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MEASURING THE PERFORMANCE OF THE HEARING AIDS ADAPTIVE DIRECTIVITY AND NOISE REDUCTION ALGORITHMS THROUGH SNR VALUES

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ABSTRACT

Audibility in noise is hearing impaired people essential problem. Hearing aids manufacturers, in recent years, have improved signal processing options such as adaptive microphone directivity and noise reduction. The most recent hearing aids use of the source localization to distinguish the useful signals (speech) from those caused by the nuisance (noise). Once localized, the noises are attenuated to let the speech out. The objective of this study is to investigate this speech intelligibility options effectiveness using the Signal to Noise Ratio (SNR) for different speech signal directions. The auditory simulation platform of the Faculty of Pharmacy audiology training center of the University of Lorraine has been used to reproduce the free-field vocal audiometry conditions and to analyze the output signals of the hearing aids placed on the ears of an artificial head. The output SNR of hearing aids is estimated using the Hagerman and Olofsson method (2004). The SNR frequency variation for the different speech signal directions allows us to discuss algorithmic strategies. Finally, listening to the analyzed signals makes it possible to compare the measured objective numerical values with our subjective intelligibility and listening comfort feeling.

1. INTRODUCTION

The purpose of hearing aids is to restore the hearing impaired's ability to communicate fully with those around. In a calm environment or with low noise, understanding is easy for a hearing-impaired person properly fitted. On the other hand, in a complex or noisy sound environment, the challenge is much more difficult. The only increase in hearing aid gain is no longer sufficient to maintain speech comprehension because it acts without distinction on the whole sound environment. The signal-to-noise ratio (SNR) index makes it possible to assess the performance of the hearing aids in its ability to highlight the useful signal, ie. speech at the expense of noise, a signal by definition useless. Many hearing aid manufacturers promise a satisfactory understanding in noise, but this requires the patient to be placed judiciously in front of his interlocutor in order to facilitate the hearing aids work in capturing the

voice signal. To navigate successfully in complex sound environments, it is necessary to have access to all sounds, in order to be able to modify the attention [1]. To give meaning to a complex auditory scene, the brain organizes sounds into different auditory "objects" which are either highlighted or in the background. The development of these acoustic objects is obtained by assembling sound elements which have similar characteristics. These functionalities can be spectral, temporal or spatial.

These audiological properties are unfortunately less efficient if they are decoded in the hearing periphery (middle ear, cochlea) of people with hearing impairment. For example, hearing loss can prevent the detection of certain frequencies, or widening cochlear filters can impair the ability to resolve spectro-temporal data. As a result, developing and switching between sound objects becomes slower for the hearing impaired. This slow process leads to difficulties, especially in dynamic sound situations with rapid evolution of the desired voice signal, such as lively family dinners [1].

Some manufacturers, such as Oticon, released a range of hearing aids with new technology that improves speech understanding in noise in complex hearing scenes. They use spatial noise estimation techniques using the microphone doublet directionality combined with multidirectional noise reduction algorithms. These techniques thus allow a rebalancing between speech signals in all directions and selective noise reduction. The purpose of this paper is to assess the performance of this new technology called OpenSound Navigator through a study of the hearing aids output SNR.

2. DESCRIPTION OF OPENSOUND NAVIGATOR TECHNOLOGY

The classic technology often used in current hearing aids uses two independent processes: directivity and noise reduction. The directivity, carried out by the two microphones weighting, is applied to suppress the sources of noise around the speaker. Thereafter, noise reduction is applied to the resulting signal, to further reduce the noise remaining in the signal (see Figure1).

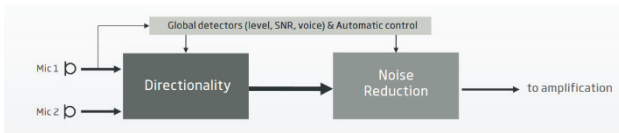


Figure 1: Block diagram of a speech improvement system with directivity, noise reduction, automation and global detectors [2].

