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Beamforming and Single-Microphone Noise Reduction: Effects on Signal-to-Noise Ratio and Speech Recognition of Bimodal Cochlear Implant Users

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Abstract

We have investigated the effectiveness of three noise-reduction algorithms, namely an adaptive monaural beamformer (MB), a fixed binaural beamformer (BB), and a single-microphone stationary-noise reduction algorithm (SNRA) by assessing the speech reception threshold (SRT) in a group of 15 bimodal cochlear implant users. Speech was presented frontally towards the listener and background noise was established as a homogeneous field of long-term speech-spectrum-shaped (LTSS) noise or 8-talker babble. We pursued four research questions, namely: whether the benefits of beamforming on the SRT differ between LTSS noise and 8-talker babble; whether BB is more effective than MB; whether SNRA improves the SRT in LTSS noise; and whether the SRT benefits of MB and BB are comparable to their improvement of the signal-to-noise ratio (SNR). The results showed that MB and BB significantly improved SRTs by an average of 2.6 dB and 2.9 dB, respectively. These benefits did not statistically differ between noise types or between the two beamformers. By contrast, physical SNR improvements obtained with a manikin revealed substantially greater benefits of BB (6.6 dB) than MB (3.3 dB). SNRA did not significantly affect SRTs per se in omnidirectional microphone settings, nor in combination with MB and BB. We conclude that in the group of bimodal listeners tested, BB had no additional benefits on speech recognition over MB in homogeneous noise, despite the finding that BB had a substantial larger benefit on the SNR than MB. SNRA did not improve speech recognition.

Keywords

cochlear implants, hearing aids, sensorineural hearing loss, speech perception, front-end processing

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Introduction

Severe-to-profound sensorineural hearing loss can be successfully treated with a cochlear implant (CI). Cochlear implantation generally results in good speech recognition in quiet. However, speech recognition deteriorates markedly in noise, and particularly in fluctuating noise (Nelson et al., 2003; Shannon et al., 2011; Stronks et al., 2020; Zeng et al., 2005). This sensitivity to fluctuating noise is believed to be caused mainly by a reduced access to pitch cues and the lack of temporal-fine structure (TFS) in the CI signal (Fu & Nogaki, 2004; Hopkins & Moore, 2009; Qin & Oxenham, 2003). Typically, speech recognition testing in the clinic is either performed in quiet, or in the presence of stationary, steady-state noise. However, real-life background noise often fluctuates in time. Previously, we have shown that introducing temporal modulation and speech-like

TFS in stationary noise decreases speech recognition in CI users (Stronks et al., 2020).

One way to potentially enhance speech recognition in noise is to make pitch cues and TFS better available for CI

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users. This can be achieved by fitting the contralateral ear with a hearing aid (HA) referred to as bimodal hearing, provided that there is sufficient residual hearing in that ear (Hopkins & Moore, 2009; Oxenham & Simonson, 2009; Qin & Oxenham, 2003; Stronks et al., 2020; Turner et al., 2004).

A second approach to reduce the effects of background noise is by means of front-end processing, such as single-microphone noise reduction and beamformers. Directional microphones, or beamformers, are spatial filters that attenuate sound (noise) originating from peripheral directions (Van Hoesel & Clark, 1995). A beamformer utilizes multiple microphones that can be located close to each other on the same behind-the-ear (BTE) unit, referred to as a monaural beamformer, or they can be placed more distant from each other at two different ears to generate a binaural beamformer (Chung & Zeng, 2009). By contrast, single-microphone noise reduction algorithms deploy the input of just a single acoustic input to reduce noise with spectral filtering.

The Advanced Bionics Naída Link™ system (Cuda et al., 2019) is a hearing solution for bimodal listeners that allows the CI and HA to communicate wirelessly. There are multiple speech enhancement algorithms available on the system, including a monaural beamformer (UltraZoom™), a binaural beamformer (StereoZoom™), and various single-microphone noise reduction algorithms including one called ClearVoice™ (on the CI) or its acoustic equivalent called NoiseBlock™ (on the HA) (Gustafson et al., 2014). These three algorithms were investigated in this study and will hereafter be referred to as MB, BB and SNRA, respectively. SNRA detects stationary noise in individual frequency bands and suppresses the gain of those that predominantly contain stationary noise (Buechner et al., 2010). There are also indications that SNRA may improve speech recognition in specific types of fluctuating noise, but not others (Holden et al., 2013).

In bimodal listeners, MB has been reported to improve the speech reception threshold (SRT) between 1.6 dB (Devocht et al., 2016) and 3.4 dB SNR (Ernst et al., 2019). BB has been shown to yield a significantly larger SRT improvement, namely approximately 4.6–4.7 dB (Ernst et al., 2019; Vroegop et al., 2018). Because of the impact of the spatial distribution of noise (Geißler et al., 2015), we chose to use a homogeneous field by presenting the noise uniformly around the listener. This facilitates a fair comparison between MB and BB, because setups with the loudspeakers positioned at specific angles can favor one over the other (Soede et al., 1993). We have used the SRT as outcome measure, which has the advantage that it can be directly linked to physical outcome measures of beamforming, including signal-to-noise ratio (SNR) improvement and directivity estimates obtained from polar patterns.

From a physical point of view, any benefit of beamforming on the SRT is expected to be attributable to an improvement of the SNR (Gannot & Cohen, 2008) that manifests

itself predominantly through the better-hearing ear (Williges et al., 2019). The benefits of beamformers are, however, frequency dependent, and thus depend on the spectral properties of the noise. There are several studies that investigated the difference in benefits of beamforming between different noise types in CI users. One study that included a relatively small population of five CI users showed that beamforming was significantly more effective in multitalker babble than in long-term speech-spectrum-shaped (LTSS) noise (Spriet et al., 2007). Other studies, however, reported no significant differences between babble noise and LTSS noise in CI users (Weissgerber et al., 2017), or bimodal listeners (Devocht et al., 2016). In this study we compared beamforming effectiveness between LTSS noise and multitalker babble noise. LTSS noise is representative of stationary noise often used in the clinic and lab settings, whereas multitalker babble is more representative of noise encountered in daily life.

Investigations on the effects of SNRA on speech recognition have yielded conflicting evidence in earlier reports. Some studies conclude that SNRA significantly improves speech recognition in stationary noise when tested by itself, or in combination with MB and BB in CI users (Buechner et al., 2010; Ernst et al., 2019; Kam et al., 2012). Other studies, however, have shown no beneficial effects of the algorithm on speech recognition (Dingemans & Goedegebure, 2015, 2018), or only when it is set to the most aggressive filter setting clinically available, i.e., the ‘high’ setting instead of the default ‘moderate’ setting as used in the present study (Noël-Petroff et al., 2013). To our knowledge, the effectiveness of SNRA has never been tested in conjunction with beamformers in bimodal listeners.

In this study, we have tested the effectiveness of MB, BB, SNRA, and combinations of them on the SRT in bimodal listeners in a field of homogeneous LTSS noise and in 8-talker babble noise. Because of the ongoing relaxation of implantation criteria and the fact that only unilateral CI is reimbursed for adults in the Netherlands and several other countries, bimodally fitted CI users are a growing population. Unlike traditional, bilaterally profoundly deaf unilateral CI users, they can use a binaural beamformer like BB, making the study clinically relevant.

The physical effects of MB and BB were also determined in both noise types by recording SNR improvements via the output of the CI when mounted on a KEMAR manikin (Burkhard & Sachs, 1975) in the same homogeneous noise field as where the speech recognition tests were performed. Directivity of the beamformers were assessed using the polar patterns. SNRA’s effects, if any, were expected to become evident only in LTSS noise.

The specific research questions we investigated were whether beamforming effectiveness differs between LTSS noise and 8-talker babble; whether BB is more effective than MB in a homogeneous noise field; whether SNRA improves the SRT in LTSS noise; and whether the SRT

benefits of MB and BB are comparable to signal-to-noise ratio (SNR) improvements as recorded with a KEMAR manikin.

Materials & Methods

Participant Population

15 study participants who were unilaterally and postlingually implanted with an Advanced Bionics CI were recruited for this study. Inclusion criteria were: (1) residual hearing in the non-implanted ear with audiometric pure-tone thresholds of 80 dB HL, or better at 125, 250 and 500 Hz, and (2) a CVC phoneme correct score of at least 80% with their CI alone in quiet, which is an above-average performance in our clinic. Before testing, informed consent was obtained. Study participants received travel reimbursements. This study was approved by the Institutional Review Board of the Leiden University Medical Center, and adhered to the tenets of Helsinki (World Medical Association, 2013). Table 1 lists the demographic details of the study participants and Figure 1 shows the median audiogram (with interquartile range) of the ear contralateral to the CI where the HA was fitted. The postoperative audiograms of the implanted ear were unavailable, because they were not routinely determined in our clinic at the time of this writing.

