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Signal Processing in High-End Hearing Aids: State of the Art, Challenges, and Future Trends

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The development of hearing aids incorporates two aspects, namely, the audiological and the technical point of view. The former focuses on items like the recruitment phenomenon, the speech intelligibility of hearing-impaired persons, or just on the question of hearing comfort. Concerning these subjects, different algorithms intending to improve the hearing ability are presented in this paper. These are automatic gain controls, directional microphones, and noise reduction algorithms. Besides the audiological point of view, there are several purely technical problems which have to be solved. An important one is the acoustic feedback. Another instance is the proper automatic control of all hearing aid components by means of a classification unit. In addition to an overview of state-of-the-art algorithms, this paper focuses on future trends.

Keywords and phrases: digital hearing aid, directional microphone, noise reduction, acoustic feedback, classification, compression.

1. INTRODUCTION

Driven by the continuous progress in the semiconductor technology, today's high-end digital hearing aids offer powerful digital signal processing on which this paper focuses. Figure 1 schematically shows the main signal processing blocks of a high-end hearing aid [1]. In this paper, we will follow the depicted signal flow and discuss the state of the art, the challenges, and future trends for the different components. A coarse overview is given below.

First, the acoustic signal is captured by up to three microphones. The microphone signals are processed into a single signal within the directional microphone unit which will be discussed in Section 2.

The obtained monosignal is further processed separately for different frequency ranges. In general, this requires an analysis filterbank and a corresponding signal synthesis. The main frequency-band-dependent processing steps are noise reduction as detailed in Section 3 and signal amplification combined with dynamic compression as discussed in Section 4.

A technically challenging problem of hearing aids is the risk of acoustic feedback that is provoked by strong signal amplification in combination with microphones and receiver being close to each other. Details regarding this problem and possible solutions are discussed in Section 5. Note that feedback suppression can be applied at different stages of the signal flow dependent on the chosen strategy. One reasonable solution is shown in Figure 1, where feedback suppression is applied right after the (directional) microphone unit.

Almost all mentioned hearing aid components can be tuned differently for optimal behavior in various listening situations. Providing different "programs" that can be selected by the hearing impaired is a simple means to account for this difficulty. However, the usability of the hearing aid can be significantly improved if control of the signal processing algorithms can be handled by the hearing aid itself. Thus, a classification and control unit, as shown in the upper part of Figure 1 and described in Section 6, is required and offered by advanced hearing aids.

The future availability of wireless technologies to link two hearing aids will facilitate binaural processing strategies involved in noise reduction, classification, and feedback reduction. Some details will be provided in the respective sections.

2. DIRECTIONAL MICROPHONES

One of the main problems for the hearing impaired is the reduction of speech intelligibility in noisy environments, which is mainly caused by the loss of temporal and spectral resolution in the auditory processing of the impaired ear. The loss

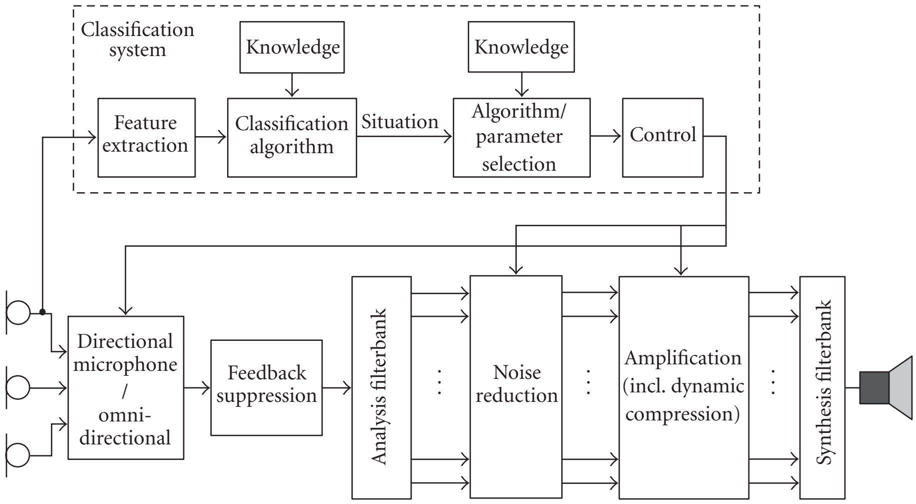


Figure 1: Processing stages of a high-end hearing aid.