More recently, binaural directivity has been introduced. In these systems, the greater distance between the microphones placed on the right and left hearing aids creates a narrower beam in low frequencies which suppresses more noise from directions other than that of the speaker's voice. Noise reduction systems, especially those working in a single frequency band, are less efficient, but this reduction improves listening comfort [3]. Noise reduction systems use the Wiener filter technique and/or the “long-term” properties of the signal to estimate noise. These techniques are slower to respond to rapid changes in complex sound environments.

These technologies effectiveness in everyday sound environments has been criticized. Recent studies have shown that directivity systems cause a decrease of speech comprehension using a radius of less than 50° [4, 5]. The current technology suppresses the information that the brain naturally uses to unravel complex acoustic environments.

Oticon proposes a technology that not only eliminates noise, but also preserves, in all directions, the important information of speech in order to facilitate the natural process to compose sound objects. This technique called MSAT for Multiple Speaker Access Technology reduces noise in complex environments, without isolating the single speaker and retaining access to multiple speakers [6]. For the OPNS hearing aid family, MSAT technology is implemented in OpenSound Navigator™ (OPNS).

The OSN is an algorithm which firstly performs a sound environment “Analyze” phase using the two microphones. A first overview of the auditory scene is taken in omnidirectional mode. The second is done using cardioid mode to pick up noise from the sides and back. This operation is updated 500 times per second and for 16 frequency bands (Figure 2 on left).



Figure 2: Left: The cardioid directivity indicates that the noise is about 6 dB louder at the back than at the side. Right: Attenuation of noise sources between speech sources [2].

In a second time, a procedure called "Balance" improves the SNR between the signals received by the “Analyze”

phase, ie the omnidirectional signal which contains speech + noise and that from the rear cardioid mode which includes noise. An MVDR algorithm (Minimal Variance without Distortion Response), originally widely used in weighting acoustic antennas techniques [7] to improve the reception of a useful signal, is implemented in this procedure. The manufacturer indicates that this procedure makes possible to subtract the noise from the omnidirectional signal to minimize it and thus make a sound balance where speech is perceived more clearly. This “Balance” procedure is performed 125 times for each frequency band. According to studies carried out by Oticon, this process makes it possible to control 16 sound sources [2] (See Figure2 on right).

A final phase of noise reduction, called Noise Removal is implemented after the “Balance” one. It is dedicated to attenuating noise located near the speech source. A module called YouMatic™ LX adapts the Balance and Noise Removal phases according to the environments and the user sound preferences (see Figure 3).

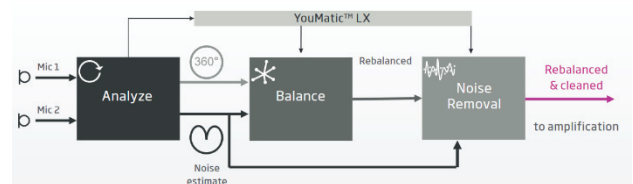


Figure 3: Block diagram of the OpenSound Navigator [2].

To assess this treatment performance, we use the auditory simulation platform of the Faculty of Pharmacy audiology training center of the University of Lorraine in order to estimate the hearing aids output SNR.

3. THE SNR MEASUREMENT PLATFORM DESCRIPTION

3.1 The platform

The hearing aids output SNR measurement platform consists of an artificial human bust located in the center of a sound multicast system. The wall facings are treated with an acoustic absorbent material to approach free field conditions, see Figure 4. The reverberation time after treatment meets the standards imposed for an audiometry room, namely much less than 0.5 s from the 125 Hz octave [8] and a background noise much less than 30 dBA, namely 20.3 dBA (see Table 1).