Study Design

This clinical trial was a prospective intervention study and was single-blinded (participants were unaware of the algorithm being tested). Test conditions were randomized. A repeated measures design was applied, i.e., every participant completed all the test conditions. Speech tests were obtained typically in 4 test sessions, performed on separate days. Most participants either preferred to complete one session per week, or to come in every other week. Per session, typically a single noise type was tested (LTSS noise or 8-talker babble). Noise type was alternated between sessions, such that a test and re-test was obtained for each microphone setting in each noise type. The different microphone settings were tested in random order within a session. This research was part of a larger study where some additional conditions were tested that are not presented here. In total, approximately 12 speech tests were run per test session, each test lasting approximately 5 min. A break halfway the session was encouraged. The LTSS noise and 8-talker babble were tested and re-tested in all but one participant for whom the re-test of the babble could not be performed. This person left the study due to health reasons unrelated to hearing or to this research.

Contralateral Hearing aid Fitting for Home use

All participants were fitted with a contralateral HA (Naída X UP, or Naída Link device; Phonak, Sonova Holding AG

Stäfa, Switzerland) using the Adaptive Phonak Digital Bimodal Fitting Formula (Cuda et al., 2019) and PhonakTarget 3.3 (Sonova Group, Stäfa, Switzerland) fitting software. The bimodal fitting rule is a dedicated formula for bimodal listeners that optimizes the acoustic gain at low frequencies and aligns loudness growth and dynamic compression behavior of the HA with that of the CI speech processor (Warren et al., 2020). Real-ear measurements were not performed. For fine tuning, the type of earpiece, and the diameter of the tube and the vent were entered in the fitting software to correct for these variables. The participants were then asked whether the acoustic gain was comfortably loud and whether the perceived loudness of the HA was on a par with the CI when conversing with the experimenter. If not, adjustments to the overall HA volume were made.

All participants had used HAs before receiving their CI and they used their own ear molds (typically a full shell) for home use and during lab testing as well. Eight participants (S02-S11) had participated in an earlier trial and had been fitted with a Naída X UP approximately 2 years before this study started. Six of them had been using that HA ever since, the remaining two (S06, S09) had stopped using the HA somewhere in the intervening period between the two studies. These two were encouraged to start using the HA again before testing was started. Of the 7 newly recruited participants, 2 used a Naída Link with the bimodal fitting formula (S12, S18) when they were recruited, and 1 wore a Naída X UP device (S17). Their HA had been fitted by a professional health care provider, and they used the HA daily. S17's HA was re-fitted with the bimodal fitting formula, leaving the remaining settings intact. The other 4 newly recruited participants (S13-S16) and the two participants who stopped using their HA after the preceding trial (S06 and S09) were newly fitted with a Naída X UP or Naída Link device. These participants were encouraged to use their HA daily for at least 4 weeks before being tested.

The programs fitted on the home-use HA generally mirrored the CI, i.e., if different front-end settings were fitted on the CI speech processor, they were duplicated on the HA. However individualized settings were sometimes required on the HA, such as whistle suppression (WhistleBlock™), wind noise suppression (WindBlock™), or suppression of impulse noise (SoundRelax™). To optimize HA functionality during everyday life, the HA fitting was adjusted in-between test sessions during the study when needed. In our clinic, adult CI users with a Q90 Advanced Bionics processor are typically fitted with at least two programs, namely: a program for regular listening conditions with omnidirectional microphone settings without SNRA, and a program for listening in background noise with a combination of MB and SNRA. Most of our participants (12 out of 15) had both MB and SNRA implemented on their CI speech processor when they were recruited for this study. One of the three remaining participants used a combination of BB and SNRA, one used only

Table 1. Participant Demographics.

ID	Age	HA use	PTA _{500–2000}	CI use (years)	CVC (%)	Array	Algorithms on home-use CI			Etiology of hearing loss
2	71	Yes	85	4.9	89	MS	O	MB	SNRA	Possibly antibiotic-induced
4	82	Yes	75	2.9	91	MS	O	MB	SNRA	Possibly familial; progressive
5	74	Yes	90	2.9	86	MS	O	MB	SNRA	Familial; progressive
6	62	No	60	4.4	100	Ij	O		SNRA	Meniere's disease; progressive
7	86	Yes	80	3.3	97	MS	O	MB		Unknown; progressive
9	86	No	65	2.4	78	MS	O	MB	SNRA	Unknown; progressive
10	67	Yes	65	3.9	76	MS	O	MB	SNRA	Congenital; hereditary (DFNA9); progressive
11	62	Yes	55	2.0	96	Ij	O			Congenital; hereditary (DFNA9); progressive
12	75	Yes	65	1.7	80	MS	O	BB	SNRA	Congenital; hereditary (DFNA9); progressive
13	62	No	120	1.8	85	MS	O	MB	SNRA	Unknown; possibly familial; progressive
14	58	No	80	2.1	91	MS	O	MB		Congenital; familial; progressive
15	79	No	120	1.5	83	MS	O	MB	SNRA	Unknown; progressive
16	85	No	110	1.7	87	MS	O	MB	SNRA	Unknown; progressive
17	78	Yes	70	1.5	92	MS	O	MB	SNRA	Unknown; progressive
18	50	Yes	70	1.5	86	MS	O	MB	SNRA	Unknown; progressive
Total N = 15	Mean 72	Total 6 no	Median 75	Mean 2.6	Mean 88		Totals 2 Ij 15 12 1 12			

Ij: HiFocus™ Ij electrode array; BB: binaural beamformer; CI: cochlear implant type; CVC: free-field consonant-verb-consonant phoneme score with the CI alone; HA: hearing aid; ID: participant number; MB: monaural beamformer; MS: HiFocus™ Mid-Scala electrode array; O: omnidirectional microphone setting; PTA_{500–2000}: average pure tone audiogram across 500, 1000 and 2000 Hz of the contralateral ear; SNRA: single-microphone noise-reduction algorithm. Three participants were excluded from the study (ID 1, 3 and 8).

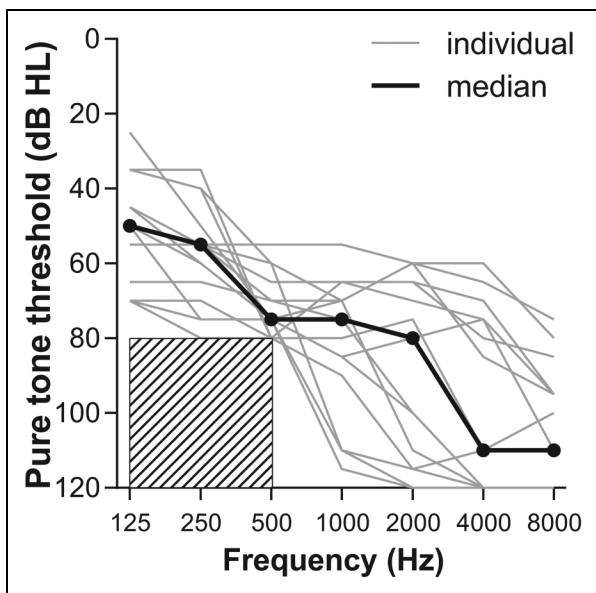


Figure 1. Individual pure-tone audiograms (thin gray lines) with the median (thick black line) for the non-implanted ear of all participants. Gray box: exclusion criterion based on residual hearing.

MB without SNRA, and one used SNRA without a beamformer (Table 1).

Listening in noise with only the HA activated was challenging; all participants performed worse with their HA

alone than with their CI alone; see Fig. 4 in Stronks et al. (2020), where the same group of participants was tested. From these results we conclude that the CI ear was the better-hearing ear when listening in noise.

Experimental CI and HA Fitting. At the start of each test session, the participants were fitted with a research Naída CI Q90™ speech processor (Advanced Bionics LLC, Valencia, CA, USA), and a research Naída Link HA. The experimental CI and HA processors were symmetrically fitted with an omnidirectional setting (O), MB, BB and a combination of these settings with SNRA. Because the Q90 processor accommodated a maximum of 5 programs, 2 sets of CI+HA were used, namely one set with a fitting without SNRA (O, MB, BB) and the other with SNRA (O + SNRA, MB + SNRA, BB + SNRA). No other front-end noise reduction algorithms were active. The Q90 speech processor was fitted with the participant's own threshold (T) levels and maximal-comfortable (M) levels. The Naída Link HA was also fitted using the participant's own HA settings.

The frequency range of the acoustic band-pass filter of the CI was set at 'standard' (350–8700 Hz), instead of the default 'extended low' (250–8700 Hz) to reduce overlapping, potentially conflicting frequency information between the CI and HA (Mok et al., 2006). Incongruent information at low frequencies can arise because of a mismatch between the insertion depth of CIs and the frequency map.

By taking the mean of the four studies with Advanced Bionics devices mentioned in Table 1 of Landsberger et al. (2015), an average insertion angle of approximately 450 degrees is obtained, corresponding to 550 Hz on the spiral ganglion map (Stakhovskaya et al., 2007). Setting the lower boundary of the acoustic band filter of the CI below this value maps low frequencies to spiral ganglion cells with higher characteristic frequency.

