

SPATIAL NOISE REDUCTION IN BINAURAL HEARING AIDS

Navin Chatlani⁽¹⁾, Egghart Fischer⁽²⁾ and John J. Soraghan⁽¹⁾

⁽¹⁾ Centre for Excellence in Signal and Image Processing, University of Strathclyde

Royal College Building, 204 George Str., Glasgow

Email: {navin.chatlani, j.soraghan}@eee.strath.ac.uk

⁽²⁾ Siemens Audiological Engineering Group, Gebbertstr. 125, Erlangen

Email: egghart.fischer@siemens.com

ABSTRACT

Binaural hearing aids are configured to have a wireless transmission link between the left and the right hearing aid. State-of-the-art hearing aids use two microphones in each instrument for compactness and power consumption constraints. In environments with multiple interferers, directional signal processing in hearing aids use techniques such as differential microphone arrays to improve speech intelligibility. However, these hearing aids have maximum sensitivity to target sources located directly in front or directly behind the user. In this paper, a novel binaural system is presented for steering the look direction of the hearing aid to other angles than 0/180 degrees. The new system was tested in a real-time environment to confirm the results.

1. INTRODUCTION

One popular solution to hearing loss is the use of hearing aids. There remain many deficiencies in hearing aid technologies. A common problem encountered in the hearing aid's signal processing is the removal of unwanted disturbances, i.e. noise from desired speech signals. Adaptive signal processing techniques such as active noise cancellation are commonly used when enhancing speech sequences, with [1] or without [2] a noise reference. Alternative approaches exist, however in the case of non-stationary signals such as speech in a complex hearing environment with multiple speakers, directional signal processing is vital to improve speech intelligibility by enhancing the desired signal. Traditional hearing aids utilize simple differential microphone arrays (DMA) [7] to focus on targets in front or behind the user. In many hearing situations, the desired speaker azimuth varies from these predefined directions. Therefore, directional signal processing which allows the beam to be steered to a focus direction would be effective at enhancing the desired source.

Recent approaches for binaural beamforming have been presented in [3], [4] and [5]. In [3], a binaural beamformer was designed using a configuration with two 3-channel hearing aids. The beamformer constraints were set based on the desired look direction to achieve a steerable beam with the use of three microphones in each hearing aid which is impractical and too costly in current hearing aids. The system performance was shown to be dependent on the propagation model used in formulating the steering vector. Binaural multi-channel Wiener filtering (MWF) was used in [4] to obtain a steerable beam by estimating the statistics of the

speech signal in each hearing aid. MWF is computationally expensive and the results presented were achieved using a perfect voice activity detector (VAD) to estimate the noise while assuming the noise to be stationary during speech activity. Another technique for forming one spatial null in a desired direction using DMA has been reported in [6]. However, this was shown to be sensitive to the microphone array geometry and therefore not applicable to a hearing aid setup.

In this paper, new techniques are proposed for a binaural system in the hearing aid which can focus on directions other than zero degrees using differential microphone arrays and filtering techniques. A realistic constraint was that each state of the art hearing aid uses two microphones. Due to the data transmission constraint in a binaural system, only one microphone signal was transmitted from each hearing aid to the other using a wireless data link. In section 2, the binaural hearing aid setup is presented along with the background necessary to understand DMA beamforming. In section 3, a "side-look" beamformer is developed which focuses its beam to either side of the head. The proposed technique decomposes the problem to process the low frequencies (≤ 1 kHz) and the high frequencies (> 1 kHz) independently. For the low frequencies, a binaural array is used and for the high frequencies, the head shadow effect is utilized to develop a system to achieve the side look. In section 4, a steerable binaural beamforming system is presented which can focus its beam to a desired source at a given azimuth for the frequency range approximately up to 750 Hz. The proposed technique involves filtering of the noisy signal using an estimate of the desired source signal and an estimate of the noise signal. Section 5 demonstrates the effectiveness of the systems using directivity plots from actual hearing aid signals in a real time environment. Finally, conclusions are made in section 6.