Figure 4: The SNR measurement platform.

| Octave (Hz) | 125 | 250 | 500 | 1k | 2k | 4k | 8k |
|--------------------|------|------|------|------|------|------|------|
| Tr (s) | 0,33 | 0,18 | 0,13 | 0,14 | 0,12 | 0,12 | 0,11 |
| $L_{eq,30s}$ (Lin) | 29,7 | 20,3 | 12,1 | 8,6 | 9,1 | 10,5 | 11,7 |

Table 1: Background noise level and reverberation time per octave bands.

The KEMAR (Knowles Electronics Manikin for Acoustic Research) mannequin from the G.R.A.S Society allows to simulate the effects of human head and torso in a sound field. Binaural sound capture considers the transfer functions relating to the head, or HRTFs (Head Related Transfer Function). For the tests, his artificial ears can be naked, fitted with hearing aids with custom made earmolds as shown in Figure 5, or even fitted with conventional headphones or earphones.



Figure 5: KEMAR artificial ear fitted with a hearing aid with a custom made earmold.

In each artificial ear hollow, a 1/4" microphone of the 26AS type simulates the eardrum and receives sounds through an IEC 60318-4 type coupler simulating the ear canal. The manikin is located 1.10 m from five active monitoring speakers (Klein & Hummel O110), as shown in the block diagram in Figure 6, to create a homogeneous sound field [9].

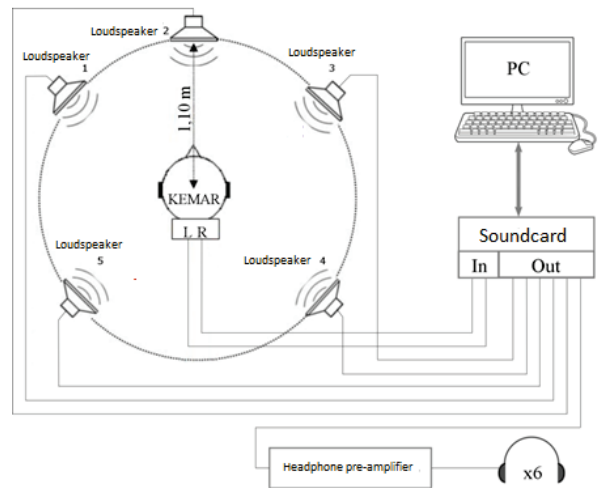


Figure 6: Platform block diagram.

The KEMAR microphones are powered by the GRAS type 12AQ conditioner which pre-amplifies the microphone signals before acquisition. A RME Fireface 802 sound card with 8 analog inputs/outputs is connected via USB 2 to the measurement PC.

3.2 Control interface

A user interface to control the platform was developed using Matlab software (see Figure 7). First, it allows controlling the audio input/output of the sound card with the *pa_wavplayrecord* function made available on the "file exchange" page of mathworks.com by Matt Frear [10]. The test signals are thus diffused and acquired in synchronization, which makes it possible to compare temporal events between two measurements.

It allows the choice of:

- Diffused sound signals (speech, noise, music) and their respective sound levels at the point manikin localization,
- Spatial distributions of the diffused signals with the different monitoring speakers,
- Display, calculation and data recording options from the recorded signals.

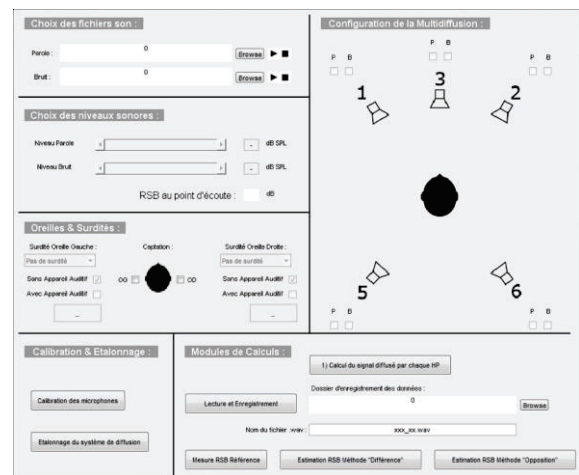


Figure 7: Control Interface

The diffused and recorded signals are sampled at a 44.1 kHz rate and digitized in 16 bits.