Noise Reduction Algorithms. MB (UltraZoom) is an algorithm that utilizes the signals of two omnidirectional microphones on the front-back axis of the BTE unit of the CI and HA (Elko & Anh-Tho Nguyen, 1995), which are spaced approximately 1 cm apart on both devices. It operates independently on the CI and HA. The algorithm is an adaptive multiband beamformer (Voss et al., 2021) and attempts to maximize the attenuation of sounds originating from the back hemisphere by estimating the region with the lowest signal-to-noise ratio (SNR) in the auditory scene. The region of maximal attenuation (the null direction) is steered toward the noise-dominated region, potentially further increasing the SNR by noise suppression (Buechner et al., 2014; Hehrmann et al., 2012). The adaptation time constant for the null direction is approximately 150 ms. Because the noise was running continuously during speech testing, MB was expected to be well adapted to the noise field throughout the test. After sentence onset, any adaptation was expected to be stably settled within a few hundred milliseconds after onset of the front-facing target (sentences were approximately 2 s each). MB is adaptive up to approximately 6 kHz, and static above.

BB (StereoZoom) is a fixed, four-microphone, binaural beamformer that combines the signals of the two microphones on the CI speech processor with the 2 signals from the HA via a bidirectional wireless streaming link (Buechner et al., 2014; Hehrmann et al., 2012). MB and BB both rely on the detection of delays of sounds arriving at the microphones (Ricketts, 2001). BB combines the signals from the CI and HA and is better able to detect delays between the microphones with better spatial selectivity than MB, especially at low frequencies. BB uses the contralateral audio input up until approximately 6 kHz and is monaural above.

SNRA (ClearVoice/NoiseBlock) is designed to reduce steady-state background noise in a target signal of fluctuating speech. It operates by decreasing the gain in the frequency bands (16 on the CI, 20 on the HA) where the output is predominantly stationary, as based on SNR estimates (Buechner et al., 2010). The SNR is estimated using the last several seconds of the incoming sound (Advanced Bionics Corporation, 2012). The algorithm performs best when a speech signal is presented in a background of stationary noise, such as LTSS, traffic or vacuum cleaner noise. In fluctuating noise, the algorithm's performance drops, because the difference between speech and noise cannot easily be

detected (Holden et al., 2013). SNRA was applied at 'medium' settings (up to 12 dB attenuation per frequency band).

Fitting of the algorithms on the experimental processors was always symmetric, i.e., MB and BB were fitted on both the CI and HA. MB and BB operate similarly in the CI and HA. In case of SNRA, ClearVoice was programmed on the CI speech processor, and the acoustic counterpart NoiseBlock on the HA.

Test Environment. Testing was performed in a sound-treated audiology room measuring $3.4 \times 3.2 \times 2.4$ m ($l \times w \times h$). Speech was presented through a loudspeaker (MSP5A monitor speaker, Yamaha Corp., Japan) placed in front of the head at eye level of the study participant at approximately 1 meter, and 1.2 meters from the floor. The distance between loudspeaker and participant was well within the reverberation distance of the room, being 2 m or more for frequencies of 500 Hz and higher. The room and setup has been described in more detail previously (Van der Beek et al., 2007). Participants were instructed to maintain their gaze on the loudspeaker and to keep their head still.

The homogeneous noise field was generated by 8 loudspeakers (Control 1, JBL Corp., Los Angeles, CA) that were evenly distributed on the walls of a sound attenuated audiometry chamber (Figure 2). Four loudspeakers were mounted in the 4 top corners of the booth, and the other 4 were situated close to the floor in-between the others. For more details on the noise setup, we refer to Van der Beek et al. (2007) and Taal et al. (2016). The homogeneous LTSS noise was created by playing it back over the 8 loudspeakers using random offsets to ensure they were uncorrelated. The 8-talker babble was generated from the 2-talker male ICRA babble noise (Dreschler et al., 2001). One of the channels of the ICRA noise was played back over 4 loudspeakers, and the other through the remaining 4 loudspeakers, again ensuring that all channels were uncorrelated. For more details on the setup, the LTSS noise and 8-talker babble we refer to Stronks et al. (2020).

Speech-Recognition Testing. The primary outcome measure was the speech reception threshold (SRT), i.e., the SNR where 50% correct word score was obtained. The SRT was determined with the Dutch/Flemish Matrix test (Luts et al., 2014) administered with the APEX 3 software platform (Francart et al., 2008). The speech corpus consisted of sentences with a fixed syntax, namely a name, verb, amount, color, and object. The words were drawn from a closed set of 50 words (10 names, 10 verbs etc.) and voiced by a Flemish female speaker. An example of a sentence (translated to English) is: "Emma has two black bicycles". Each run consisted of 20 sentences. After sentence presentation, participants repeated each sentence verbally. The researcher scored each correct word manually on a computer. Participants were encouraged to repeat every word, and

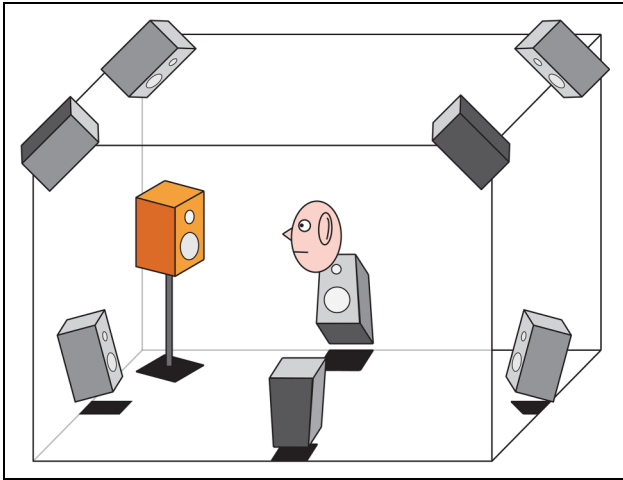


Figure 2. Loudspeaker configuration used to generate a homogenous noise field of long-term speech-spectrum-shaped (LTSS) noise or 8-talker babble. Speech was presented in front of the listener, i.e., at an angle of 0° .

guessing was allowed. No feedback on their performance during testing was provided.

To determine the SRT, the speech level was adaptively varied using a staircase procedure. The noise was presented at a constant level of 60 dBA. The step size of the staircase was dynamically decreased after each reversal. The step size reduction after a reversal depended on the number of reversals and the correct score in the previous trial (Francart et al., 2008). Typically, the speech level was varied by several dB in the first few trials of the run, while in the last few trials the variation was typically approximately 0.1 dB. The SRT was determined by calculating the average SNR across the last 6 trials. The first two tests of a session were practice runs to minimize training effects. They were performed bimodally with omnidirectional microphone settings.

Noise Stimuli. Two noise types were used, namely LTSS noise based on the speech material (Luts et al., 2014) and 8-talker male babble extracted from the ICRA noise files, both described in detail previously (Stronks et al., 2020). ICRA babble is derived from speech material that has been made unintelligible by filtering procedures. As a result, it does not lead to informational masking, but its temporal fine structure, envelope and periodicity still resemble those of speech (Dreschler et al., 2001). The LTSS and ICRA babble noise have differing spectral properties (Stronks et al., 2020), which may influence the effectiveness of MB and BB. The background noise was continuously presented, rather than intermittently together with sentence presentation. This ensured that the automatic gain control (AGC) was stably set and that SNRA was active throughout sentence presentation. The AGC in Advanced Bionics devices is a dual-

loop broadband system, but its gain is usually dominated by the slow-acting compressor that has attack and release time constants of 139 ms and 383 ms, respectively (Dwyer et al., 2021). SNRA has an activation time of 1.3 s (Holden et al., 2013). Because Matrix sentences last but a second or two, these activation times can considerably impact audibility.

LTSS noise was calibrated with a free-field sound level meter (Rion NA-28, Rion Co. Ltd., Tokyo, Japan) placed approximately at ear level. For the homogeneous noise field, the eight loudspeakers were calibrated separately to yield a final sound level of 60 dBA when all were active at the same time. The speech was calibrated using LTSS noise on the center loudspeaker, which had the same long-term spectral and amplitude characteristics as the speech (Luts et al., 2014). The 8-talker babble was calibrated digitally by matching the long-term rms amplitude to that of the LTSS noise.

KEMAR Manikin Recordings. The physical SNR benefits of MB and BB were measured in the homogeneous noise setup as used for speech testing by recording the output of a Q90 speech processor mounted on a KEMAR manikin (Burkhard & Sachs, 1975) using stimuli of 60 dBA. The speech processor was connected to a digital oscilloscope (SmartScope, Antwerp, Belgium) using a ListeningCheck™ module (Advanced Bionics, LLC, Valencia, CA, USA). SNR measurements were performed on the CI speech processor, but not on the HA, because the effects of MB and BB on SNR were assumed to be equal. Small differences in beamformer effectivity between the two devices cannot be entirely ruled out, however, because of minor differences in shape of the BTE units of the HA and CI. The ListeningCheck module allows for the audio signal to be listened to in a relatively early stage of the speech processing chain, namely after the pre-emphasis and beamforming, but before the adaptive gain control and filter banks. Therefore, SNRA could not be investigated by this method, because it operates on the envelopes of the signal.

