA prescription formulas on bimodal auditory performance was not clear. We compared the NAL-NL2 prescription formula and Audiogram+, a prescription formula from Resound, and found no difference in real ear aided response. Possibly, due to the loudness balancing procedure which was performed, differences between the two fitting rules were reduced. Another reason may be that due to feedback restrictions or limited maximum power output the gain for the higher frequencies was restricted and therefore an identical real ear aided response for the two fitting formulas was found. Also, no differences in bimodal auditory performance were noted, most probably caused by the similar HA gain provided. We also compared NAL-NL2 with the Adaptive Phonak Digital Bimodal (APDB) fitting formula. For this comparison, no loudness balancing was performed. The two fitting formulas differed in prescribed gain for the higher frequencies. Furthermore, a higher compression ratio for the APDB fitting formula was found. However these relatively small differences between the two fitting formulas did not result in different bimodal auditory performance among subjects.

So, in our studies a bimodal benefit was found, however this benefit was independent of the fitting strategies chosen. This shows that, if an adequate frequency response is provided by the HA, bimodal hearing will be beneficial over CI use alone.

This is possibly because, for these severe hearing losses, the differences between the prescription formulas are small in general, which is also shown by the real ear measurements we performed. Also, most of the bimodal CI users have low frequency residual hearing. As the energy for low frequencies in speech is relatively high, restoring audibility for low frequencies is often quite easily achieved. Another possible cause is that the bimodal effect itself is relatively small. Possibly, a higher sample size is needed to distinguish between the effect sizes of two fitting formulas.

In most of the assessed studies in the systematic review, individual fine tuning of the HA was performed, often based on balancing loudness between HA and CI. However, its additional value was not clear. Therefore we decided to investigate the effect of loudness balancing on provided gain for three different prescription formulas: NAL-NL2, Audiogram+, and APDB. We also compared two different loudness balancing methods. However, loudness balancing did not result in large deviations from the prescribed gain by the initial fitting formula. This means that fitting based on NAL-NL2, APDB or Audiogram+ generally results in a good loudness balance with the CI. Interestingly, in some individuals in our study larger deviations (8-10 dB) were found. In other words, the prescription formulas does not provide equal loudness balance between HA and CI for all bimodal users, and fine-tuning may still be relevant for the individual user. Reasons why certain users might need adjustments are possibly a deviating type of hearing loss. As most of the subjects included in the study have a severe high frequency hearing loss, loudness balancing is mainly performed at the low frequencies, because the high frequencies are not heard at the HA side. However, it can be expected that for subjects with relatively flat hearing loss, which is less severe for the high frequencies, the loudness balancing requires more attention. Another reason is that CI fittings cannot be performed according to general prescription formulas and instead subjective measurements have to be used. This may cause additional variation in the CI fitting from person to person, which possibly also affect the loudness balancing.

Using wireless technology

Although bimodal listening outperforms using a CI only, speech perception in noise is still far worse compared to that of normal hearing persons. Additional technology can possibly provide better speech recognition in difficult listening conditions. The second part of this thesis therefore investigated the effect of different wireless technologies in order to improve bimodal auditory functioning.

The first technology evaluated was the use of a binaural directional beamformer. The concept of this system is that speech information from a certain direction can be enhanced by combining the input of the CI and HA microphones. For speech within the spot of the beamformer, a clinically relevant benefit was found, on top of the benefit due to bimodal hearing. This result was not that trivial, as both concepts, bimodal hearing as well as the beamformer, are based on separating the speech source and the noise source. It was possible therefore, that both concepts would have caused a comparable benefit, but not an additional one.

However, in more dynamic listening conditions, in which the speech sources were unpredictable, the additional benefit of the beamformer was lost. Instead, the benefit of bimodal listening remained stable for these unpredictable situations. This shows that although the bimodal benefit is small, it remains effective for a broad range of listening conditions, while additional wireless techniques are mainly effective in static listening conditions. As we tested this beamformer only in a test set-up in the clinic, no data is available about the experiences of this beamformer in real-life daily listening conditions. This would be interesting to know, as the question is how frequent the more static conditions are where it is mainly effective compared to the more dynamic conditions. It can be expected that for some CI users dynamic conditions are more frequently present, like for example work situations with group conversations and meetings, and then bimodal listening seems to be the best option and binaural beamforming becomes less effective.