3.3 SNR estimation method

Whatever noise the patient faces, the more the SNR decreases, the more the energy masking increases, and the more the intelligibility decreases [11, 12]. This is reflected in the classic intelligibility score curves as a function of SNR, for example that of Figure 8. The word recognition score (WRS) represents the correctly recognized words percentage and SRT (Speech Recognition Threshold) the SNR level corresponding to 50% of intelligibility.

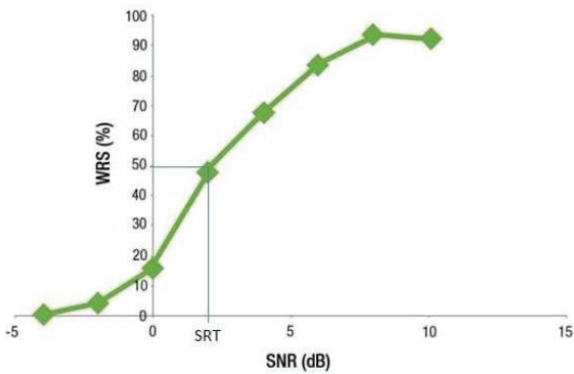


Figure 8: Intelligibility score example according to the SNR, extracted from [13].

The SNR parameter is often used in studies on intelligibility in noise [14] because it is an objective physical quantity. The difficulty lies in estimating the level of the noise $n'(t)$ regardless of the speech level $s'(t)$ at the hearing aids output when the intelligibility enhancing options are activated. As signal processing in devices varies with the sound environment, these levels should be estimated when the hearing aids are immersed in a real mixture of noise and speech.

The method proposed by Hagerman et al. in 2004 [15] makes it possible to estimate the SNR at the hearing aids output when speech and noise are emitted simultaneously, by means of 2 measurements, one of which with emitted noise in phase opposition. This allows, by adding and subtracting the 2 signals measured at the hearing aids output, to "separate" the speech signal $\hat{s}'(t)$ from the noise signal $\hat{n}'(t)$. The estimated signals $\hat{s}'(t)$ and $\hat{n}'(t)$ lead to the estimation of the SNR at the hearing aid output such as:

$$SNR = \widehat{L}_{s'} - \widehat{L}_n,$$

with $\widehat{L}_{s'}$ et \widehat{L}_n the sound pressure levels of separated speech $\hat{s}'(t)$ and separated noise $\hat{n}'(t)$.

The processing implemented in hearing aids can be non-linear and induce harmonic and temporal distortion on the output signal. This induces "spurious" noise overlapped on the speech and noise signals generated at the input, which can bias the SNR estimation. There is a computational method using the Hilbert transform to assess the distortion

level at the hearing aids output: the Distortion/Signal Ratio (DSR). This indicator makes it possible to decide whether the output SNR estimate is correct or not according to its value. The calculated SNR is reliable when the DSR is less than -20 dB over the entire spectrum [16, 17].

For this study, we want to separate several speech signals (simulating several interlocutors around the manikin) and noise in order to estimate a hearing aid output SNR value for each speech signal contributing to the sound scene. For this purpose, the method proposed by Hagerman et al is extended for 2 speech signals $s_1(t)$ and $s_2(t)$, by means of 3 measurements:

$$a(t) = s_1(t) + s_2(t) + n(t),$$

$$b(t) = s_1(t) + s_2(t) - n(t),$$

$$\text{and } c(t) = s_1(t) - s_2(t) - n(t).$$

Operating:

$$a'(t) - b'(t),$$

$$b'(t) + c'(t),$$

$$\text{and } a'(t) - b'(t)$$

allows to separate the 3 signals at the hearing aids output.