Because LTSS and 8-talker babble noise had differing spectral properties, as shown in Figure 3 of Stronks et al. (2020), the physical effects of MB and BB on the SNR were determined for both LTSS noise and 8-talker babble separately. The frontally delivered speech signal was substituted by LTSS noise. The output levels of the CI speech processor to the signal and noise were recorded separately to determine SNRs. Recordings were bandpass filtered with a digital, 6th order Butterworth filter (MATLAB v. 2020a, MathWorks Inc., Natick, MA, USA) with cut-off frequencies matching those of the CI speech processor (350–8700 Hz). The filter roll-off (30 dB/octave) was similar to that of the CI speech processor (40 dB/octave). To extrapolate SNR benefits to SRT improvements, we applied articulation index (AI) weighting by converting the recorded audio into

17 one-third octave bands using inverse FFT filtering. The bands were AI-weighted according to the ‘Count-The-Dots’ method from Killion & Mueller (2010) by using the number of dots in each third-octave band as weighting factor. The number of dots is an integer representation of the relative importance of that band for speech recognition of normal hearing and aided listeners and was applied in a linear fashion to the narrowband audio signals. The band importance functions of CI users, however, shows substantial between-subject variability and CI users may rely more heavily on low-frequency bands than normal-hearing listeners (Bosen & Chatterjee, 2016). After AI-weighting, the SNR was calculated from the reconstituted broadband signals using $20 \cdot \log_{10}(\text{rms}_{\text{signal}}/\text{rms}_{\text{noise}})$. The effect of AI weighting was limited to no more than several tenths of a dB.

Polar patterns of the different microphone settings were created in a similar fashion by mounting a CI speech processor on the right ear of a KEMAR manikin and rotating it in 15° steps around its axis. The acoustic output of the speech processor was recorded with and without beamforming in response to pink noise presented at 80 dB SPL from a loudspeaker placed 1 meter away from KEMAR’s head. Microphone output after beamforming, but before pre-emphasis and automatic gain control, was recorded using the research tool BEPSnet (Advanced Bionics, LLC, Valencia, CA, USA). The recordings were performed by Advanced Bionics, LLC in an anechoic chamber. From the broadband polar patterns, a measure of directivity was obtained using equations (3) and (4) from Chung & Zeng (2009). This measure equals the average sound pressure level across the measured locations in the polar pattern of the beamformer divided by that of the reference omnidirectional microphone setting and is expressed in dB. This measure is different from the directivity index. Rather, it is a measure of how much more directive the beamformers were than the omnidirectional setting.

For the KEMAR recordings, a static version of MB without null steering was used, instead of the clinical, adaptive algorithm used for participant testing, because signal and noise were recorded separately during the SNR recordings with KEMAR. This sequential recording was anticipated to result in unexpected behavior of the clinical adaptive MB algorithm, because it cannot unambiguously establish the region with the lowest SNR. The static representation of the algorithm (Advanced Bionics, LLC, Valencia, CA, USA) had its null directed at 120°, which was expected to be the most efficient orientation based on the directivity index in a homogeneous broadband noise field with speech presented frontally. The adaptive variant of MB was used during psychophysical testing, because we wished to obtain a clinically relevant evaluation of the benefits of the algorithm.

Figure 3A shows the broadband polar patterns of the processor microphone at omnidirectional setting and of the beamformers recorded from the right ear of KEMAR.

Figure 3B to F show the corresponding narrowband patterns, illustrating that BB is more effective than the static version of MB, particularly at low frequencies. The shapes of the polar patterns differ between the two beamformers as well; BB is most effective when speech comes from the front and noise from the sides, whereas MB is most effective when speech comes from 45° to the CI side and noise from the back. In this study, speech was presented from the front, and we expected BB to be more effective than MB.

Statistical Testing. Statistics were carried out in SPSS 23 for Windows (IBM Corp., Armonk, NY) and Prism 7.04 (GraphPad Software, La Jolla, CA). The effectiveness of the beamformers and SNRA was tested using two linear mixed models (LMM). LMMs are an extension of simple linear models such as ANOVA that allow the inclusion of both fixed and random effects, and they are particularly useful for the analysis of repeated measures (Brauer & Curtin, 2018).

The first LMM was designed to test 1) whether the SRT benefits of beamforming differ when measured in a background of homogeneous LTSS noise or 8-talker babble, (2) whether BB is more effective than MB in LTSS noise and 8-talker babble, and (3) whether SNRA improves the SRT in LTSS noise. The SRT was entered as the dependent variable, and the fixed main effects were: (a) microphone setting (mic) with 3 levels (omnidirectional, MB, BB), (b) SNRA with 2 levels (yes/no), (c) noise type (LTSS noise or 8-talker babble), and (d) session number, which was entered as an ordinal variable with 4 levels (1 to 4). Two interaction factors were included in the model to test the effect of noise type on microphone setting, namely [mic · noise type], and [SNRA · noise type]. Trial number was entered as a fixed, ordinal covariate, adding up to a total of 7 factors in the LMM. Session and trial number were included to investigate learning effects and fatigue. Participant ID was entered as a categorical random variable. For the fixed and random effects, an estimation of the intercept was included. Covariance type was set at ‘unstructured’ for the fixed effects, and ‘identity’ for the random effect. All LMM settings were left at their defaults (using SPSS version 23), except when explicitly stated otherwise. As a control, we tested whether trial and session number affected SRT *differences* obtained with the beamformers or SNRA. The model was identical to the first, but SRT differences were entered as the dependent variable, which were calculated by subtracting the omnidirectional control condition (no beamformer, no SNRA) from the conditions where a beamformer and/or a SNRA was used.

The last research question, namely 4) whether the SRT benefits of MB and BB are comparable to the SNR improvement, was investigated by comparing the AI-weighted SNR benefits obtained with KEMAR with the effect of the beamformers on the SRT. The differences were tested for significance using Bonferroni-corrected one-sample *t*-tests.

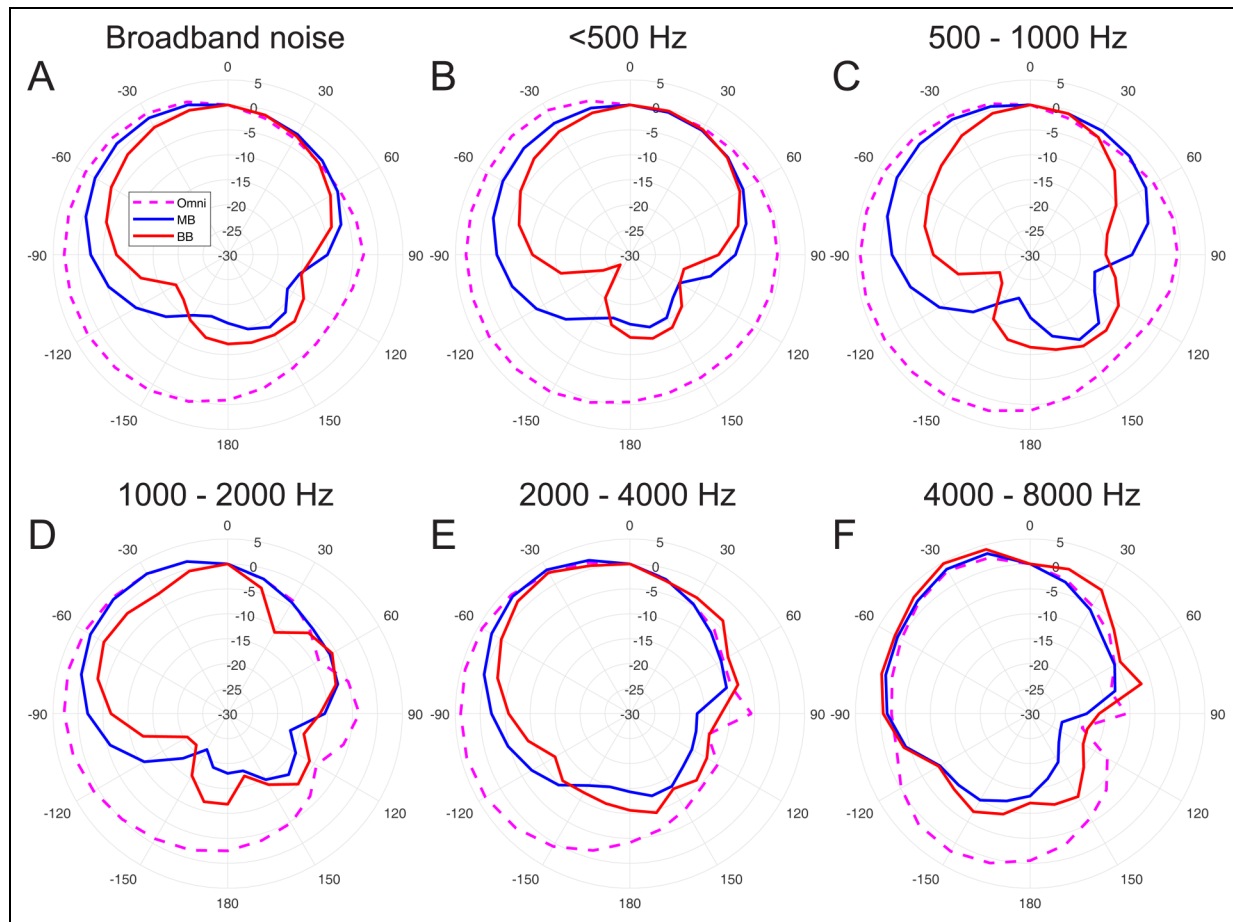


Figure 3. Polar patterns of the different microphone settings. Broadband (pink noise, A) and narrowband (B-F) polar patterns for the omnidirectional microphones (omni, hashed purple line), monaural beamformer (MB, solid blue line) and binaural beamformer (BB, solid red line). Concentric circles indicate the gain in dB relative to the frontal 0°.

