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# Microphone arrays for hearing aids: An overview

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**Abstract**

Using the difference in the spatial domain (direction or location) between a target signal and noise, a system with a microphone array can achieve the goal of noise reduction and speech enhancement in various environments, especially, in speech-like noisy environments. This paper deals with various issues related to the use of microphone arrays in hearing aids. It includes overall principles, algorithms, real-time implementation, configuration, processing mode, geometry, combination with binaural processing, directivity pattern, frequency responses, and compensation for mismatch and misplacement of microphones. A practical microphone array device will be presented that uses microphone array based processing techniques to help hearing aids deliver improvement in signal-to-noise ratio, reduction of the effects of reverberation, and reduction of the feedback, with appropriate configuration and connections.


## 1. Introduction

The use of microphone array signal processing techniques in digital hearing aids is becoming increasingly widespread. Motivations behind this use mainly come from the following aspects.

First, the benefits of amplification alone in hearing aids are limited. In a noisy place, hearing aids will amplify the noise as well as the desired speech signal. Second, in a reverberant place, hearing aids will amplify late multipath arrivals as well as the direct first-arrival signal. Furthermore, feedback associated with high output hearing aids distorts the frequency response of the hearing aid, which was carefully tuned to compensate for the individual's hearing loss and sometimes causes oscillation. To overcome these limitations, a hearing aid should not only amplify the input signals but also improve the signal-to-noise ratio, reduce the effect of reverberation, and cancel feedback.

As a matter of fact, many schemes exist to enhance the desired speech and reduce interference to improve the signal-to-noise-ratio of hearing-aid outputs (Kates, 1997). These available schemes mainly employ various differences between the desired speech and the noise in the frequency domain and in the time domain. For example, modulation index (defined as the dB difference of the maximal envelope and the minimum envelope of a signal) based schemes employ the property that noises have a lower modulation index than the desired speech, and assign a high amplification gain for the band where the signal is more like speech (high modulation index) and assign a low amplification gain for the band where the signal is more like noise (low modulation index) (Edwards et al., 1998). Another example is spectral-subtraction based schemes which include the following steps: (1) detecting pauses in speech; (2)

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estimating the noise spectrum during the pauses; (3) subtracting the estimated noise spectrum from the entire spectrum (in power spectrum domain) to get an estimation of the desired speech; and (4) amplifying this estimated speech signal with appropriate gains. With these steps, an improvement of signal-to-noise ratio can be achieved (Boll, 1979). Obviously, these schemes will deliver a degraded performance in speech-like noisy environments such as in restaurants because in these situations there is no distinct difference in the frequency domain and in the time domain between the noise and the desired speech. More importantly, there would be unaccepted artifacts in the system output if the calculation of modulation index or detection of speech pause are not accurate when using these schemes (Virag, 1999).

Microphone array based processing techniques which mainly employ the difference in spatial domain (in location or direction) between the target speech signal and the noise can overcome some of the problems mentioned above and have the capability of further enhancing speech understanding for hearing-impaired patients (Lehr and Widrow, 1998). Microphone arrays are called beamforming systems or directional systems and are illustrated in Fig. 1. Let us denote the received signals of the $i$th microphone in the time domain and the frequency domain as $x_i(n)$ and $X_i(f)$ (for $i = 1, 2, \ldots, N$), respectively. The $i$th microphone is followed by a filter with frequency response $W_i(f)$ and the impulse response $h_i(n)$, which gives an output signal $Y_i(f)$ in the frequency domain and $y_i(n)$ in the time domain for this channel, that is,

$$Y_i(f) = W_i(f)X_i(f) \tag{1}$$

or

$$y_i(n) = h_i(n) * x_i(n), \quad i = 1, 2, \ldots, N, \tag{2}$$

where "$*$" represents convolution. The outputs of these filters are summed and provide an input to the other processing parts of hearing aids such as compression amplification or other noise reduction and speech enhancement processing. Let us denote this input $z(n)$ in the time domain and $Z(f)$ in the frequency domain, respectively, then we have

$$Z(f) = \sum_{i=1}^{N} Y_i(f) = \sum_{i=1}^{N} X_i(f)W_i(f), \tag{3}$$

$$z(n) = \sum_{i=1}^{N} y_i(n) = \sum_{i=1}^{N} x_i(n) * h_i(n). \tag{4}$$

The key problem in this microphone array based processing system is how to improve the