Each separated signal $\widehat{s}_1'(t)$, $\widehat{s}_2'(t)$ and $\widehat{n}'(t)$ is then filtered by critical frequency band thanks to a 3rd order Butterworth digital bandpass filter. It allows to estimate the sound levels and thus SNR values for each critical band.

From these SNR values, a weighted average SNR has been determined. This was calculated by multiplying respectively each SNR values by a coefficient linked to the importance of speech on each critical band. The coefficients are given by the method for calculation of the speech intelligibility index (SII) [18].

4. MEASUREMENT OF THE SNR

4.1 Hearing aid settings

A latest generation pair of hearing aids type OPN S1 from Oticon was used in our experimental protocol. The hearing aid settings are representative of those necessary for the hearing impaired with deafness:

- Bilateral presbycusis type,
- An average hearing level of 43.75 dB HL (Figure 9).

This hearing loss degree is common and hearing aids of the OPN S1 type have been adjusted to compensate for it via NAL-NL2 methodology [19].

OpenSound Navigator (OPNS) technology was activated on program 2 (P2) with maximum management of noise reduction and a very high level of spatial processing. Program 1 (P1) contains an equally high noise reduction

processing as well as an adaptive directivity algorithm. Program 0 (P0) does not contain any signal processing (no noise reduction) and is in omnidirectional mode. These programs are conventionally used in the hearing-impaired patient treatment. By comparing these three programs, we wanted to verify the OpenSound Navigator algorithm performance. These programs were carried out on the Genie 2019.2 software. As we can see in Figure 5, custom made earmolds were designed for the artificial head ears. There are no vents in these earmolds. Thus, they guarantee maximum tightness so that the received signal is exclusively coming from the hearing aid output.

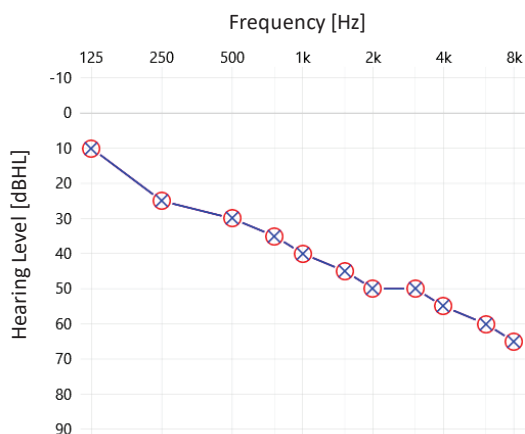


Figure 9: Presbycusis and bilateral hearing losses.

4.2 Description of the studied sound scenes

Our objective being to evaluate the OpenSound Navigator system performances allowing to locate and highlight the speech sources between those emitting noise, we designed three complex environments. The first one (ENV 1) consists of:

- Two speech sources (ISTS signal, International Speech Test Signal) at a respective sound level of 67 dB_{SPL} and placed respectively at 0° (ISTS₁ 0°) and 90° (ISTS₂ 90°) from the artificial head,
- Two noise sources each emitting a vocal noise signal (Cocktail Party) [20]. These noise sources are placed respectively at -60° and 30°. They emit simultaneously a total noise level equal to 65 dB_{SPL}.

The second environment (ENV 2) is identical to the first one, we added a third source of noise at the rear 180° from the head, total noise level also equal to 65 dB_{SPL}.

For the third environment (ENV 3), the two speech sources are placed respectively at 90° (ISTS₁ 90°) and 180° (ISTS₂ 180°) from the artificial head. The noise sources are placed respectively at -60°, 0° and 30° for a total level equal to 65 dB_{SPL}.

For all environments, 2nd speech signal ISTS₂ is delayed in time by 3 seconds compared to ISTS₁, to avoid that they are coherent acoustic sources.

In the three complex sound environments, the input SNR has been set at 2 dB corresponding to uncomfortable noisy conditions (in restaurants for example) to challenge the processing algorithms of the three programs P0, P1 and P2.