Results

Figure 4A and B show the average SRTs obtained in both noise types, and Figure 4C and D show the benefits on the SRT of the beamformers and SNRA relative to the omnidirectional condition without any noise reduction algorithm. MB and BB substantially improved the SRTs, and these benefits appeared to be similar between MB and BB, and between noise types. By contrast, SNRA seemed to have little effect. To statistically test these observations, an LMM was built with SRTs as outcome measure. To verify that a quantitative analysis was appropriate, a normal distribution of the SRT and SRT difference was first confirmed (D'Agostino & Pearson test on the pooled data, $P=0.126$ and 0.178 , respectively). The LMM showed significant overall effects of beamforming ($F_{(2,330)}=146$, $P < 0.001$) and noise type ($F_{(1,330)}=1344$, $P < 0.001$), but not of SNRA ($F_{(1,330)}=0.268$, $P=0.605$). The two interaction factors [*mic · noise type*] and [*SNRA · noise type*] did not significantly affect SRTs ($F_{(2,330)}=0.83$, $P=0.436$, and $F_{(1,330)}=0.008$, $P=0.930$, respectively).

Parameter estimates of the LMM and *post hoc* testing revealed that 8-talker babble resulted in 6.2 dB higher SRTs overall than LTSS noise ($t_{(333)}=36.7$, $P < 0.001$). The effectiveness of MB across conditions (2.6 dB) did not differ significantly from that of BB (2.9 dB, $t_{(330)}=1.11$, $P=0.269$). These LMM-based estimated benefits of MB and BB differ slightly from the arithmetic averages of the raw data depicted in Figure 4C and D, because the reported LMM values represent the overall effect across conditions, taking the various fixed and random factors included in the model into account. The LMM estimates can, therefore, be considered as superior representations of the estimated population means.

The difference of the recorded AI-weighted SNR improvements was negligible when comparing them across noise types (solid green lines), but SNR improvements differed substantially between MB and BB (3.3 and 6.6 dB, respectively, averaged across homogeneous LTSS noise and 8-talker babble). The measure of directivity as used by Chung & Zeng (2009) derived from the unweighted SNRs of the polar patterns resembled the AI-weighted SNR

improvements reasonably well (4.8 dB and 6.4 dB, respectively). To address the last research question, namely whether SNR and SRT benefits matched, one sample *t*-tests were carried out to determine whether the SRT benefits obtained with the beamformers differed significantly from the SNR reduction recorded with the KEMAR manikin (Figure 4C and D). The *t*-tests were carried out using the raw SRT benefit data (i.e., uncorrected for any of the factors from the LMM), without SNRA, and *p* values were Bonferroni corrected to compensate for the four comparisons made. The SRT benefit of MB in LTSS noise and 8-talker babble (arithmetic mean: 2.9 and 2.7 dB, respectively), did not significantly differ from the AI-weighted SNR benefits of 3.1 dB ($t_{14}=0.560$, $p=1$ after Bonferroni correction) and 3.4 dB ($t_{14}=2.322$, $p=0.143$), respectively. By contrast, the SRT improvement of BB in LTSS noise and 8-talker babble (2.8 and 3.2 dB, respectively) were significantly less than the corresponding SNR benefits in both LTSS and babble noise, i.e., 6.6 dB in both noise types ($t_{14}=10.580$, $p < 0.0001$ and $t_{14}=11.121$, $p < 0.0001$, respectively).

A hypothetical reason for the ineffectiveness of SNRA in our study was that the algorithm may not have been able to extract the speech signal from the noise at low SNRs. To investigate this, the SRT effect of SNRA ($SRT_{SNRA} - SRT_{omni}$) needs to be correlated with the SRT without SNRA (SRT_{omni}). Because SRT_{omni} is present in both factors, the terms are mathematically coupled. To uncouple the data, the effect of SNRA was correlated to the average SRT $(SRT_{SNRA} + SRT_{omni})/2$ instead (Carlyon et al., 2018; Stronks et al., 2021). Only SRTs obtained in LTSS noise and omnidirectional microphone settings (i.e., no beamforming) were used. Overall SRTs varied from -10 to 0 dB. No significant correlation was found ($r^2=0.073$, $F(1,28)=2.19$, $P=0.150$).

Within-session learning effects were significant ($F_{(1,330)}=10.2$, $P=0.00156$), and each subsequent trial resulted in an average SRT decrease of 0.08 dB, on average. Between-session learning effects were significant as well ($F_{(3,330)}=43.1$, $P < 0.001$), and were most pronounced between the 1st and 2nd session, with an overall improvement of approximately 1.5 dB. Additional improvements of 0.5 and 0.7 dB were seen in the 3rd and 4th session, respectively. Conditions were randomized across trials and sessions. Any remaining learning effects will have been corrected for by the inclusion of these factors in the LMM.

A second LMM was built to investigate whether the observed learning effects were also evident on the effects of beamforming and SNRA. The LMM was identical to the first, except that the SRT differences relative to the omnidirectional control condition were used as dependent variable. In contrast to the first LMM, this LMM did not reveal significant effects of either trial or session number ($F_{(1,276)}=0.23$, $P=0.630$ and $F_{(3,275)}=1.88$, $P=0.134$ respectively).

Discussion

In this study, the effectiveness of MB, BB and SNRA were tested in a group of bimodal listeners using homogeneous fields of LTSS noise or 8-talker babble. We have shown that MB and BB statistically significantly improved the SRT across both noise types. There was no significant effect of noise type on the SRT benefit, and there was no significant difference between the performance of MB and BB (2.6 and 2.9 dB SRT benefit across both noise types for MB and BB, respectively). By contrast, KEMAR recordings revealed a substantially higher SNR improvement by BB (6.6 dB) than MB (3.3). No significant effect of SNRA on SRTs was found. On the basis of these results, we conclude: (1) that SRT benefits of beamforming do not significantly differ between LTSS noise and 8-talker babble, (2) that there is no significant difference between the benefit of MB and BB on the SRT, (3) that the SNR increase of MB approximates the SRT improvement, but that BB's effect on the SNR is significantly larger than its effect on the SRT, and lastly (4) that SNRA does not improve speech recognition in either noise type by itself, nor in combination with a beamforming algorithm.

SRTs and SNRs were obtained by using the processor mic on the BTE unit. Most CI recipients with a Naída Q processor use a Tmic instead when listening in omnidirectional microphone setting (Frohne-Büchner et al., 2004). The Tmic is placed near the entrance of the ear canal and therefore it benefits from the directional shaping of sound by the pinna (Aronoff et al., 2011; Korhonen, 2013). This “pinna-shadow effect” has been shown to improve SRTs in diffuse noise (Gifford & Revit, 2010). Because MB and BB depend on the processor microphones that are situated on top of the BTE unit, they do not benefit from the pinna shadow effect (Festen & Plomp, 1986). KEMAR recordings performed in our lab in a homogeneous noise field showed that the Tmic improves the AI-weighted SNR by approximately 0.3 dB in LTSS noise. Thus, the reported benefits in the present study are expected to be approximately 0.3 dB lower when the CI would have been sourced by the Tmic.

The SNR recordings and directivity estimates in the current study both revealed a substantially better performance of BB over MB, but the SRT measurements did not. The SNR and SRT benefits of MB matched reasonably, but the SRT improvement with BB was substantially less than expected. Given that SNR measures have previously been shown to be accurate predictors for SRT benefits of beamformers in HAs (Soede et al., 1993) and CIs (Buechner et al., 2014), the discrepancy between the two measures for BB is unexpected. A possible explanation may have been that the study participants invested less effort under conditions of improved SNR (Sarampalis et al., 2009), which may have effectively counteracted BB's additional SNR benefit. In addition, the mixing of the two monaural signals by BB