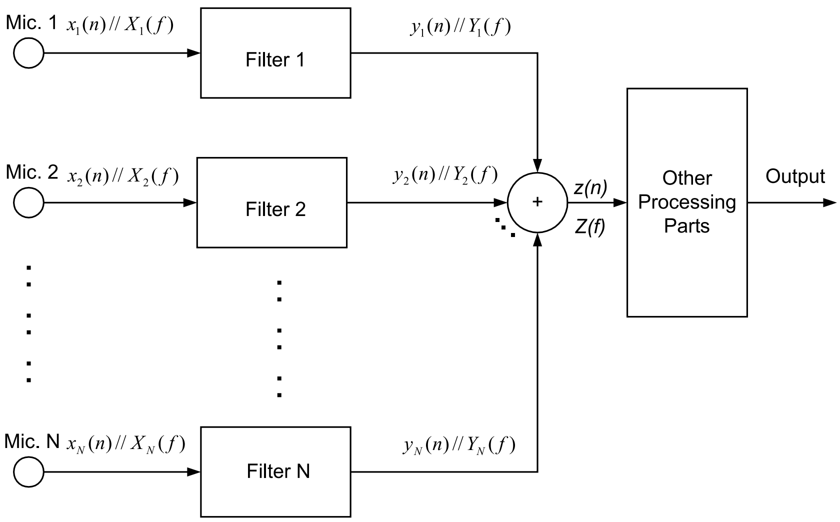


Fig. 1. Schematic diagram of an $N$-microphone array for hearing aids.

signal-to-noise ratio for $z(n)$ so as to help hearing-aid wearers hear better in various noisy environments. There are many factors to affect the performance of this system and its use in hearing aids. These factors mainly include the distance, number, geometry shape and configuration of microphones, the processing mode (fixed processing mode or adaptive processing mode), mismatch of microphones, frequency response of filters, connections and integration with other processing parts of hearing aids, etc. In the following, we will deal with these issues in more detail.

## 2. Practical issues of microphone array based hearing aids

### 2.1. Processing mode

There are usually two processing modes for microphone array systems, that is, the fixed processing mode and the adaptive processing mode (Soede et al., 1993). In the fixed processing mode, the frequency response of each filter in the beamforming system of Fig. 1 remains unchanged when the system operates. In this processing mode, one first specifies the kind of directivity pattern that is required and then gets the coefficients of each filter in an off-line mode by solving a non-linear optimization problem. In this design, the directivity pattern (beam) is a diagram which is used to describe the average power of the output signal $z(n)$ versus the direction of sound arrival. Furthermore, the directivity pattern is also a function of the frequency of the sound. From a noise reduction point of view, a system with a narrow beam around the direction of arrival of the target signal and with a flat frequency response in the frequency range of interest (say, 500–8000 Hz for hearing aids) is desired. The simplest fixed beamforming system is a delay-sum system where each filter in Fig. 1 is in effect a fixed delay unit. However, this simple system has poor directivity and poor frequency response. For time-varying and moving noise environments, any fixed directional system will deliver a degraded performance and the system with an adaptive processing mode will be highly desirable. The coefficients or the frequency response of each filter in an adaptive directional system will be adaptively updated under certain optimization criterion so as to track varying or moving noise sources and to always deliver an optimized performance. As an example, we have the constraint optimization problem such as

$$\min_{H(n)} E(|z(n)|^2) = \min_{H(n)} E\left(\left|\sum_{i=1}^{N} x_i(n) * h_i(n)\right|\right)^2,$$
$$\text{s.t. } H(n) * X(n) = c(n), \tag{5}$$