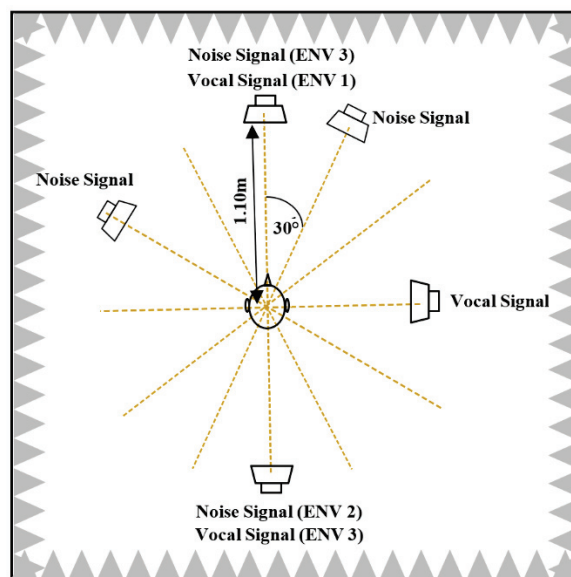


Figure 10: Speaker arrangement around the artificial head.

4.3 Results

For sound environments ENV 1, ENV 2 and ENV 3, the performances of programs P0, P1 and P2 are evaluated by comparing SNR values estimated for both artificial left and right ears, and both ISTS₁ and ISTS₂ speech signals, *ie* SNR₁ and SNR₂.

4.3.1 ENV 1

The first sound environment simulates aided sound perception of 2 speakers, one in the front and the second on the right side, in noisy conditions. Figures 11 to 14 present SNR values as a function of critical frequency band, for left and right ears and ISTS₁ and ISTS₂ signals.

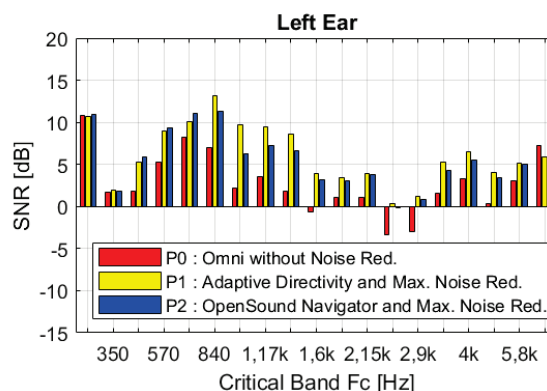


Figure 11: SNR₁ as a function of critical band, estimated between front speech (ISTS₁ 0°) and cocktail party noise, left ear, ENV 1.

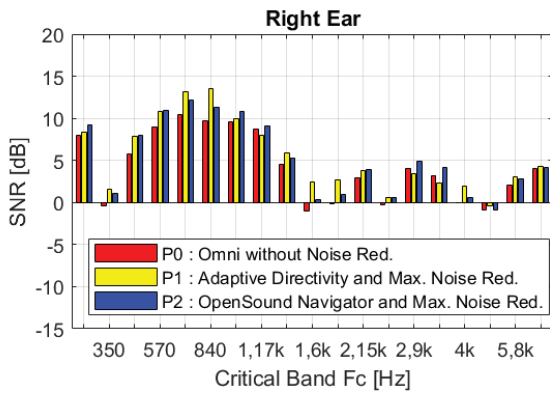


Figure 12: SNR_1 as a function of critical band, estimated between front speech ($ISTS_1 0^\circ$) and cocktail party noise, right ear, ENV 1.