2. BINAURAL HEARING AID SET-UP

The proposed scheme for the binaural hearing aid is illustrated in Figure 1(a) where the left and the right hearing aids are connected by a bidirectional wireless link. Size constraints impose that each hearing aid has two microphones separated by a distance of approximately 10 mm. Due to practical rate constraints and minimization of power consumption, only one microphone signal is transmitted from one hearing aid to the other. In Figure 1(a), the signals $x_{L,j}[n]$ and $x_{R,j}[n]$ are the j^{th} microphone signals from the left and the right hearing aid respectively where $j=1,2$. In Figure 1(a), $x_{L,1}[n]$ corresponds to the signal being transmitted from the front microphone in the left hearing aid to the right hearing

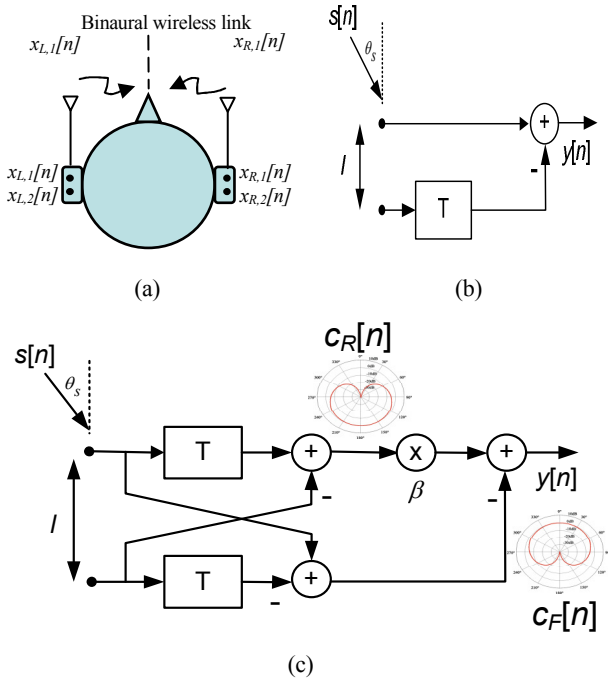


Figure 1 (a) Binaural hearing aid config. with wireless link (b) First order diff. mic. array (c) Adaptive diff. mic. array

aid. Without loss of generality, $x_{R,1}[n]$ corresponds to the right front microphone signal being transmitted to the left hearing aid.

2.1 Differential Microphone Array (DMA)

Traditional hearing aids use a first order DMA [7] with two omni-directional microphones separated by a distance l (approx. 10 mm in an individual hearing aid) to generate a directional response. Its response is independent of frequency as long as the assumption of small spacing to acoustic wavelength, λ , holds. Consider the signal $s[n]$ impinging on the first order DMA at an angle θ_s as illustrated in Figure 1(b). Under farfield conditions, the magnitude of the frequency and angular dependent response of the array is given by [7]:

$$|H(\Omega, \theta)| = \left| 1 - e^{-j\Omega(T + \frac{l}{v} \cos \theta_s)} \right| \quad (1)$$

where v is the speed of sound. The delay T may be adjusted to cancel a signal from a certain direction to obtain the desired directivity response. In hearing aids, this delay T is fixed to match the microphone spacing l/v and the desired directivity response is instead achieved using a back-to-back cardioid system [7] as shown in the adaptive differential microphone array (ADMA) in Figure 1(c). From Figure 1(c), $c_F[n]$ is the cardioid beamformer output that attenuates signals from the back direction and $c_R[n]$ is the anti-cardioid beamformer output which attenuates signals from the front direction. The array output $y[n]$ is given by:

$$y[n] = c_F[n] - \beta c_R[n] \quad (2)$$

For $y[n]$ from (2), the signal from 0° is not attenuated and a single spatial notch is formed in the direction θ_{null} for a value of β given by [7]:

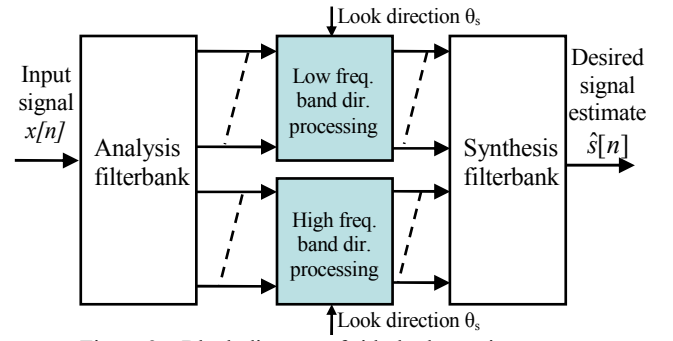


Figure 2 – Block diagram of side-look steering system

$$\theta_{null} \approx \arccos \frac{\beta - 1}{\beta + 1} \quad (3)$$

In ADMA for hearing aids, the parameter β is adapted to steer the notch to direction θ_{null} of a noise source to optimize the directivity index. This is performed by minimizing the MSE of the output signal $y[n]$. Using a gradient descent technique to follow the negative gradient of the MSE cost function, the parameter β is adapted by:

$$\beta[n+1] = \beta[n] - \mu \frac{\delta}{\delta \beta} \varepsilon(y^2[n]) \quad (4)$$

where μ is the update step size as detailed in [7].

3. BINAURAL SIDE LOOK STEERING

In hearing situations, the desired speaker may be on one side of the hearing aid user. Therefore, a system which performs side-look beam steering is realized using binaural hearing aids with a bidirectional audio link. It is known that at high frequencies, the Interaural Level Difference (ILD) between measured signals at both sides of the head is significant due to the head-shadowing effect. The ILD increases with frequency. This head-shadow effect is exploited in the design of the binaural Wiener filter for the high frequencies (>1 kHz). At low frequencies (≤ 1 kHz), the acoustic wavelength λ is long with respect to the head diameter. Therefore, there is minimal change between the sound pressure levels at both sides of the head and the Interaural Time Difference (ITD) is the more significant acoustic cue. At low frequencies, a binaural first-order DMA is designed to create the side-look. Therefore, this side-look steering is decomposed into two smaller problems with a binaural DMA for the low frequencies and a binaural Wiener filter approach for the high frequencies. The proposed system diagram is shown in Figure 2.

The input noisy speech signal $x[n]$ from Figure 2 is given by:

$$x[n] = s[n] + d[n] \quad (5)$$

where $s[n]$ is the desired speech signal from direction $\theta_s \in [90^\circ - 90^\circ]$ and $d[n]$ is the noise signal incident from direction θ_d where $\theta_d = -\theta_s$. The input signal is decomposed into sub-bands by the analysis filterbank [8]. The proposed directional processing is applied and then the signal is reconstructed using a synthesis filterbank [8]. The signal from the side of the interferer is termed the interferer side and the signal on the side of the desired source is termed the focus side.

A bidirectional audio link between the hearing aids is assumed.

3.1 High Frequency Side Look

The head shadowing effect is exploited in the design of a binaural system to perform the side-look at high frequencies ($f > 1$ kHz). The signal from the interferer side is attenuated across the head at these high frequencies and the analysis of the proposed system is given below.

3.1.1 System Model

Consider a target speaker $s[n]$ on the left side (-90°) of the hearing aid user and an interferer $d[n]$ on the right side (90°). From Figure 1(a), the left ear signal model $x_{L,l}[n]$ recorded at the front left microphone and the right ear model $x_{R,l}[n]$ recorded at the front right microphone are given by:

$$x_{L,l}[n] = s[n] + h_{L,l}[n] * d[n] \quad (6)$$

$$x_{R,l}[n] = h_{R,l}[n] * s[n] + d[n] \quad (7)$$

where $h_{L,l}[n]$ is the transfer function from the front right microphone to the left front microphone and $h_{R,l}[n]$ is the transfer function from the front left microphone to the front right microphone. Transformation of equations (6) and (7) into the frequency domain gives:

$$X_{L,l}(\Omega) = S(\Omega) + H_{L,l}(\Omega)D(\Omega) \quad (8)$$

$$X_{R,l}(\Omega) = H_{R,l}(\Omega)S(\Omega) + D(\Omega) \quad (9)$$

Let the short-time spectral power of signal $X_a(\Omega)$ be denoted as $\Phi_{X_a}(\Omega)$. Since the left side is the focus side and the right side is the interferer side, a classical Wiener filter can be derived as:

$$W(\Omega) = \frac{\Phi_{X_{L,l}}(\Omega)}{\Phi_{X_{L,l}}(\Omega) + \Phi_{X_{R,l}}(\Omega)} \quad (10)$$

For analysis, assume that $\Phi_{H_{L,l}}(\Omega) = \Phi_{H_{R,l}}(\Omega) = \alpha(\Omega)$.

$\alpha(\Omega)$ is the frequency dependent attenuation corresponding to the transfer function from one hearing aid to the other across the head. Therefore, (10) can be simplified to:

$$W(\Omega) = \frac{\Phi_S(\Omega) + \alpha(\Omega)\Phi_D(\Omega)}{(1 + \alpha(\Omega))(\Phi_S(\Omega) + \Phi_D(\Omega))} \quad (11)$$

As explained earlier, at high frequencies the ILD attenuation $\alpha(\Omega) \rightarrow 0$ due to the head-shadowing effect and (11) tends to a traditional Wiener filter. At low frequencies, the attenuation $\alpha(\Omega) \rightarrow 1$ and the Wiener filter gain $W(\Omega) \rightarrow 0.5$. The output filtered signal at each side of the head is obtained by applying the gain $W(\Omega)$ to the omni-directional signal at the front microphones on both hearing aid sides. \mathbf{X} is given as the vector $\mathbf{X} = [X_{L,l}(\Omega) \ X_{R,l}(\Omega)]$ and the output \mathbf{Y} from both hearing aids is denoted as $\mathbf{Y} = [Y_{L,l}(\Omega) \ Y_{R,l}(\Omega)]$ and is given by:

$$\mathbf{Y} = W(\Omega)\mathbf{X} \quad (12)$$

Therefore, the spatial impression cues from the focussed and interferer sides are preserved since the gain is applied to the original microphone signals on either side of the head.

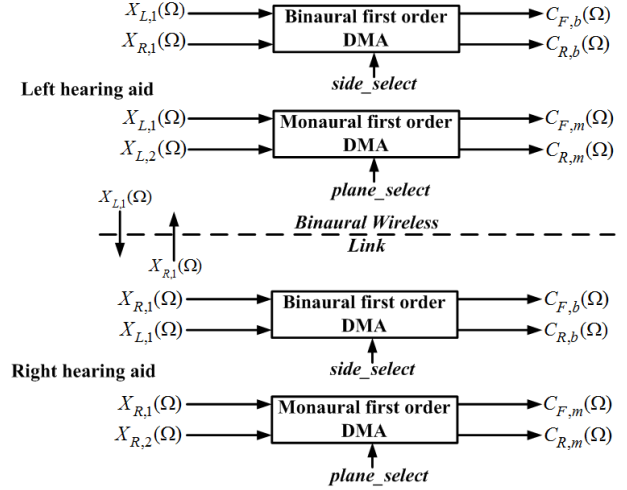


Figure 3 - System diagram for steerable binaural beamformer

3.2 Low Frequency Side Look

At low frequencies, the signal's wavelength is long compared to the distance l_{head} across the head between the two hearing aids. Therefore spatial aliasing effects are not significant. Assuming $l_{head} = 17$ cm, the maximum acoustic frequency to avoid spatial aliasing is approximately 1 kHz. The proposed system for the low frequency side look is designed using the first-order ADMA from Figure 1(c) across the head which is described below.