Another limitation of the binaural beamformer is that it requires near field situations where the sound source is located close by. In case the sounds source is located in the far field, a wireless remote microphone can be used instead to improve hearing in demanding listening situations. We investigated the possible additional benefit of two different types of wireless remote microphones for bimodal CI users. Both remote microphones connected to the CI resulted in improved in speech recognition in noise. An additional benefit was found when also using the HA, next to the CI. This is probably the result of central bimodal processes, which consists of the complementarity effect and the binaural redundancy. The results of the study show that it seems sufficient to connect the wireless microphone to the CI only. As no additional benefit is shown for also connecting the wireless microphone to the HA, apparently the central processes are not influenced by a better SNR at the HA side.

Combining bimodal listening with wireless techniques results in a substantial benefit up to seven dB compared with listening with the CI only in certain complex listening conditions. Interestingly, the two effects of bimodal listening and wireless technology add up, resulting in a large improvement. This enables bimodal listeners to participate more frequently in challenging conditions, improving social interaction, etc.

FUTURE DIRECTIONS FOR HA FITTING IN BIMODAL CI USERS

Like all other studies, our study has its limitations. One of the main limitations is the reduced generalizability of our results, due to our sample size, inclusion criteria and set-up of the tested listening conditions. Overall no differences were found in bimodal auditory performance in our studies for different HA fitting strategies, however individual differences were found. This is comparable with other HA fitting studies in bimodal CI users. Our study, like all others in the systematic review, is limited in its capability to individually assess the added value of specific fitting factors, due to small sample size and relatively small differences that can be expected. There is urgent need for larger (multicenter) studies in order to determine critical factors which influence bimodal functioning.

Another opportunity for future research is to use broader inclusion criteria. In our studies only CI users with a relatively good speech perception performance were included. It can be hypothesized that for CI users with poor speech perception abilities, the HA fitting needs to fulfill other requirements. Furthermore, we included subjects in our study with a relatively good residual hearing. Possibly for CI users with more severe contralateral hearing loss, also other requirements for the HA fitting are needed.

Future studies on the current topic should also incorporate different listening conditions. As no differences in bimodal benefit were found in our studies for different fitting strategies, it is possible when measuring in other listening conditions more differences would be found. By example, two-talker or multitalker noise could be used, instead of the stationary noise which was used in our study set-up. Furthermore, we investigated the different fitting strategies for noise from the left, right or front. Possibly a set-up with a diffuse noise field, noise from both left and right side together or other noise set-ups will provide more differences between strategies.

A more specific limitation of our study is the balancing method used. We did not test repeatability of the balancing procedures in our study. This should be incorporated in future research as it improves the interpretation of the findings of studies that include loudness balancing. Moreover, a clinically applicable reliable balancing procedure can then be obtained.

In addition, we assessed the experience of the subjects with the different fitting strategies only with the SSQ. It is possible that the SSQ is not sensitive enough for small differences between the fitting strategies. Moreover, we did not incorporate questions which are important for the bimodal CI users itself. In future research, bimodal CI users should be asked what is important for them and tests should incorporate and measure the needs of the CI users.

Future studies to the binaural beamformer should incorporate field studies to evaluate the effect of the beamformer in daily life. From these field studies it valuable information will be obtained for which listening situations subjects really benefit from the beamformer, which will be helpful for clinicians to counsel new users in the use of it.

IMPLICATIONS AND RECOMMENDATIONS FOR CLINICAL PRACTICE

Due to the relatively high auditory performance of most of the CI users and the rapid evolution of implant technology, selection criteria for CI are expanding. Nowadays, more and more candidates with residual hearing are receiving a cochlear implant. Subsequently the amount of CI users with a contralateral hearing aid will expand, forcing CI-clinics to upgrade their knowledge on HA fittings. In the most optimal situation the CI and HA can be fitted by the same clinical specialist and fitting rooms are equipped with both HA fitting and CI fitting equipment. Naturally, clinical evaluations of auditory performance, should involve the CI, HA and the bimodal performance.