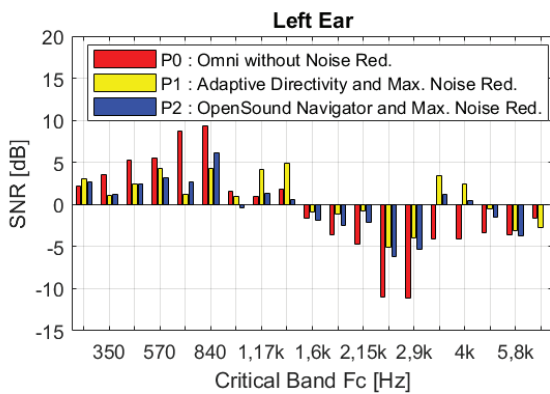


Figure 13: SNR_2 as a function of critical band, estimated between side speech ($ISTS_2 90^\circ$) and cocktail party noise, left ear, ENV 1.

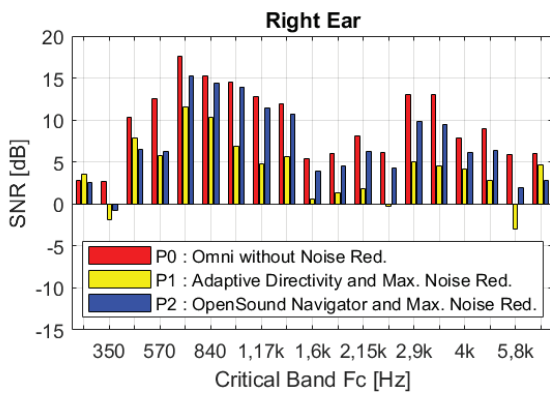


Figure 14: SNR_2 as a function of critical band, estimated between side speech ($ISTS_2 90^\circ$) and cocktail party noise, right ear, ENV 1.

At the first level of analysis, SNR_1 values are quite symmetrical for left and right ears whereas SNR_2 values are clearly better for the right ear because of its direct exposition to $ISTS_2$ speech source. Left ear undergoes the shadow effect of the head and SNR_2 attenuation at medium frequencies is due to the attenuation of $ISTS_2$ signal by the head whereas noise is surrounding.

Appearance of SNR values evolution among frequency is quite comparable for the 3 programs. P1 and P2 do not alter the spectral composition of speech and noise signal, for cocktail party noise.

For the front speech $ISTS_1$, the programs P1 (Adaptive directivity and noise reduction) and P2 (OPNS and noise reduction) give the best SNR_1 values (slight better for P1). Binaural balance is ensured for these two programs unlike P0 which gives a lower output SNR that differs between the two ears. For P1 and P2, there is a better SNR on the left ear than on the right one. On the left side, the $ISTS_1 0^\circ$ speech signal is further from the noise source compared to the other noise source on the right side. The noise source on the left is easily spatially filtered due to the directivity algorithms of the programs P1 and P2. For P0, the SNR is significantly poorer on the left because of its omnidirectional mode which naturally acquires the noise source placed closer to the left ear than the $ISTS_1 0^\circ$ speech signal.

In the case of the 90° speech signal ($ISTS_2 90^\circ$), the P1 program, which is focused forward due to the fixed directivity lobe, has a harder time picking up the speech signal at 90° . It is clearly observed that the algorithms of the OPNS (P2) which can locate the useful sources (speech signal) offer a better SNR_2 compared to P1. However, since the speech signal ($ISTS_2 90^\circ$) remains closest to the right ear, the P0 program in omnidirectional mode provides a good SNR_2 for the right ear.

Weighted mean SNR values presented Figure 15 highlight the binaural added value of programs P1 and P2 for front speech $ISTS_1$.

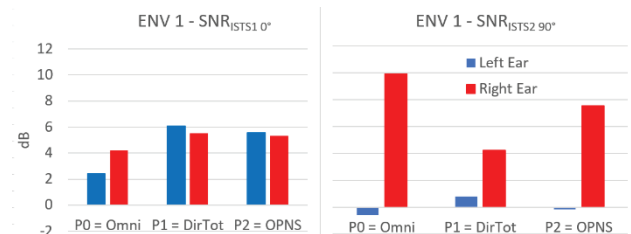


Figure 15: Global SNR estimated for the ENV 1, for P0, P1 and P2 programs respectively, for $ISTS_1 0^\circ$ (on the left) and $ISTS_2 90^\circ$ (on the right).