3.2.1 Binaural First Order ADMA

The left side is the focussed side of the user and the right side is the interferer side. Therefore a system is designed which performs directional signal processing to steer to the side of interest. As described in section 2, consider the left ear signal $x_{L,l}[n]$ and the right ear signal $x_{R,l}[n]$. A binaural first order ADMA is implemented along the microphone sensor axis across the head pointing to $\theta_s = -90^\circ$. Two back-to-back cardioids are thus resolved setting the delay to l_{head}/v . The array output is a scalar combination of a forward facing cardioid $c_F[n]$ (pointing to -90°) and a backward facing cardioid $c_B[n]$ (pointing to 90°) as in (2).

4. STEERABLE BINAURAL BEAMFORMER

The main goal of the steerable system is to achieve specific look directions $\theta_{s,n}$ where:

$$\theta_{s,n} = 45 * n^\circ \quad \forall n = 0, \dots, 7 \quad (13)$$

Beam steering to 0° and 180° is achieved using the basic first order DMA [7]. Section 3 described the proposed system for steering to 90° and 270° . This section details the proposed model for focusing the beam to the subset of angles $\theta_{steer} \subset \theta_{s,n}$ where $\theta_{steer} \in [45^\circ, 135^\circ, 225^\circ, 315^\circ]$. First the proposed parametric model is presented for achieving these desired look directions. This model is used to derive an estimate of the desired signal and an estimate of the interfering signal for enhancing the input noisy signal.

4.1 Parametric Steering Model

The desired signal incident from angle θ_{steer} and the interfering signal are estimated by a linear combination of directional signal outputs. The directional signals used in this estimation are derived as shown in Figure 3. The outputs of the system $C_{F,b}(\Omega)$ and $C_{R,b}(\Omega)$ result from the binaural first order DMA and respectively denote the forward facing and backward facing cardioids. $C_{F,m}(\Omega)$ and $C_{R,m}(\Omega)$ result from the monaural first order DMA and follow the same naming convention as in the binaural case. The parameter “*side_select*” selects which microphone signal from the binaural array is delayed and subtracted and therefore is used to select the direction to which $C_{F,b}(\Omega)$ and $C_{R,b}(\Omega)$ point. When “*side_select*” is set to one, $C_{F,b}(\Omega)$ points to the right at 90° and $C_{R,b}(\Omega)$ points to the left at -90° as indicated in Figure 4(a) and vice versa when “*side_select*” is set to zero as indicated in Figure 4(b). Conversely, the parameter “*plane_select*” selects which microphone signal from the monaural array is delayed and subtracted. Therefore, when “*plane_select*” is set to one, $C_{F,m}(\Omega)$ points to the front plane at 0° and $C_{R,m}(\Omega)$ points to the back plane at 180° as indicated in Figure 4(c) and vice versa when “*plane_select*” is set to zero as indicated in Figure 4(d). For conciseness, the frequency-domain variable Ω will be omitted from now on.

4.2 Signal and Noise Estimation

Consider the desired speaker $s[n]$ to be at azimuth θ_{steer} of 45° . Since the direction of the desired signal θ_{steer} is known, an estimate of the desired signal power can be obtained from measuring the minimum of the power obtained from the directional outputs which mutually have maximum response in the direction of the signal. For this orientation, the parameters “*side_select*” and “*plane_select*” are both set to 1 to give binaural and monaural outputs as indicated in Figure 4(a) and Figure 4(c) respectively. From the frequency domain signals, hypercardioids [9] Y_1 and Y_2 are obtained and signals Y_3 and Y_4 create notches at $90^\circ/-90^\circ$ and $0^\circ/180^\circ$ respectively in:

$$\begin{bmatrix} Y_1 \\ Y_2 \\ Y_3 \\ Y_4 \end{bmatrix} = \begin{bmatrix} C_{F,m} \\ C_{F,b} \\ C_{F,m} \\ C_{F,b} \end{bmatrix} - \beta_{hyp} \begin{bmatrix} C_{R,m} \\ C_{R,b} \\ C_{R,m} / \beta_{hyp} \\ C_{R,b} / \beta_{hyp} \end{bmatrix} \quad (14)$$