For a good HA fitting in bimodal CI users adequate amplification seems the most important factor. Real ear verification of the aided response should therefore be the standard when fitting the HA. In our study, average adjustments until 5 dB were needed for the different frequency bands to compensate for the earmould in order to obtain the needed amplification. We advise to use a standard fitting formula for which the target HA aided response is known, like the NAL-NL2.

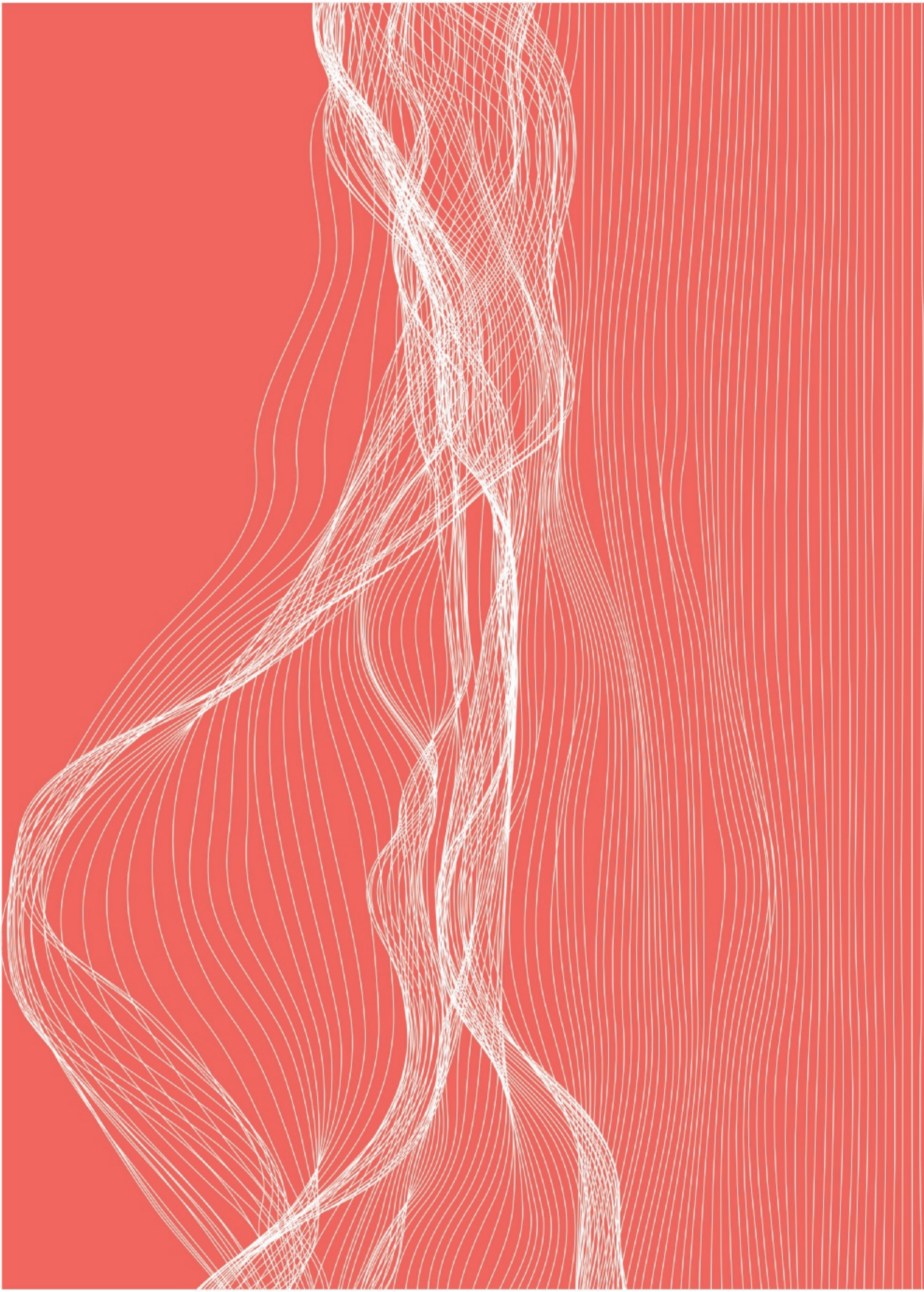
However, if a prescription formula is used and no target is known, it is advisable to start with a standard fitting formula to investigate the effect of the earmould.