in signal-to-noise ratio (SNR) is estimated to be about 4–10 dB [2]. Additionally, the natural directivity of the outer ear is not effective when behind-the-ear (BTE) instruments are used. To compensate for these disadvantages, directional microphones have been used in hearing aids for several years and have proved to significantly increase speech intelligibility in various noisy environments [3].

2.1. First-order differential arrays

In advanced hearing aids, directivity is achieved by differential processing of two nearby omnidirectional microphones in endfire geometry (first-order differential array) to create a direction-dependent sensitivity. As depicted in Figure 2, the signal of the rear microphone is delayed and subtracted from the signal picked up by the front microphone. The directivity pattern of the system is defined by the ratio r of the internal delay T_i and the external delay due to the microphone spacing d (typically 7–16 mm). In this example, the ratio was set to $r = 0.57$ resulting in a supercardioid pattern also shown in Figure 2. To compensate for the highpass characteristic introduced by the differential processing, an appropriate lowpass filter (LPF) is usually added to the system.

Compared to conventional directional microphones utilizing a single diaphragm with two separate sound inlet ports (and an acoustic damper to introduce an internal time delay), the advantage of this approach is that it allows to automatically match microphone sensitivities and that the user can switch to an omnidirectional characteristic, when the direction of the target signal differs from the assumed zero-degree front direction, for example, when having a conversation in a car.

To protect the amplitude and phase responses of the microphones against mismatch caused by aging effects (e.g., loss of electric charge in electret) or environmental influences (condensed moisture and smoke on microphone membrane, corrosion due to aftershave and sweat, etc.), adaptive matching algorithms are implemented in high-end hearing aids.

The performance of a directional microphone is quantified by the directivity index (DI). The DI is defined by the power ratio of the output signal (in dB) between sound incidence only from the front and the diffuse case, that is, sound coming equally from all directions. Consequently, the DI can be interpreted as the improvement in SNR that can be achieved for frontal target sources in a diffuse noise field. The hypercardioid pattern ($r = 0.34$) provides the best directivity with a DI of 6 dB, which is the theoretical limit for any two-microphone array processing [4]. However, in practical use, these DI values cannot be reached due to shading and diffraction effects caused by the human head. Figure 3 illustrates the impact of the human head on the directivity of a BTE with a two-microphone array. The most remarkable point is that the direction of maximum sensitivity is shifted aside by approximately 40 degrees, if the device is mounted behind the ear of a KEMAR (Knowles Electronic Manikin for Acoustic Research). Consequently, the DI, which is related to the zero-degree front direction, decreases typically by 1.5 dB compared to the free-field condition.

The performance related to speech intelligibility is quantified by a weighted average of the DI across frequency, commonly referred to as the AI-DI. The weighting function is the importance function used in the articulation index (AI) method [5] and takes into account that SNR improvements in different frequency bands contribute differently to the speech intelligibility. As shown in Figure 4 for a hypercardioid pattern, the AI-DI (as measured on KEMAR) of two microphone arrays in BTE instruments ranges from 3.5 to 4.5 dB. For speech intelligibility tests in mainly diffuse noise, the effect of directional microphones typically leads to improvements of the speech reception threshold (SRT) in the range from 2 to 4 dB (e.g., [6]).