where β_{hyp} is set to a value to create the desired hypercardioid. Equation (14) can be rewritten as:

$$\mathbf{Y} = \mathbf{C}_{F,1} - \beta_{hyp} \mathbf{C}_{R,1} \quad (15)$$

where $\mathbf{Y} = [Y_1 \ Y_2 \ Y_3 \ Y_4]^T$, $\mathbf{C}_{F,1} = [C_{F,m} \ C_{F,b} \ C_{F,m} \ C_{F,b}]^T$ and $\mathbf{C}_{R,1} = [C_{R,m} \ C_{R,b} \ C_{R,m} / \beta_{hyp} \ C_{R,b} / \beta_{hyp}]^T$. An estimate of the short time desired signal power $\hat{\Phi}_S$ is obtained from measuring the minimum short time power of the four signal components in \mathbf{Y} as given by:

$$\hat{\Phi}_S = \min(\Phi_Y) \quad (16)$$

The noise estimate is obtained by measuring the maximum power from two directional signals which mutually have a null placed in the direction θ_{steer} of the desired source. For the same parametric values of “*side_select*” and “*plane_select*” as before, let $\mathbf{C}_{R,2} = [C_{R,m} \ C_{R,b}]^T$ and

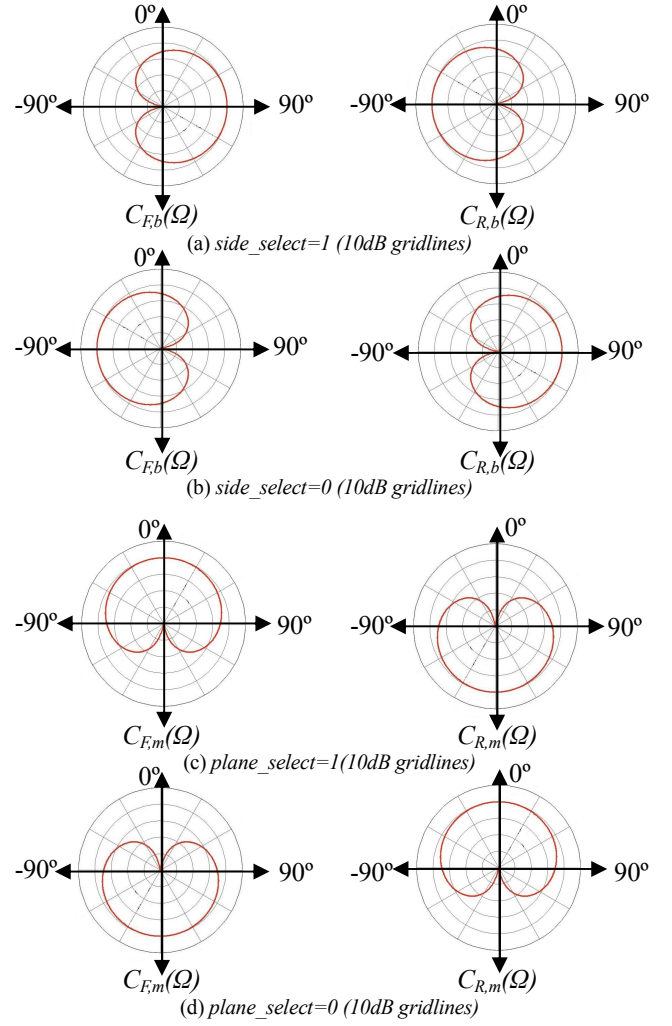


Figure 4 – Differential microphone array outputs for (a)-(b) binaural and (c)-(d) monaural cases