where $E(\cdot)$ stands for the statistical average; $H(n)$ stands for the set of impulse responses of all the filters, that is $H(n) = \{h_1(n), h_2(n), \ldots, h_N(n)\}$; and $X(n)$ is a vector consisting of $x_i(n)$ $(i = 1, 2, \ldots, N)$ when the sound comes from a specific direction; $c(n)$ is a constant sequence and can be considered as unity for simplicity. If the target signal comes from the specified direction (for example, from the zero direction, that is, straight in front of the listener), the system obtained by solving Eq. (5) can provide the maximum signal-to-noise ratio. However, it would be very difficult in real time to solve the optimization problem of Eq. (5), especially in the case of a large number of microphones. The simplest case is a system with only two microphones in broadside configuration. For this case, we can use the concept developed in (Widrow and Stearns, 1985; Griffiths and Jim, 1982) by considering the summation of two microphones as the primary signal and the difference of two microphones as the reference signal, respectively. All adaptive algorithms given in (Widrow and Stearns, 1985) can be directly used in this system. For the case with more than two microphones, more efficient and simpler adaptive algorithms are still highly desirable.

### 2.2. Configurations of microphone array

The performance of a microphone array system depends not only on the processing mode but also on other various configuration issues such as geometry, number, distances and placement of microphones. Usually, the more microphones there are in the array, the better the performance of the system but more complicated it is. The most common case is a linear uniform microphone array

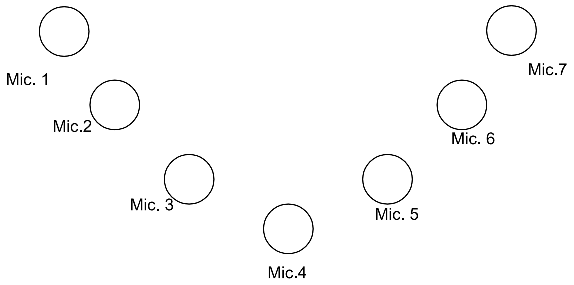


Fig. 2. A planar V-shaped seven-microphone array.

in end-fire configuration or broadside configuration. For this case, the distance between each pair of nearby microphones should be less than half wavelength so as to avoid the spatial ambiguity problem. This problem can also be avoided by arranging the array in a non-uniform way without reducing the entire length of the array. A more complicated case is the microphone array arranged in a planar V-shape as shown in Fig. 2, where seven microphones are used. One advantage of this configuration is that good directivity in both azimuth and elevation can be provided. Concerning the V-shaped array, we will discuss more details in the next section.

The microphone array geometry can be arranged in many other ways, which will mainly depend on the integration and connection to other processing parts of hearing aids. This will be dealt with in the next subsection.

### *2.3. Integration and connection of the microphone array with other parts of hearing aids*

As shown in Fig. 1, the microphone array processing is only a part of the entire hearing-aid process which may include multiband compression amplification, spectral contrast enhancement, single-channel noise reduction and feedback cancellation processing. One important problem in the application of microphone array processing to hearing aids is how to transmit the output signal $z(n)$ of the array to the other parts of the hearing-aid system. The microphone array may be placed on eye glasses or arranged as a necklace or placed around the head of the user, and because the other parts of hearing aids are usually located behind the ear (BTE) or in the ear (ITE), a form of connection is required. There are several ways, with wire or wireless, to realize this connection. As an example, let us consider a microphone array worn on the chest as part of a necklace. A processed signal from the array drives current through a conducting neck loop thus creating a time-variable magnetic field that is representative of the received sound. The magnetic field provides a wireless means for carrying the sound signal to conventional hearing-aid devices located in the ears of the wearer. In order to receive the signal, the hearing aid must be equipped with a "telecoil", a small induction coil contained within the hearing aid whose output can be switch selected by the wearer to serve in place of the hearing aid's microphone signal. When switching the hearing aid to telecoil position, the wearer hears the sound received by the microphone array. When switching the hearing aid to the microphone position, the wearer hears the usual sound received by the hearing aid's own microphone. As a matter of fact, the original purpose of the telecoil was to enable the hearing-aid wearer to converse over the telephone. A hearing-aid compatible telephone receiver radiates a time-varying magnetic field corresponding to the telephone signal. This is generally leakage flux from the receiver. Using the telecoil, many patients can hear over the telephone much more effectively. We are able to take advantage of the telecoil, which is commonly available in the most powerful behind-the-ear hearing-aid types, to provide a wireless link between the chest-mounted array and the hearing aid. Telecoils can be fitted to almost all hearing aids.