If you would like to use a proprietary HA fitting formula from a manufacturer, we advise to start the fitting process with a standard fitting formula and optimize the frequency response with the real ear measurements. Afterwards, add the compensation due to the earmould to the gain of the fitting formula you want to use.

On average, loudness balancing did not result in large deviations from the prescribed HA gain. However, for some individuals larger deviations were found. We recommend to perform a simple broadband loudness balancing procedure to check if equal loudness between HA and CI is established, e.g. adjusting the volume till broadband speech in front is heard in the middle. More evidence is needed for more precise/extended procedures.

For CI users with an AB Naida Q90 CI and a Naída Link HA it is advisable to make a Stereozoom program, as it is able to substantially improve speech perception in noise. But be aware that counseling of the CI user is really important, as it likely works only in certain specific listening conditions where the sound source is known and near-by.

For bimodal CI users experiencing difficulties in noisy situations when the sound source is further away, the use of wireless remote microphones is advised. It seems sufficient to connect the wireless microphone to the CI only, as long as the HA is worn.



Chapter 9
SUMMARY

Hearing impairment is a major health problem affecting more than 450 million people worldwide. For individuals with severe to profound hearing loss, cochlear implants (CI) have become a viable treatment option. A CI is a surgically implanted electronic device that allows people with severe hearing loss to access sound and to communicate more effectively with their peers. Cochlear implantation has caused a major shift in the treatment of severe to profound sensorineural hearing loss. The high performance of most of the CI recipients coupled with the rapid evolution of implant technology lead to a distinct expansion in selection criteria for cochlear implantation. Therefore, more and more candidates with residual hearing are receiving a cochlear implant and contralateral hearing aids are worn in many cases. The combination of a CI in one ear and a HA in the other ear is called bimodal hearing, aiming to restore binaural hearing as much as possible. It has been shown to be beneficial compared to unilateral CI use alone in difficult listening situations. It can be expected that, in order to achieve optimal bimodal hearing, specific requirements for the HA fitting are needed to reach the full hearing potential of each patient. To achieve this, more evidence about HA fitting for bimodal patients should be collected. Therefore this thesis focusses on HA fitting to optimize bimodal auditory functioning. This thesis evaluates the effect of wireless technology for optimizing bimodal hearing as well.

In chapter two a review of the literature is given on the effect of different hearing aid fitting strategies on auditory performance in bimodal CI users. Not many studies did fit to the inclusion criteria and only seventeen studies were included in the review. In general, a bimodal benefit was found in most studies, using various strategies for the HA fitting. Using a standard prescription rule is a good starting point in children and adults. However, there is no clear evidence how certain choices in hearing aid fitting contribute to optimal bimodal performance. Current evidence suggests that frequency lowering is not beneficial. Individual fine tuning based on loudness or general preference is often applied, but its additional value for auditory performance should be investigated more thoroughly.

Chapter three describes a study which investigates the effect of hearing aid fitting procedures on provided gain and bimodal auditory functioning in bimodal CI users. The hearing aid fitting methods differed in initial gain prescription formula and loudness balancing method. No differences in provided gain and in bimodal performance were found for the different hearing aid fittings. For all hearing aid fittings a bimodal benefit was found for speech in noise as well as sound localization. These results confirm that cochlear implant users with residual hearing in the contralateral ear substantially benefit from bimodal stimulation. However, on average, no differences were found between different types of fitting methods, varying in prescription rule and loudness balancing method.