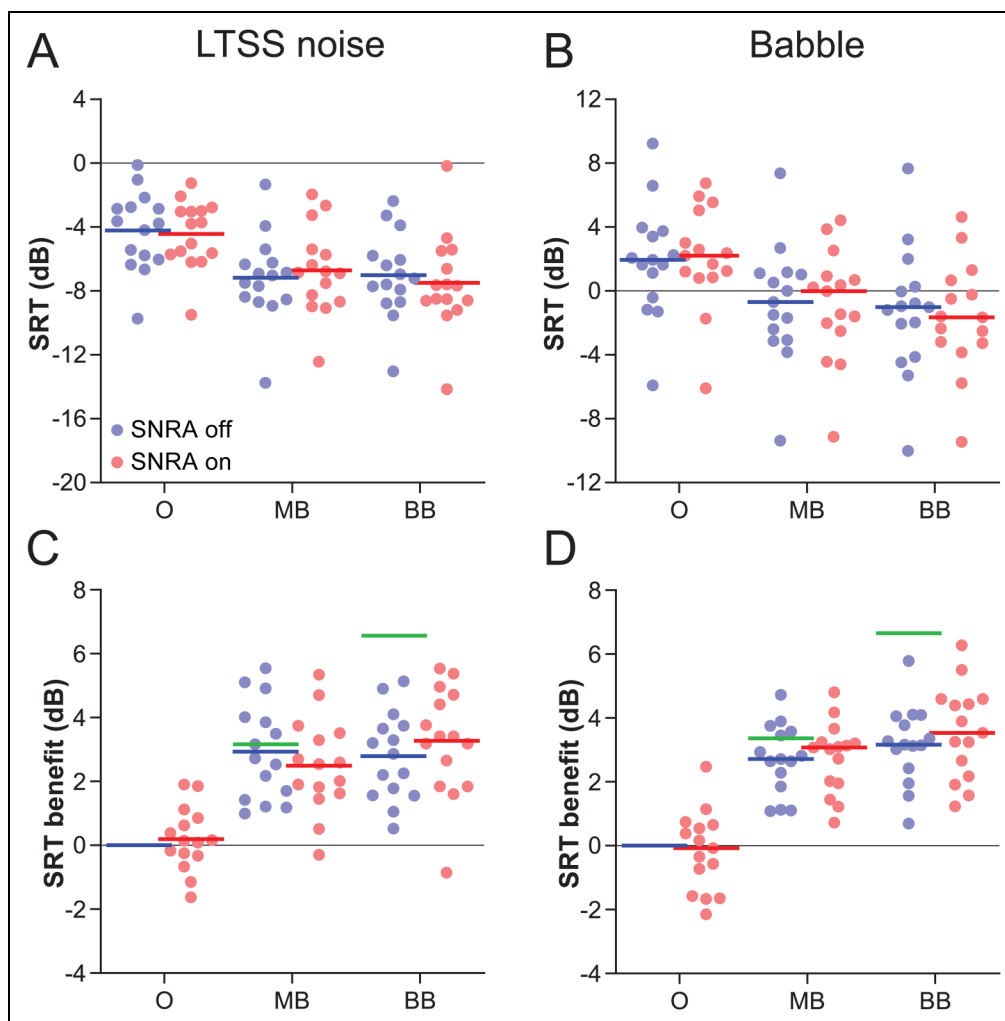


Figure 4. Scatter plots of SRTs and SRT differences obtained with different microphone settings and different noise types. SRTs obtained in long-term speech-spectrum-shaped (LTSS) noise (panel A) and in 8-talker babble when using omnidirectional microphone settings (O), monaural beamforming (MB), binaural beamforming (BB), and/or a single-microphone noise reduction algorithm (SNRA) on (red dots) or off (blue dots), and the corresponding benefits (panels C and D, positive values correspond to a benefit, negative to a disadvantage). Horizontal blue and red lines: averages; horizontal green lines: SNR reductions after articulation index-based weighting.

has resulted in lower SRT benefits by reducing complementary TFS cues at low frequencies or eliminating binaural cues (Morera et al., 2005). Whether bimodal listeners actually utilize binaural cues is, however, contested (Dieudonné & Francart, 2020). It is also possible that the use of a static variant of MB for the KEMAR recordings has underestimated the SNR benefit, especially in 8-talker babble. The clinical variant could have steered its null to the direction with the most dominant interference, resulting in a smaller difference between the SNR effects of both beamformers, i.e., more conform the SRT findings. Another factor that may have contributed is the use of variable speech levels to determine the SRT at a constant noise level of 60 dBA. Because SRTs were predominantly negative, beamforming

may have pushed speech levels down to the point where speech recognition abilities become negatively affected (Donaldson & Allen, 2003).

Another reason behind the lack of difference between the two beamformers on SRTs in the current study was the specific noise setup used. Specific locations of the noise sources can yield conditions that better favor BB over MB (Buechner et al., 2014). MB is most effective when the target speech is presented from 45° at the CI side and the noise is coming from the back, whereas BB performs optimally when speech is presented from the front and noise is directed to the sides (Figure 3). Two other studies have compared MB and BB, both in unilateral CI users, where speech was presented from the front as in this study, but the noise setups

were different (Buechner et al., 2014; Ernst et al., 2019). In Ernst et al. (2019) two setups were used where the noise sources were placed uniformly around the listener, or predominantly in the frontal hemisphere. Compared to omnidirectional microphone settings using the Tmic, MB yielded significant benefits of 3.4 dB and 1.4 dB in the two setups, respectively. BB yielded significantly higher benefits on speech recognition, namely 4.6 dB and 2.6 dB. Buechner et al. (2014) also reported significantly greater benefits of BB (7.1 dB) than MB (5.3 dB) on the SRT using noise sources located laterally and in the back hemisphere. By contrast, we were unable to show a statistically significant difference between MB and BB. In a homogeneous noise field as used in the present study, a substantial amount of noise came from the same direction as the target speech, which remains unattenuated by MB or BB (Figure 3). As a result, the setup used in our study is not well suited to show differences in effectivity between MB and BB. The reason we nonetheless used it is that it, arguably, better reflects real-world noisy conditions, because noise may also originate behind the target speaker, or from any other direction.

DeVocht et al. (2016) investigated MB and compared LTSS noise and babble in bimodal listeners, as in the present study. MB was implemented on both the CI and HA. The noise was presented diffusely, with loudspeakers positioned in the frontal and back hemisphere. The reported SRT benefits were hence relatively small and comparable to ours, namely 2.7 dB in LTSS noise, and 2.6 dB in 8-talker babble, using an omnidirectional microphone (Tmic) as reference. In line with our findings, the difference between noise types was not statistically significant.

We did not find a statistically significant effect of SNRA on SRTs, in line with a previous study (Dingemanse & Goedegebure, 2015). However, others have reported benefits (Buechner et al., 2010; Buechner et al., 2014; Ernst et al., 2019; Kam et al., 2012; Noël-Petroff et al., 2013), or mixed results with high inter-individual variability (Dingemanse & Goedegebure, 2018; Holden et al., 2013). It has been reported that for SNRA to be effective, threshold (T) and maximum comfort levels (M) need to be adjusted (Dingemanse & Goedegebure, 2018; Noël-Petroff et al., 2013), because overall sound levels are decreased by the algorithm. Unfortunately, it has never been investigated whether the reported improvement of speech understanding after adjustment of the T and M levels are due to the combination of SNRA and the raising of M and T levels, or whether the adjustment of the M and T levels alone are enough to improve speech understanding.

The effect of SNRA on SRT was independent on the SRT in our study. However, overall SRT levels were relatively low in our study population, averaging to approximately -4 dB without beamforming in LTSS noise (Figure 4A). The associated SNRs may have been too low for SNRA to work. It is imaginable that the algorithm can effectively distinguish a fluctuating speech signal from steady-state

background noise at favorable SNRs, but that it becomes less effective when SNRs deteriorate. Average SNRs used in studies that reported benefits of the same SNRA were higher than ours overall, ranging from -3 to $+5$ dB (Buechner et al., 2010; Buechner et al., 2014; Ernst et al., 2019; Kam et al., 2012; Noël-Petroff et al., 2013). However, SNRs in studies reporting marginal, mixed, or no benefits were also much higher than ours, *i.e.*, approximately $+5$ dB (Dingemanse & Goedegebure, 2015, 2018; Holden et al., 2013) and we conclude that SNR alone cannot explain the ineffectiveness of SNRA in the present study. All the studies discussed above, including those reporting benefits and those that did not, deployed SNRA set at Medium or High. Thus, the setting of SNRA cannot easily explain the discrepancies between studies either. Possibly other variables, such as the speech material and type of masker may have affected study outcomes. Dingemanse & Goedegebure (2015, 2018) report marginal benefits of SNRA on speech recognition in noise under some of the conditions tested, but they found a consistent enhanced noise tolerance, *i.e.*, higher acceptable noise levels with SNRA. Noise reduction algorithms may improve SNRs, but fail to improve SRTs nonetheless, because of the introduction of speech distortion and artifacts, or because listeners simply invest less effort in the task at better SNRs. Nonetheless, often users prefer them nonetheless because they decrease cognitive load (Sarampalis et al., 2009). These results show that SNRA may not improve speech recognition, but that it may benefit the user in other cognitive dimensions not addressed here.

We included trial number as a covariate in the LMM which revealed that the SRT reduced by 0.08 dB per trial. This learning effect may not have been linear, because the first few trials are expected to have had a greater learning effect than later ones, just as the between-session learning effect was most pronounced between the 1st and 2nd session. Thus, the training effect as reported here should be interpreted with caution. We do not expect learning effects to have affected our results, because all the within (beamforming, SNRA) and between-session conditions (noise type) were randomized. In addition, the second LMM, where SRT differences were entered as the dependent variable, did not reveal significant learning effects either within, or between sessions in terms of the effectiveness of beamforming and SNRA. In other words, the study participants' SRTs improved, but they did not gain more benefit from the noise reduction algorithms during and across sessions.