In Fig. 1, beamforming processing is separated from other signal processing needed in hearing aids. In other words, the design of each filter in Fig. 1 is independent of compensation processing for hearing loss. As an alternative, beamforming processing and compensation processing can be combined in some way so as to achieve better performance. For example, each filter in Fig. 1 can first be decomposed into multifrequency bands and then designed in combination with the multiband compression amplification used in most current digital hearing-aid products so as to obtain a better directivity pattern, a better frequency response and to get a better compression gain in various noisy environments. However, this system will become more complicated and many trade-offs will be needed in practical applications.

### 2.4. Combination of microphone array processing with binaural processing

The combination of beamforming with binaural processing is receiving increasing attention because this combination can provide both the spatial-filtering benefits of the microphone array and the natural benefits of binaural listening to sound localization ability and speech intelligibility. Usually, the interaural time difference (ITD) and interaural level difference (ILD) are considered as two important binaural cues. However, how to effectively combine these two kinds of processing methods is very challenging because it is difficult to preserve binaural cues in designing spatial domain processing system while aiming to target speech enhancement.

The combination of these two kinds of processing can be accomplished in either fixed mode or adaptive mode. As a result, we can have four types of multimicrophone based systems which could be used in hearing aids, that is, (1) fixed beamforming mode, (2) adaptive beamforming mode, (3) fixed binaural beamforming mode and (4) adaptive binaural beamforming mode. The first two modes are beamforming-alone systems whose outputs have noise reduction performance but do not preserve binaural cues. The last two modes are the combination of beamforming with binaural processing and their outputs not only have noise reduction performance but also preserve binaural cues.

Among these systems, adaptive binaural beamforming systems are the most comprehensive and the most difficult. For these systems, additional constraints should be added in Eq. (5), that is,

$$\begin{aligned} &\min_{H(n)} E(\|z_{\mathrm{R}}(n)|^2 + |z_{\mathrm{L}}(n)|^2), \\ &\quad \text{s.t. } H(n) * X(n) = c(n), \\ &\quad \text{s.t. } E(f) = 0, \\ &\quad \text{s.t. } L(f) = 0, \end{aligned} \tag{6}$$

where $E(f)$ and $L(f)$ are the ITD difference and ILD difference, respectively, obtained before processing and after processing, $z_{\mathrm{R}}(n)$ and $z_{\mathrm{L}}(n)$ correspond to the output of the system for the right ear and the left ear, respectively. Eq. (6) represents a non-linear constrained optimization problem, and it would be very difficult to find the optimum solution of Eq. (6) in real time. However, sub-optimal algorithms have been proposed for adaptive binaural beamforming systems. A good representative is the system developed in (Welker et al., 1997). In this scheme, the combination of adaptive array processing with binaural listening is accomplished by dividing the frequency spectrum, devoting the low-pass part to binaural processing and the high-pass part to adaptive array processing. Another effective adaptive binaural beamforming system is proposed in (Luo et al., 2000). In this system two adaptive spatial processing filters are employed. These two adaptive spatial processing filters have the same reference signal which comes from both ear microphones, but they have different primary signals which correspond to the right ear microphone signal and the left ear microphone signal, respectively. Also, these two adaptive spatial filters have the same structure and the same adaptive algorithm, which reduces the complexity in the hardware implementation. With this two-adaptive-filters based system, some shortcomings of the system in (Welker et al., 1997) could be overcome. Regardless of these efforts, a good compromise between computational complexity and performance has not been achieved still, mainly because these adaptive binaural beamforming systems include only two microphones. To achieve better performance, the development of more effective algorithms with more than two microphones would be highly desirable.