In chapter four a study to the possible advantage of the use of a dedicated bimodal hearing aid fitting formula is described. In this study this fitting formula is compared with a frequently used standard hearing aid fitting formula. We evaluated the effects of bimodal hearing aid fitting on provided hearing aid gain and on bimodal auditory functioning in a group of experienced bimodal cochlear implant users. We also described the effect of broadband loudness balancing on the prescribed gain of those two fitting formulas. Significant differences between the two fitting formulas were found for the high frequencies. For the dedicated bimodal fitting formula a higher compression ratio the mid and high frequencies was found. Loudness balancing did not result in large deviations from the prescribed gain by the initial fitting formula. Bimodal benefit was found for speech perception in quiet as well as for speech perception in noise. No differences in auditory performance were found between the two fitting formulas for any of the auditory performance tests. So, although significant differences between the output and compression ratio of the two fitting formulas existed, no differences in bimodal auditory performance were observed. Next, additional loudness balancing has only a marginal effect on the provided hearing aid output.

Chapter five investigates the use of binaural directional microphones, a so called binaural beamformer, on auditory bimodal performance. A significant improvement for speech perception in noise was found with the binaural beamformer for predictable or static listening conditions. For dynamic listening conditions no improvement for the beamformer was found. The two effects of bimodal listening and wireless technology add up, resulting in a large improvement.

In chapter six and seven, the effect of two different wireless remote microphones was investigated. Both microphones did improve the speech perception in noise in bimodal CI users. The most benefit was found for the bimodal listening condition.

This thesis shows the importance for CI users of wearing a contralateral hearing aid in case of residual hearing. This bimodal hearing improves speech perception in noise. However, the different HA fitting methods studied in this thesis did not result in different bimodal auditory functioning. Adequate amplification seems to be the most important factor for bimodal benefit, the exact fitting strategy is of less importance. However, individual differences exist and therefore there is a need for larger comparative studies to investigate factors influencing bimodal auditory performance. The use of wireless technology to improve speech perception in noise has been shown beneficial, for binaural beamformers as well as for wireless remote microphones in specific listening conditions.

Chapter 10

NEDERLANDSE SAMENVATTING

Slechthorendheid is een groot gezondheidsprobleem dat wereldwijd meer dan 450 miljoen mensen treft. Voor mensen met ernstige tot zeer ernstige gehoorverliezen zijn cochleaire implantaten (CI) een belangrijke behandeloptie geworden. Een CI is een chirurgisch geïmplanteerd elektronisch apparaat waarmee mensen met ernstig gehoorverlies toegang hebben tot geluid en effectiever kunnen communiceren. Cochleaire implantatie heeft een belangrijke verschuiving veroorzaakt in de behandeling van deze ernstige gehoorverliezen.

De goede prestaties van de meeste CI-gebruikers in combinatie met de snelle evolutie van de implantaattechnologie leiden tot een duidelijke uitbreiding van indicatiecriteria voor cochleaire implantatie. Daarom ontvangen steeds meer mensen met restgehoor een CI en wordt na implantatie in veel gevallen aan de andere zijde een hoortoestel gedragen. De combinatie van een CI in het ene oor en een hoortoestel in het andere oor wordt bimodaal horen genoemd. Dit is gericht op het zo veel mogelijk herstellen van het binauraal horen. Het is aangetoond dat het voordeel oplevert in vergelijking met eenzijdig CI-gebruik in moeilijkere luistersituaties. Het is te verwachten dat, om een optimaal bimodaal auditief functioneren te bereiken, er specifieke vereisten zijn voor de hoortoestelaanpassing. Echter, op dit moment is wetenschappelijk bewijs over hoortoestel aanpasstrategieën bij bimodale CI-patiënten schaars.

Dit proefschrift concentreert zich daarom op het exploreren van manieren om de hoortoestelaanpassing te verbeteren om hiermee het bimodaal auditief functioneren verder te optimaliseren. Ook wordt in dit proefschrift het effect van draadloze technologie voor het optimaliseren van het bimodaal horen geëvalueerd.