In high-end hearing aids, the directivity is normally adaptive in order to achieve a higher noise suppression effect in coherent noise, that is, in situations with one dominant noise source [2, 7]. As depicted in Figure 5, the primary

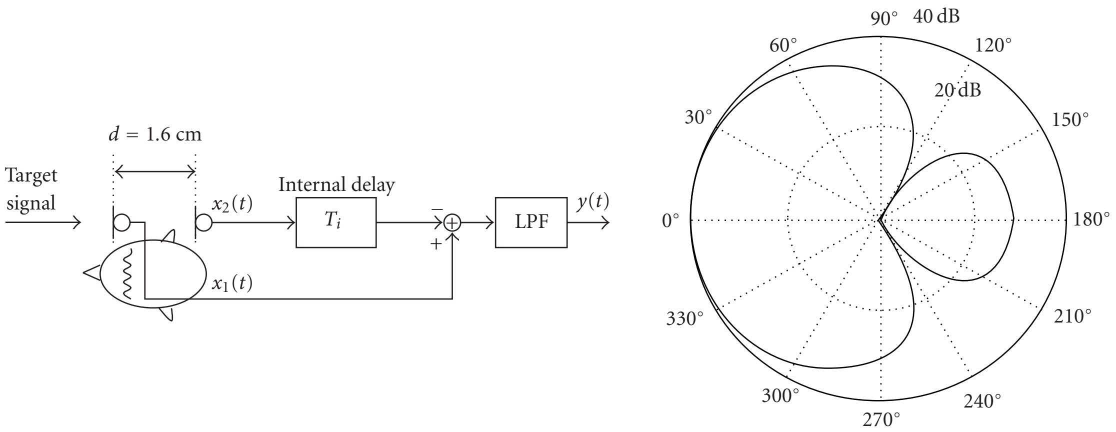
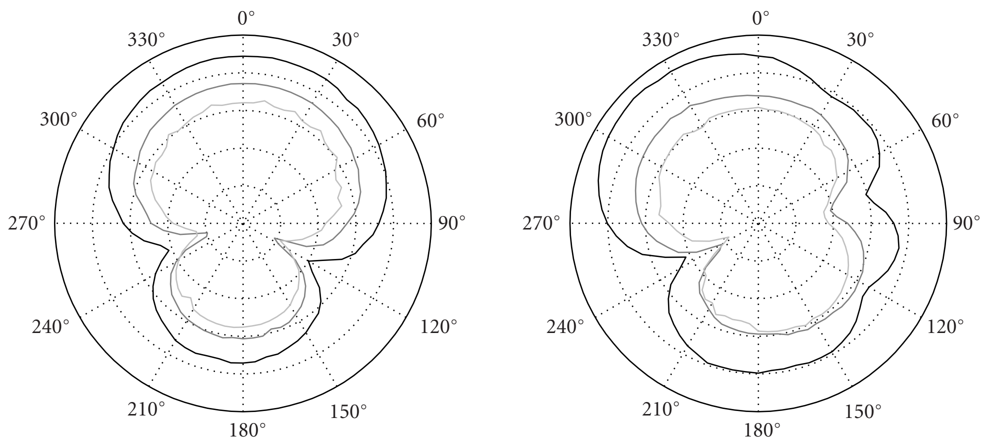


FIGURE 2: Signal processing of a first-order differential microphone.



(a) (b)

FIGURE 3: Impact of head shadow and diffraction on the directivity pattern of a BTE with a two-microphone differential array (a) in free field and (b) mounted behind the left ear of a KEMAR. The black, dark gray, and light gray curves show the directivity pattern for 2 kHz, 1 kHz, and 500 Hz, respectively (10 dB grid).

direction from which the noise arrives is continually estimated and the directivity pattern is automatically adjusted so that the directivity notch matches the main direction of noise arrival. Instead of implementing computationally expensive fractional delay filters, the efficient method proposed by Elko and Pong [8] can be used. In this approach, the shape of the directivity pattern is steered by a weighted sum of the output signals of a bidirectional and a cardioid pattern. The position of the directivity notch is monotonically related to the weighting factor. Great demands are made on the adaptation algorithm. The steering of the directional notch has to be reliable and accurate and should not introduce artefacts or perceivable changes in the frequency response for the zero-degree target direction, which would be annoying for the user. The adaptation process must be fast enough ($<$ 100 milliseconds) to compensate for head movements and to track moving sources in common listening situations, such as conversation in a street cafe with interfering traffic noise. To ensure that no target sources from the front hemisphere are suppressed, the directivity notches are limited to the back hemisphere (90°–270°). Finally, the depth of the notches is limited to prevent hazardous situations for the user, for example, when crossing the street while a car is approaching.