### 2.5. Non-ideal factors in microphone array processing

There are many non-ideal factors that arise when implementing a beamforming system, and these factors will affect greatly the performance of the system. For example, in most available algorithms for designing the filters of Fig. 1, the magnitude response and phase response of all microphones are assumed to be identical. However, in practical applications, there is a significant mismatch in phase and magnitude among these microphones. The mismatch in magnitude and phase will result in degraded performance of the beamforming system. For the adaptive beamforming systems with two microphones in broadside configuration mentioned in Section 2.1, the mismatch means that there is some target speech signal in the reference signal and the assumption that the reference signal contains only noise is no longer correct and hence the system will reduce not only the noise but also the target signal. As a result, some preprocessing (called matching filters) to compensate for the mismatch needs to be added in each channel of the system in Fig. 1. Matching filters can be determined in either fixed processing mode or adaptive processing mode. In effect, with careful design, a first-order IIR matching filter can compensate for the mismatch in magnitude response very well. However, concerning the phase mismatch, the problem is more serious. First, it would be difficult to measure phase mismatch for each device in application situations. Second, even if the phase mismatch measurement is available, the corresponding matching filter would be complicated, that is, a simple (with first or second order) filter cannot effectively compensate for the phase mismatch. In addition, the matching filter for compensating for magnitude mismatch will introduce its own phase delay; this means that both phase mismatch and magnitude mismatch have to be taken into account simultaneously in designing the desired matching filter. One alternative practical way to overcome this mismatch is to make good calibration among these microphones. This approach would apply also to the compensation for the misplacement of microphones from their specific positions.

With all of the above issues being considered, a planner V-shaped microphone array device with six microphones has been implemented. In the next two sections, we will deal with the above issues with this real array device as an example.

## 3. A real microphone array device for hearing aids

The array design and geometry of the device is shown in Fig. 3. The device is constructed in the form of a necklace including an array of microphones mounted on a housing supported on the chest of a user by a conducting loop encircling the user's neck. The device is comprised of an array of six microphones, four pushbuttons for control and a plastic case designed to fit both the adult male and female torso. The plastic case was designed by computer, completely specified in software. It contains batteries and all of the signal processing electronics. Two custom ASIC chips were designed for this device, one for signal processing and the other to serve as an interface between a PC computer and the signal processing chip when this chip is being programmed. Custom chips were needed because of the tight space requirements and the requirements for low battery drain. The connection between this array device and the hearing aids is accomplished by a "telecoil" equipped in hearing aids and the magnetic field produced by the device as mentioned in Section 2.3. This connection can also reduce feedback because the microphones in this device are located at larger distances from the loudspeakers of the hearing aids than the microphones on the hearing aids themselves.

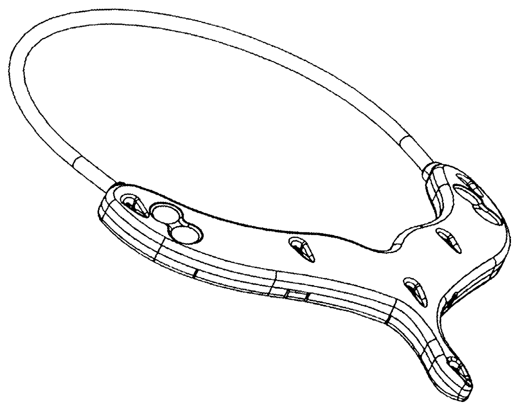

Fig. 3. Actual geometry of the directional hearing array of Hearingpoint Systems, Inc.

As suggested in Section 2.3, the audio spectrum from 200 Hz to 6 kHz is divided into 12 bands in this device, each with its own digital gain control. The six microphone signals are amplified and weighted and then fed to each of the 12 bandpass filters. Different microphone-signal weightings were designed for each frequency band so that the beam width was able to be held at approximately 60° over the entire frequency range of interest. The microphone weights were designed off-line by solving a non-linear optimization problem to achieve the desired beam shape and to achieve a specified robustness to inherent variations in microphone characteristics. A least square error criterion was used for the design. By band-pass filtering the weighted microphone signals with a set of filters covering the audio frequency range and summing the filtered signals, a receiving microphone array with a small aperture size is caused to have a directivity pattern that is essentially uniform over frequency in three dimensions. This method enables the design of highly directive hearing instruments which are comfortable, inconspicuous, and convenient to use. The array provides the user with a dramatic improvement in speech perception over existing hearing-aid designs, particularly in the presence of diffuse noise, reverberation, and feedback.