For ENV1, the P2 program (OPNS) provides the best overall SNR output from hearing aids when both speech signals are emitted simultaneously.

4.3.2 ENV 2

For the ENV 2, we generally observe the same trends except that binaurality is better assured by P1 and P2 for SNR_2 than for ENV 1. For the speech signal located in front of the subject ($ISTS_1 0^\circ$), the P1 program gives a better SNR. The fixed directivity of this program allows better spatial filtering of the rear source. On the whole of this environment, for the two speech signals emitted simultaneously, the program P1 seems to offer a slight better hearing comfort than P2 (OPNS). We would have

expected that the P2 would guarantee better filtering of the rear noise source.

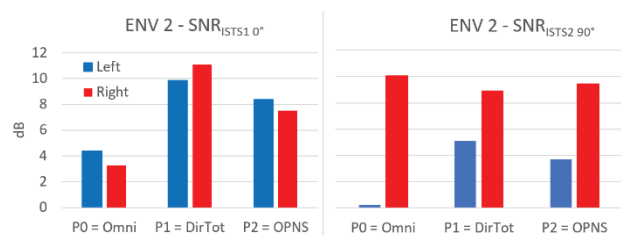


Figure 16: Global SNR estimated for the ENV 2, for P0, P1 and P2 programs respectively, for $ISTS_1 0^\circ$ (on the left) and $ISTS_2 90^\circ$ (on the right).

4.3.3 ENV 3

For ENV 3, we unfortunately no longer see any binaural gain: the SNR on the left is always better in omnidirectional mode (P0) compared to the two other programs P1 and P2 even when the voice source is located at the rear ($ISTS_2 180^\circ$). If we compare the P1 and P2 programs, OPNS is slightly better than Adaptive directivity especially for a 90° speech signal ($ISTS_1 90^\circ$). One can suspect that the localization process of the OPNS is reaching its limits especially for a low input SNR which leaves little speech emergence compared to noise. We can regret that in this case the OPNS does not switch to omnidirectional mode which offers in this very complex environment a better output SNR.

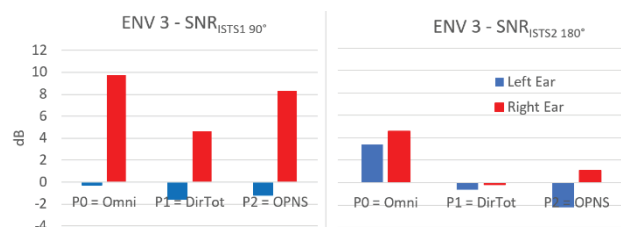


Figure 17: Global SNR estimated for the ENV 3, for P0, P1 and P2 programs respectively, for $ISTS_1 90^\circ$ (on the left) and $ISTS_2 180^\circ$ (on the right).

5. CONCLUSIONS

The localization process of useful speech signals in a noisy environment is an increasingly challenge among hearing aid manufacturers. The binaural hearing aids connectivity has made it possible to improve the source detection algorithms thanks to the continuous analysis of the sound environment and the use of efficient directivity. In the situations of speakers placed at the front and/or on the side in a noisy acoustic context, this study has shown that the treatment processes are quite efficient by giving a hearing aids output SNR that can reach more of 8 dB. An improvement in binaural balance was observed for the OpenSound Navigator process for signals coming from the front with distribution of noise sources also coming

from the front and rear. However, in the case where the speech emergence is low, it is still difficult to restore hearing comfort for the algorithms when two speech sources are emitted simultaneously for large incidence (90° and 180°). We can criticize our protocol for which the input SNR was low, the incidences of speech sources very large, and the fact that these speech sources were emitted simultaneously. In the future output SNRs should be determined for other higher input SNRs and for separately transmitted speech signals.

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