Figure 5 shows a measurement in an anechoic test chamber with an adaptive directional microphone BTE instrument mounted on the left KEMAR ear. A noise source was moved around the head and the output level of the hearing aid was recorded (dashed line). Compared to the same measurement for a nonadaptive supercardioid directional microphone (solid line), the higher suppression effect for noise incidence from the back hemisphere is clearly visible.

2.2. Second-order arrays

The latest development is the realization of a combined first- and second-order directional processing in a hearing aid with three microphones [7], which is shown in Figure 6. Due to

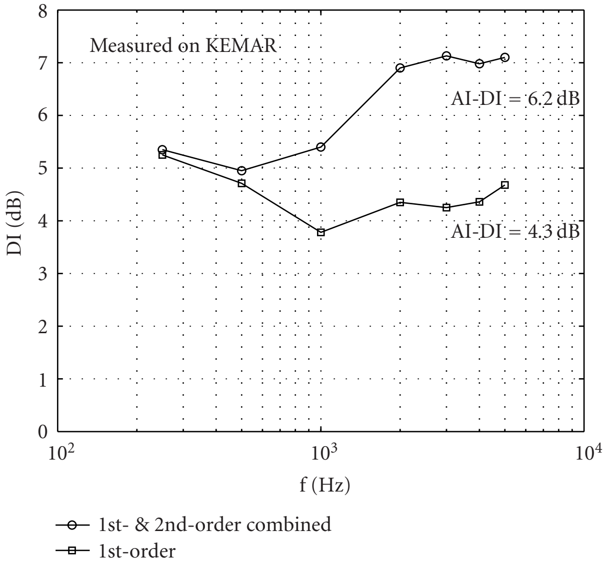


Figure 4: DI and AI-DI for a fist-order array (Siemens Triano S) and the combination with a second-order array in the upper frequency range (Siemens Triano 3).

the high sensitivity to microphone noise in the low frequency range, the second-order processing is limited to the frequencies above approximately 1 kHz which are most important for speech intelligibility.

As shown in Figure 4, calculation of the AI-DI leads to values of 6.2 dB, that is, an improvement in AI-DI of about 2 dB compared to a first-order system. It should be noted that for many listening situations, improvements of 2 dB in the AI-DI can have a significant impact on speech understanding [9].

2.3. Challenges and future trends

Although today's directional microphones in hearing aids provide a significant improvement of speech understanding in many noisy hearing situations, there are still several open problems and ways for further improvement. Some of these are outlined below.

2.3.1. Extended (adaptive) directional microphones

In the past decade, various extended directional microphone approaches have been proposed for hearing aid applications in order to increase either the directional performance or the robustness against microphone mismatch or head shadow effects, for example, adaptive beamformers (e.g., [10, 11, 12, 13]), beamformer taking head shadow effects into account [14], and blind source separation techniques (e.g., [15, 16]).

Adaptive beamformers can be considered as an extension of differential microphone arrays, where elimination of potential interferers is achieved by adaptive filtering of several microphone signals. Usually the adaptation needs to be constrained such that the target signal is not affected.

An attractive realization form of adaptive beamformers is the generalized sidelobe canceller (GSC) structure [17].

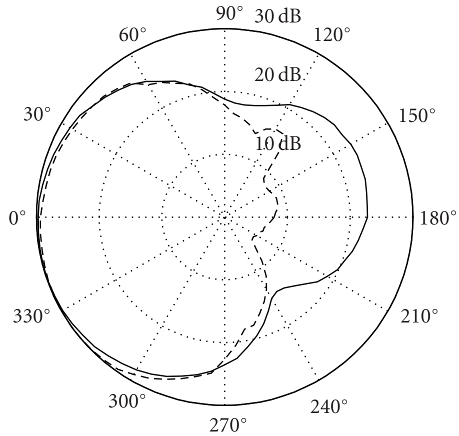


Figure 5: Suppression of a noise source moving around the KEMAR for a BTE instrument (mounted on left ear) with directional microphone in adaptive mode (dashed line) and nonadaptive mode (solid line).