Conclusion

We conclude that BB does not have additional benefit on speech recognition over MB in a group of bimodal listeners in either homogenous stationary noise or multitalker babble, in contrast to what was expected based on SNR recordings.

SNRA does not seem beneficial to improve speech recognition either alone, or in combination with MB or BB. Speech recognition in 8-talker babble was substantially (6 dB) and significantly higher than in LTSS noise. Fluctuating noise is especially difficult for CI users (Nelson et al., 2003) and the TFS of babble noise as opposed to the broadband character of TSS may have additional masking effects as well (Stronks et al., 2020). We conclude that MB and BB resulted in significant and substantial benefits even in the most adverse of listening conditions.

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Declaration of Conflicting Interests

The author(s) declared no potential conflicts of interest with respect to the research, authorship, and/or publication of this article.

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References

Advanced Bionics Corporation. (2012). ClearVoice™. Technical Facts, 1–4.

Aronoff J. M., Freed D. J., Fisher L. M., Pal I., & Soli S. D. (2011). The effect of different cochlear implant microphones on acoustic hearing individuals' binaural benefits for speech perception in noise. *Ear and Hearing, 32*(4), 468–484. <https://doi.org/10.1097/AUD.0b013e31820dd3f0>

Bosen A. K., & Chatterjee M. (2016). Band importance functions of listeners with cochlear implants using clinical maps. *The Journal of the Acoustical Society of America, 140*(5), 3718–3718. <https://doi.org/10.1121/1.4967298>

Brauer M., & Curtin J. J. (2018). Linear mixed-effects models and the analysis of nonindependent data: A unified framework to analyze categorical and continuous independent variables that vary within-subjects and/or within-items. *Psychological Methods, 23*(3), 389–411. <https://doi.org/10.1037/met0000159>

Buechner A., Brendel M., Saalfeld H., Litvak L., Frohne-Buechner C., & Lenarz T. (2010). Results of a pilot study with a signal enhancement algorithm for HiRes 120 cochlear implant users. *Otology & Neurotology, 31*(9), 1386–1390. <https://doi.org/10.1097/MAO.0b013e3181f1cdc6>

Buechner A., Dyballa K.-H., Hehrmann P., Fredelake S., & Lenarz T. (2014). Advanced beamformers for cochlear implant users: acute measurement of speech perception in challenging listening conditions. *PloS One, 9*(4), e95542. <https://doi.org/10.1371/journal.pone.0095542>

Burkhard M. D., & Sachs R. M. (1975). Anthropometric manikin for acoustic research. *The Journal of the Acoustical Society of America, 58*(1), 214–222. <https://doi.org/10.1121/1.380648>

Carlyon R. P., Cosentino S., Deeks J. M., Parkinson W., & Arenberg J. G. (2018). Effect of stimulus polarity on detection thresholds in cochlear implant users: relationships with average threshold, gap detection, and rate discrimination. *Journal of the Association for Research in Otolaryngology: JARO, 19*(5), 559–567. <https://doi.org/10.1007/s10162-018-0677-5>

Chung K., & Zeng F.-G. (2009). Using hearing aid adaptive directional microphones to enhance cochlear implant performance. *Hearing Research, 250*(1), 27–37. <https://doi.org/10.1016/j.heares.2009.01.005>

Cuda D., Murri A., Mainardi A., & Chalupper J. (2019). Effectiveness and efficiency of a dedicated bimodal fitting formula. *Audiology Research, 9*(1), 219–219. <https://doi.org/10.4081/audiore.2019.219>

Devocht E. M. J., Janssen A. M. L., Chalupper J., Stokroos R. J., & George E. L. J. (2016). Monaural beamforming in bimodal cochlear implant users: effect of (A)symmetric directivity and noise type. *PloS One, 11*(8), e0160829–e0160829. <https://doi.org/10.1371/journal.pone.0160829>

Dieudonné B., & Francart T. (2020). Speech understanding with bimodal stimulation is determined by monaural signal to noise ratios: No binaural cue processing involved. *Ear and Hearing, 41*(5), 1159–1171. <https://doi.org/10.1097/AUD.0000000000000834>

Dingemans J. G., & Goedegebure A. (2015). Application of noise reduction algorithm ClearVoice in cochlear implant processing: effects on noise tolerance and speech intelligibility in noise in relation to spectral resolution. *Ear and Hearing, 36*(3), 357–367. <https://doi.org/10.1097/AUD.0000000000000125>

Dingemans J. G., & Goedegebure A. (2018). Optimising the effect of noise reduction algorithm ClearVoice in cochlear implant users by increasing the maximum comfort levels. *International Journal of Audiology, 57*(3), 230–235. <https://doi.org/10.1080/14992027.2017.1390267>

Donaldson G. S., & Allen S. L. (2003). Effects of presentation level on phoneme and sentence recognition in quiet by cochlear implant listeners. *Ear and Hearing, 24*(5). <https://doi.org/10.1097/01.AUD.0000090340.09847.39>

Dreschler W. A., Verschuure H., Ludvigsen C., & Westermann S. (2001). ICRA Noises: Artificial noise signals with speech-like spectral and temporal properties for hearing instrument assessment. *International collegium for rehabilitative audiology. Audiology, 40*(3), 148–157. <https://doi.org/10.3109/00206090109073110>

Dwyer R. T., Chen C., Hehrmann P., Dwyer N. C., & Gifford R. H. (2021). Synchronized automatic gain control in bilateral cochlear implant recipients yields significant benefit in static and dynamic listening conditions. *Trends in Hearing, 25*, 1–14. <https://doi.org/10.1177/23312165211014139>

Elko G. W., & Anh-Tho Nguyen P. (1995, 15–18 Oct. 1995). A simple adaptive first-order differential microphone. Paper presented at the Proceedings of 1995 Workshop on Applications of Signal Processing to Audio and Acoustics.

Ernst A., Anton K., Brendel M., & Battmer R.-D. (2019). Benefit of directional microphones for unilateral, bilateral and bimodal cochlear implant users. *Cochlear Implants International, 20*(3), 147–157. <https://doi.org/10.1080/14670100.2019.1578911>