## 4. Testing of the real microphone array device

Anechoic chamber testing was used to verify the design. Theoretical and measured beam patterns turned out to be remarkably close. More importantly, subject testing was performed to evaluate the effectiveness of the microphone array and to compare listening with the hearing aid alone with listening to the array and hearing aid in telecoil mode. The patient was seated before a loudspeaker that carried the sound of a male test voice. Four loudspeakers on the floor in the four corners of the room carried spectrally weighted band-pass noise. Four additional loudspeakers in the four corners at the ceiling were also used to carry the same noise. The room was not anechoic but had some sound damping. The noise carried by the eight corner loudspeakers produced a noise field that was approximately isotropic.

The test voice and the test noise were stored in a PC computer. The voice and noise data were obtained from Dr. Sig Soli of the House Ear Institute in Los Angeles. We performed a modified version of his HINT test (hearing in noise test) on nine subjects. The HINT was utilized to assess speech understanding of sentences at above threshold level under varying noise conditions.

With the patient seated at a prescribed location marked on the floor, the volume control of the hearing aid and the volume control of the array were set so that the measured volume delivered to the patient's ear would be the same when listening to the test voice through the hearing aid and through the array. The volume level of the test voice was set to be comfortable for the patient in the absence of noise.

Word phrases were spoken to the patient by the test voice with some noise applied. The patient was asked to repeat the words. If any word in the phrase was repeated incorrectly, the response was considered to be incorrect. The noise level was reduced by 2 dB, and another randomly chosen phrase was read. If the response was incorrect again, the noise was lowered by another 2 dB and so forth. When a correct response was obtained, the noise level for the next phase was raised by 2 dB. If another correct response was obtained, the noise level was raised by another 2 dB and so forth. The noise level went up and down, and the average noise level was observed over 10 or 20 phrases.

The average noise level when using the hearing aid was compared to that when using the array. The improvement in signal-to-noise ratio when using the array was significant for these nine test patients. This improvement averages more than 10 dB, which is consistent with anechoic chamber measurements and theoretical calculations.

Other testing was done with the noise volume fixed and the volume level of the test voice fixed. Individual words randomly selected were presented by the test voice. The responses of the patients were observed when using the hearing aid and when using the array. Most patients have only around 25% correct response with the hearing aid and a 75% correct response with the array. These improvements are rather dramatic.

From all testing, we can see that this array device enhances the patient's hearing in the following three ways:

(1) The array enhances signal-to-noise ratio. The patient aims his or her body toward the

person to be listened to. The array beam is 60° wide in both azimuth and elevation. The sound in the beam is enhanced relative to the surrounding sound. The speech of interest is enhanced relative to omnidirectional background noise by about 10 dB, from about 200 Hz to 6 kHz. The gains of the array sidelobes vary between 20 and 35 dB below the gain at the center of the main beam.

(2) The array reduces the effects of reverberation. The array is generally steered toward the sound of interest. The direct primary path is thus aligned with the beam. The secondary paths for the most part arrive at angles outside the beam and are thus attenuated by the array. Reducing reverberation enhances sound clarity since the ear and the brain are somewhat confused by multiple arrivals. This is commonly the case with hearing-impaired individuals.

(3) Use of the array reduces feedback by about 15 dB, since the chest is at a much greater distance from the hearing-aid loudspeaker than is the microphone on the hearing aid itself. Reduction of feedback makes available louder sound for the patient, without oscillation, and allows the hearing aid to function with a frequency response closer to the desired compensation curve. It should be noted that microphone array processing itself does not contribute anything to the reduction of the feedback. The main contribution to the reduction of the feedback comes from the geometry configuration and the placement of the microphone array.

More information about the real microphone array device discussed above and shown in Fig. 3 can be obtained at the website: http://www.hearingpoint.com.

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