In hoofdstuk twee wordt een overzicht gegeven van de literatuur over het effect van verschillende aanpassingsstrategieën voor hoortoestellen bij bimodale CI-gebruikers. Maar enkele gevonden studies voldeden aan de inclusiecriteria en slechts zeventien studies werden na inhoudelijke, kwalitatieve beoordeling in de analyse opgenomen. Over het algemeen werd een bimodaal voordeel gevonden in de meeste onderzoeken, waarbij allerlei verschillende strategieën voor de hoortoestel-aanpassing werden gebruikt. Het gebruik van een standaard rekenregel is een goed uitgangspunt bij kinderen en volwassenen. Er is echter geen duidelijk bewijs hoe bepaalde keuzes in de aanpassing van het hoortoestel bijdragen aan optimale bimodale prestaties. Huidig bewijs suggereert dat frequentieverlaging niet bijdraagt aan een beter functioneren. Individuele fijnafregeling op basis van luidheid of algemene voorkeur wordt vaak toegepast, maar de toegevoegde waarde voor het auditief functioneren moet nog verder worden onderzocht.

Hoofdstuk drie beschrijft een onderzoek dat het effect van verschillende hoortoestel aanpassingsprocedures op de versterking en het bimodaal auditief functioneren onderzocht. De methoden voor het aanpassen van de hoortoestellen verschilden in de gebruikte aanpassingsformule en manier van het balanceren van de luidheid tussen het hoortoestel en het CI (luidheidsbalanceringsmethode). Voor de verschillende aanpassingsstrategieën, werden geen verschillen gevonden tussen de geleverde versterking en tussen het bimodaal auditief functioneren. Voor alle fittingsmethoden werd een bimodale winst gevonden voor het spraakverstaan in rumoer evenals voor geluidslokalisatie. Deze resultaten bevestigen dat CI-gebruikers met restgehoor in het contralaterale oor substantieel baat hebben bij bimodale stimulatie. Gemiddeld werden echter geen verschillen gevonden tussen verschillende soorten aanpasmethoden, variërend in aanpasregel en luidheidsbalanceringsmethode.

In hoofdstuk vier wordt een onderzoek beschreven naar het mogelijke voordeel van het gebruik van een speciale bimodale aanpasregel van het hoortoestel. In deze studie is deze aanpasregel vergeleken met een veelgebruikte standaard hoortoestelrekenregel. We evalueerden het effect van de beide rekenregels op de geleverde versterking uit het hoortoestel en op het bimodaal auditief functioneren in een groep ervaren bimodale CI-gebruikers. Ook beschreven we het effect van breedbandige luidheidsbalancering op de gegeven versterking van deze twee fittingregels. Significante verschillen in versterking tussen de twee aanpasformules werden gevonden voor de hoge frequenties. Voor de speciale bimodale hoortoestelformule werd een hogere compressieverhouding gevonden voor de midden- en hoge frequenties. Luidheidsbalancering resulteerde niet in grote afwijkingen van de initiale voorgeschreven versterking. Bimodaal voordeel werd gevonden voor spraakverstaan van zachte spraak in stilte en voor spraakverstaan in rumoer. Er werd geen verschil in bimodaal auditief functioneren gevonden voor de twee verschillende rekenregels.

Voor het verbeteren van spraakverstaan in rumoer kan men gebruik maken van directionele microfoons. Het geluid wat van de achterkant komt wordt dan meer verzwakt, dan geluiden die van de voorzijde komen. Wanneer je de informatie van deze directionele microfoons combineert voor twee oren spreekt men van binaurale directionele microfoons. In hoofdstuk vijf is het gebruik van binaurale directionele microfoons, een zogenaamde binaurale beamformer, onderzocht op het bimodaal auditief functioneren. Een significante verbetering voor spraakverstaan in ruis werd gevonden met de binaurale beamformer voor voorspelbare of statische lusiteromstandigheden. Voor dynamische luisteromstandigheden werd geen verbetering voor de beamformer gevonden.

In hoofdstuk zes en zeven werd het effect van twee verschillende draadloze microfoons onderzocht. Beide microfoons verbeterden de spraakperceptie in ruis bij bimodale CI-gebruikers. Het meeste voordeel werd gevonden voor de bimodale luisterconditie, waarbij het voordeel van bimodaal horen én van de draadloze microfoon bij elkaar optellen, resulterend in een grote verbetering.