$\mathbf{C}_{F,2} = [C_{F,m} \ C_{F,b}]^T$. These two signals are used to measure the signal \mathbf{V} which is used for the noise power estimation as given by:

$$\mathbf{V} = \mathbf{C}_{R,2} - \beta_{steer} \mathbf{C}_{F,2} \quad (17)$$

where $\mathbf{V} = [V_1 \ V_2]^T$ and β_{steer} is set to place a null at the direction of the desired source. An estimate of the short time noise power $\hat{\Phi}_D$ is obtained from the maximum of the short time power of the two noise components in \mathbf{V} as given in:

$$\hat{\Phi}_D = \max(\Phi_V) \quad (18)$$

The corresponding Wiener filter gain $W(\Omega)$ is obtained from:

$$W(\Omega) = \frac{\hat{\Phi}_S}{\hat{\Phi}_S + \hat{\Phi}_D} \quad (19)$$

The enhanced desired signal is obtained by filtering the locally available omni-directional signal. Steering to the other directions of 135° , 225° or 315° is done by setting the parameter values of [*side_select*, *plane_select*] to [1, 0], [0, 0] or [0, 1] respectively.

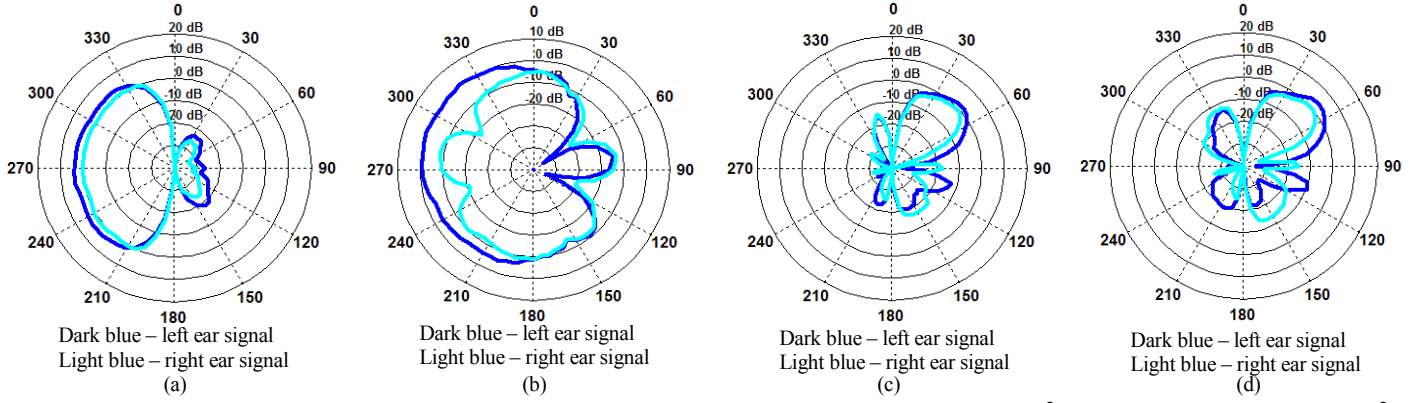


Figure 5 (a) Beam steered to left side at 250 Hz (b) Beam steered to left side at 2 kHz (c) Beam steered to 45° at 250 Hz (d) Beam steered to 45° at 500 Hz

5. PERFORMANCE EVALUATION

The performance of the side-look beamformer and the steerable beamformer was evaluated by examining the output directivity patterns. A binaural hearing aid system was set up as described in section 2 with two “Behind the Ear” (BTE) hearing aids on each ear and only one signal being transmitted from one ear to the other. The measured microphone signals were recorded on a KEMAR dummy head and the beampatterns were obtained by radiating a source signal from different directions at a constant distance.

5.1 Binaural side-look steering beamformer

The binaural side-look steering beamformer was decomposed into two subsystems to independently process the low frequencies (≤ 1 kHz) and the high frequencies (> 1 kHz). In this scenario, the desired source is located on the left side of the hearing aid user at -90° ($\equiv 270^\circ$ on the plots) and the interferer on the right side of the user at 90° . The effectiveness of these two systems is demonstrated with representative directivity plots at 250 Hz (low frequency) in Figure 5(a) and at 2 kHz (high frequency) in Figure 5(b). In both plots, the responses from both ears are shown together to illustrate the desired preservation of the spatial cues. It can be seen that the attenuation is more significant on the interfering signal impinging on the right side of the hearing aid user. Similar frequency responses were obtained across all frequencies for focussing on desired signals located either at the left (270°) or the right (90°) of the hearing aid user.