- Festen J. M., & Plomp R. (1986). Speech-reception threshold in noise with one and two hearing aids. *Journal of the Acoustical Society of America*, 79(2), 465–471. <https://doi.org/10.1121/1.393534>
- Francart T., van Wieringen A., & Wouters J. (2008). APEX 3: A multi-purpose test platform for auditory psychophysical experiments. *Journal of Neuroscience Methods*, 172(2), 283–293. <https://doi.org/10.1016/j.jneumeth.2008.04.020>
- Frohne-Büchner C., Büchner A., Gärtner L., Battmer R. D., & Lenarz T. (2004). Experience of uni- and bilateral cochlear implant users with a microphone positioned in the pinna. *International Congress Series*, 1273, 93–96. <https://doi.org/10.1016/j.ics.2004.08.047>
- Fu Q.-J., & Nogaki G. (2004). Noise susceptibility of cochlear implant users: the role of spectral resolution and smearing. *Journal of the Association for Research in Otolaryngology*, 6(1), 19–27. <https://doi.org/10.1007/s10162-004-5024-3>
- Gannot S., & Cohen I. (2008). Adaptive beamforming and postfiltering. In J. Benesty, M. M. Sondhi, & Y. A. Huang (Eds.), *Springer handbook of speech processing* (pp. 945–9978). Springer Berlin Heidelberg. https://doi.org/10.1007/978-3-540-49127-9_47
- Geißler G., Arweiler I., Hehrmann P., Lenarz T., Hamacher V., & Büchner A. (2015). Speech reception threshold benefits in cochlear implant users with an adaptive beamformer in real life situations. *Cochlear Implants International*, 16(2), 69–76. <https://doi.org/10.1179/1754762814Y.0000000088>
- Gifford R. H., & Revit L. J. (2010). Speech perception for adult cochlear implant recipients in a realistic background noise: Effectiveness of preprocessing strategies and external options for improving speech recognition in noise. *Journal of the American Academy of Audiology*, 21(7), 441–451. <https://doi.org/10.3766/jaaa.21.7.3>
- Gustafson S., McCreery R., Hoover B., Kopun J. G., & Stelmachowicz P. (2014). Listening effort and perceived clarity for normal hearing children with the use of digital noise reduction. *Ear and Hearing*, 35(2), 183–194. <https://doi.org/10.1097/01.aud.0000440715.85844.b8>
- Hehrmann P., Fredelake S., Hamacher V., Dyballa K. H., & Buechner A., (2012, Sept). Improved speech intelligibility with cochlear implants using state-of-the-art noise reduction algorithms. Paper presented at the ITG Symposium; Speech Communication., Braunschweig, Germany.
- Holden L. K., Brenner C., Reeder R. M., & Firszt J. B. (2013). Postlingual adult performance in noise with HiRes 120 and ClearVoice low, Medium, and high. *Cochlear Implants International*, 14(5), 276–286. <https://doi.org/10.1179/1754762813Y.0000000034>
- Hopkins K., & Moore B. C. J. (2009). The contribution of temporal fine structure to the intelligibility of speech in steady and modulated noise. *The Journal of the Acoustical Society of America*, 125(1), 442–446. <https://doi.org/10.1121/1.3037233>
- Kam A. C. S., Ng I. H. Y., Cheng M. M. Y., Wong T. K. C., & Tong M. C. F. (2012). Evaluation of the ClearVoice strategy in adults using HiResolution fidelity 120 sound processing. *Clinical and Experimental Otorhinolaryngology*, 5(Suppl 1), S89–S92. <https://doi.org/10.3342/ceo.2012.5.S1.S89>
- Killion M. C., & Mueller H. G. (2010). Twenty years later: A NEW count-the-dots method. *The Hearing Journal*, 63(1), 10–17. <https://doi.org/10.1097/01.HJ.0000366911.63043.16>
- Korhonen P. (2013). Horizontal localization with pinna compensation algorithm and inter-ear coordinated dynamic-range compression. Paper presented at the 4th symposium on Auditory and Audiological Research, Nyborg, Denmark.
- Landsberger D. M., Svrakic M., Roland J. T.Jr., & Svirsky M. (2015). The relationship between insertion angles, default frequency allocations, and spiral ganglion place pitch in cochlear implants. *Ear and Hearing*, 36(5), e207–e213. <https://doi.org/10.1097/AUD.000000000000163>
- Luts H., Jansen S., Dreschler W., & Wouters J. (2014). Development and normative data for the flemish/Dutch matrix test. *Katholieke universiteit Leuven, Belgium and Academic Medical Center Amsterdam, The Netherlands., Unpublished article.*
- Mok M., Grayden D., Dowell R. C., & Lawrence D. (2006). Speech perception for adults who use hearing aids in conjunction with cochlear implants in opposite ears. *Journal of Speech, Language, and Hearing Research : JSLHR*, 49(2), 338–351. [https://doi.org/10.1044/1092-4388\(2006/027\)](https://doi.org/10.1044/1092-4388(2006/027))
- Morera C., Manrique M., Ramos A., Garcia-Ibanez L., Cavalle L., & Huarte A.,Castillo, C., & Estrada E. (2005). Advantages of binaural hearing provided through bimodal stimulation via a cochlear implant and a conventional hearing aid: A 6-month comparative study. *Acta Oto-Laryngologica*, 125(6), 596–606. <https://doi.org/10.1080/00016480510027493>
- Nelson P. B., Jin S.-H., Carney A. E., & Nelson D. A. (2003). Understanding speech in modulated interference: cochlear implant users and normal-hearing listeners. *The Journal of the Acoustical Society of America*, 113(2), 961–968. <https://doi.org/10.1121/1.1531983>
- Noël-Petroff N., Mathias N., Ulmann C., & Abbeele T. V. D. (2013). Pediatric evaluation of the ClearVoice™ speech enhancement algorithm in everyday life. *Audiology Research*, 3(1), e9. <https://doi.org/10.4081/audiores.2013.e9>
- Oxenham A. J., & Simonson A. M. (2009). Masking release for low- and high-pass-filtered speech in the presence of noise and single-talker interference. *The Journal of the Acoustical Society of America*, 125(1), 457–468. <https://doi.org/10.1121/1.3021299>
- Qin M. K., & Oxenham A. J. (2003). Effects of simulated cochlear-implant processing on speech reception in fluctuating maskers. *The Journal of the Acoustical Society of America*, 114(1), 446–454. <https://doi.org/10.1121/1.1579009>
- Ricketts T. A. (2001). Directional hearing aids. *Trends in Amplification*, 5(4), 139–176. <https://doi.org/10.1177/108471380100500401>
- Sarampalis A., Kalluri S., Edwards B., & Hafter E. (2009). Objective measures of listening effort: effects of background noise and noise reduction. *Journal of Speech, Language, and Hearing Research*, 52(5), 1230–1240. [https://doi.org/10.1044/1092-4388\(2009/08-0111\)](https://doi.org/10.1044/1092-4388(2009/08-0111))
- Shannon R. V., Cruz R. J., & Galvin J. J. (2011). Effect of stimulation rate on cochlear implant Users' phoneme, word and sentence recognition in quiet and in noise. *Audiology and Neurotology*, 16(2), 113–123. <https://doi.org/10.1159/000315115>
- Soede W., Bilsen F. A., & Berkhout A. J. (1993). Assessment of a directional microphone array for hearing-impaired listeners. *The Journal of the Acoustical Society of America*, 94(2), 799–808. <https://doi.org/10.1121/1.408181>

- Spriet, A., Van Deun, L., Eftaxiadis, K., Laneau, J., Moonen, M., Van Dijk, B., Van Wieringen, A., & Wouters, J. (2007). Speech understanding in background noise with the two-microphone adaptive beamformer BEAM™ in the nucleus freedom™ cochlear implant system. *28*(1), 62–72. <https://doi.org/10.1097/01.aud.0000252470.54246.54>
- Stakhovskaya O., Sridhar D., Bonham B. H., & Leake P. A. (2007). Frequency map for the human cochlear spiral ganglion: Implications for cochlear implants. *Journal of the Association for Research in Otolaryngology : JARO*, *8*(2), 220–233. <https://doi.org/10.1007/s10162-007-0076-9>
- Stronks H. C., Briaire J. J., & Frijns J. H. M. (2020). The temporal fine structure of background noise determines the benefit of bimodal hearing for recognizing speech. *Journal of the Association for Research in Otolaryngology*, *21*(6), 527–544. <https://doi.org/10.1007/s10162-020-00772-1>
- Stronks H. C., Tops A. L., Hehrmann P., Briaire J. J., & Frijns J. H. M. (2021). Personalizing transient noise reduction algorithm settings for cochlear implant users. *Ear and Hearing*, *42*(6), 1602–1614. <https://doi.org/10.1097/AUD.0000000000001048>
- Taal C., van Barneveld D., Soede W., Briaire J., & Frijns J. (2016). Benefit of contralateral routing of signals for unilateral cochlear implant users. *The Journal of the Acoustical Society of America*, *140*(1), 393–401. <https://doi.org/10.1121/1.4955307>
- Turner C. W., Gantz B. J., Vidal C., Behrens A., & Henry B. A. (2004). Speech recognition in noise for cochlear implant listeners: benefits of residual acoustic hearing. *The Journal of the Acoustical Society of America*, *115*(4), 1729–1735. <https://doi.org/10.1121/1.1687425>
- Van der Beek F. B., Soede W., & Frijns J. H. M. (2007). Evaluation of the benefit for cochlear implantees of two assistive directional microphone systems in an artificial diffuse noise situation. *Ear and Hearing*, *28*(1), 99–110. <https://doi.org/10.1097/AUD.0b013e31802d0a55>
- Van Hoesel R. J., & Clark G. M. (1995). Evaluation of a portable two-microphone adaptive beamforming speech processor with cochlear implant patients. *Journal of the Acoustical Society of America*, *97*(4), 2498–2503. <https://doi.org/10.1121/1.411970>
- Voss S. C., Pichora-Fuller M. K., Ishida I., Pereira A., Seiter J., El Guindi N., Kuehnel V., & Qian, J. (2021). Evaluating the benefit of hearing aids with motion-based beamformer adaptation in a real-world setup. *International Journal of Audiology*, 1–13. <https://doi.org/10.1080/14992027.2021.1948120>
- Vroegop J. L., Homans N. C., Goedegebure A., Dingemans J. G., van Immerzeel T., & van der Schroeff M. P. (2018). The effect of binaural beamforming technology on speech intelligibility in bimodal cochlear implant recipients. *Audiology and Neurotology*, *23*(1), 32–38. <https://doi.org/10.1159/000487749>
- Warren S. E., Noelle Dunbar M., Bosworth C., & Agrawal S. (2020). Evaluation of a novel bimodal fitting formula in advanced bionics cochlear implant recipients. *Cochlear Implants International*, *21*(6), 323–337. <https://doi.org/10.1080/14670100.2020.1787622>
- Weissgerber T., Rader T., & Baumann U. (2017). Effectiveness of directional microphones in bilateral/bimodal cochlear implant users - impact of spatial and temporal noise characteristics. *Otology & Neurotology*, *38*(10), e551–e557. <https://doi.org/10.1097/MAO.0000000000001524>
- Williges B., Wesarg T., Jung L., Geven L. I., Radeloff A., & Jürgens T. (2019). Spatial speech-in-noise performance in bimodal and single-sided deaf cochlear implant users. *Trends in Hearing*, *23*, 1–16. <https://doi.org/10.1177/2331216519858311>
- World Medical Association. (2013). World medical association declaration of Helsinki: ethical principles for medical research involving human subjects. *JAMA*, *310*(20), 2191–2194. <https://doi.org/10.1001/jama.2013.281053>
- Zeng F.-G., Nie K., Stickney G. S., Kong Y.-Y., Vongphoe M., Bhargava A., & Cao K. (2005). Speech recognition with amplitude and frequency modulations. *Proceedings of the National Academy of Sciences of the United States of America*, *102*(7), 2293 – 22298. <https://doi.org/10.1073/pnas.0406460102>