Dit proefschrift toont aan hoe belangrijk het is voor CI-gebruikers om contralateraal een hoortoestel te dragen, in het geval er nog restgehoor is. Dit bimodaal horen kan het spraakverstaan in ruis verbeteren. De verschillende aanpasmethoden voor het hoortoestel die in dit proefschrift werden onderzocht, resulteerden echter niet in een verschil in bimodaal auditief functioneren. Adequate versterking lijkt de belangrijkste factor voor bimodale winst te zijn, de exacte aanpasstrategie is van minder belang. Er zijn echter individuele verschillen en daarom is er behoefte aan studies met een groter aantal proefpersonen die onderzoeken welke factoren precies het bimodaal auditief functioneren beïnvloeden. Het gebruik van draadloze technologie om het spraakverstaan in ruis te verbeteren voor bimodale CI-gebruikers, is nuttig gebleken. Zowel voor binaurale beamformers als voor draadloze microfoons.

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CURRICULUM VITAE

Jantien Vroegop is geboren op 17 september 1984 en opgegroeid in Sint-Annaland, in het mooie Zeeland. Na het eindexamen VWO aan het Calvijn College te Goes in 2002 begon zij met de opleiding Biomedische Technologie aan de Technische Universiteit te Eindhoven. Ze koos voor de Master Medical Engineering, die ze in 2007 afsloot. Tijdens haar afstudeeronderzoek werkte ze aan een mathematisch model om de effecten van een arterioveneuze fistel op de hartfunctie te onderzoeken.

Na haar afstuderen maakte ze de overstap naar de audiologie en is ze gestart met de opleiding tot klinisch fysicus – audioloog in het Erasmus MC. Deze opleiding rondde ze in 2011 af. Direct aansluitend startte Jantien als audioloog in het Erasmus MC en specialiseerde zich in de zorg rondom cochleair implantaten. Inmiddels is zij verantwoordelijk voor de audiologische zorg binnen het CI-team Rotterdam, de stad waar ze met veel plezier woont met Hijmen, Tijn en Fiene.

Naast haar werkzaamheden als audioloog startte zij met een promotietraject onder begeleiding van dr. ir. A. Goedegebure en dr. M.P. van der Schroeff, met professor dr. R.J. Baatenburg de Jong als promotor, waarvan dit proefschrift het resultaat is.

PHD PORTFOLIO

Summary of PhD training and teaching
PhD period: 2016 – 2018
Name PhD student: J.L. Vroegop
Promotor: Prof. Dr. R.J. Baatenburg de Jong
Erasmus MC Department: KNO-heelkunde

1. PhD training	**Year**	**Workload** ECTS
Specific courses		
– Introduction to data analysis, NIHES	2010	0.7
– BROK cursus, ErasmusMC	2014	1.5
– Teach the Teacher III	2017	0.3
– Wetenschappelijke integriteit, ErasmusMC	2018	0.3
– BROK herregistratie	2018	0.3
Presentations		
– ESPCI Istanbul	2013	1
– ESPCI Toulouse	2015	1
– CI2016 Toronto 2x	2016	2
– Symposium Cochlear	2017	1
– CI2018 Antwerpen 3x	2018	6
(inter) national conferences		
– IERASG Rio de Janeiro	2009	1
– Widex Pediatric conference Dubai	2010	1
– IERASG Moskou	2011	1
– Phonak pediatric conference Istanbul	2012	1
– KKAU conferentie audiologie	2015	0.5
– NVA 2012, 2013, 2014, 2015	2015	1
– VU symposium Language & Hearing	2016	0.25
– Symposium CI Radboud UMC (2x)	2016	1
– ESPCI Lissabon	2017	1
– Symposium NSDSK	2018	0.25
– Scientific meeting AB Venetië	2018	1
– KKAU conferentie audiologie	2018	0.5

2. Teaching

Support of different research internships of students of speech and language therapy and medicine.

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