5.2 Steerable beamformer: Example 45° steering

The performance of the steerable beamformer is demonstrated for the scenario described in section 4.2 where the desired speaker $s[n]$ is at azimuth θ_{steer} of 45° . From (16) and (18), estimates of the signal power $\hat{\Phi}_S$ and the noise power $\hat{\Phi}_D$ were obtained. A null is placed at 45° for the noise estimate from (18). The corresponding value of β_{steer} can be calculated as in (3). The polar plot of the beampattern of the proposed steering system to 45° is shown from the left and right hearing aids at 250 Hz and 500 Hz in Figure 5(c) and Figure 5(d) respectively. As required, the maximum gain is in the direction of θ_{steer} . These simulations were performed using actual recorded signals. The steering of the beam can be

adjusted to the direction θ_{steer} by fine-tuning the ideal value of β_{steer} from (3) for real implementations.

6. CONCLUSION

The systems presented in this paper were shown to be effective at steering the look direction of binaural hearing aids to directions other than $0^\circ/180^\circ$. These results were achieved using real hearing aid recordings under the constraint of two microphones in each hearing aid and only one signal being wirelessly transmitted from one side to the other. The technique presented for the steerable beamformer operated up to approximately 750 Hz since the estimates were obtained from a combination of monaural and binaural array outputs (limited due to large spacing between hearing aids across the head). Future work involves extending this system to operate for the high frequencies. Both systems were tested in a real-time environment using hearing aids with multiple interferers from different directions to confirm the results.

REFERENCES

- [1] N. Chatlani, J. J. Soraghan, "Speech Enhancement using Adaptive Empirical Mode Decomposition," in *16th Int'l Conference on Digital Sig. Proc (DSP 2009)*, July 2009.
- [2] N. Chatlani, J. J. Soraghan, "EMD-based Noise Estimation and Tracking (ENET) with application to speech enhancement," in *17th European Signal Proc. Conference (EUSIPCO)*, Aug 2009.
- [3] T. Rohdenburg, V. Hohmann, B. Kollmeier, "Robustness Analysis of Binaural Hearing Aid Beamformer Algorithms by Means of Objective Perceptual Quality Measures," in *2007 IEEE Workshop on Applications of Signal Processing to Audio and Acoustics*, pp.315-318, Oct 2007.
- [4] S. Doclo, M. Moonen, T. Van den Bogaert, J. Wouters, "Reduced-Bandwidth and Distributed MWF-Based Noise Reduction Algorithms for Binaural Hearing Aids," *IEEE Transactions on Audio, Speech, and Language Processing*, vol.17, no.1, pp.38-51, Jan 2009.
- [5] S. Srinivasan, A.C. den Brinker, "Analyzing Rate-Constrained beamforming schemes in wireless binaural hearing aids," in *17th European Signal Proc. Conference (EUSIPCO)*, Aug 2009.
- [6] M. Ihle, "Differential Microphone Arrays for Spectral Subtraction," in *Int'l Workshop on Acoustic Echo and Noise Control (IWAENC 2003)*, Sep 2003.
- [7] H. Teutsch, G. Elko, "First and Second Order Differential Microphone Arrays," in *7th International Workshop on Acoustic Echo and Noise Control (IWAENC 2001)*, pp. 35-38, Sep 2001.
- [8] R. Bauml, W. Soergel, "Uniform polyphase filter banks for use in hearing aids: Design and Constraints," in *16th European Signal Proc. Conference (EUSIPCO)*, Aug 2008.
- [9] S. L. Gay, J. Benesty, *Acoustic Signal Processing For Telecommunication*, Springer, 1st Edition, 2000.