Here, the underlying idea is to split the constrained adaptation into an unconstrained adaptation of the noise reduction and a fixed (nonadaptive) beamformer for the target signal.

An extension is the TF-GSC where transfer functions (TF) from the source to the microphones can be included in the concept [18]. Multiple microphones on each side of the head can be used to increase the number of possible spatial notches to suppress unwanted directed sound sources. The fixed filter-and-sum beamformer can also be designed for lateral target signal directions. This makes sense when the target signal beamformer is adaptive so that it is able to follow the desired speaker.

One crucial problem of the application of the TF-GSC approach for hearing aids occurs when the wearer turns his head, since the beamformer has to adapt again. However, the hearing aid does not know which the desired sound source is. Note that this difficulty is common to all algorithms forming an adaptive beam. In standard directional microphone processing, this problem is circumvented by defining the frontal direction as the direction of the desired sources. Although this strategy has proved to be practical, the directional benefit in everyday life is limited due to this assumption. Examples for critical situations are conversation in a car or with a person one is sitting next to at a table. Thus, sophisticated solutions for selecting the desired source (direction) have to be developed.

2.3.2. Binaural noise reduction

So far, algorithms for microphones placed in one device have been discussed. However, future availability of a wireless link between a left and a right hearing aid gives the opportunity to combine microphone signals from both hearing aids. Envisioned algorithms are, for instance, the binaural spectral subtraction [19] or the "cocktail-party" processors, which mimic some aspects of the processing in the human ear (e.g., [20, 21]).

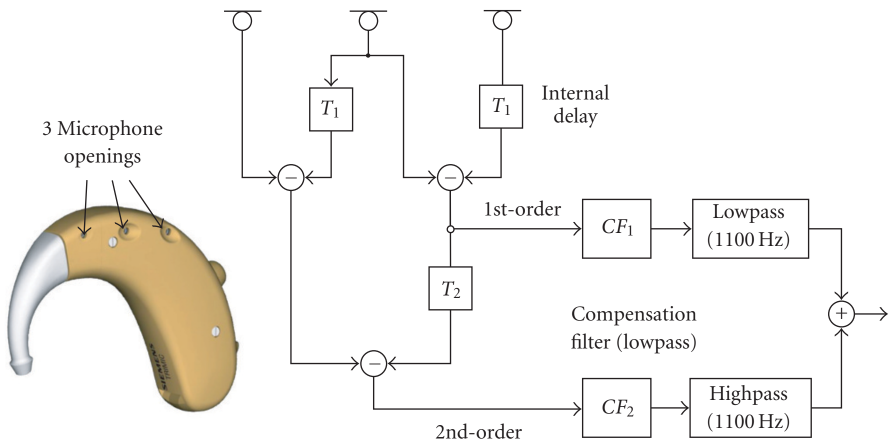


Figure 6: Combined first- and second-order processing in a behind-the-ear (BTE) hearing aid with three microphones.

The binaural spectral subtraction [19] utilizes cross-correlation analysis of the two microphone signals for a more reliable estimation of the monaural noise power spectrum without requiring stationarity for the interfering noise as the single-microphone versions do. An interesting variant of the binaural noise-power estimator assumes the noise field only to be diffuse and the microphones to pick up mainly direct sound of the target source. That means the hearing aid user must be located inside the reverberation radius of the target source. Consequently, in contrast to most other multi-microphone approaches, no specific direction of arrival is required for the target signal. It is expected that due to the minimal need of head alignment, this will be more appropriate in noisy situations with multiple target sources, for example, talking to nearby persons in a crowded cafeteria.

Another approach is to combine the principles of binaural spectral subtraction and (monaural) differential arrays (see Section 2.1). The advantage arises from the fact that the SNR improvement due to the differential arrays in both hearing aids improves the condition for the sequencing binaural spectral subtraction algorithm. By means of this combination, an efficient reduction of localized and diffuse noise is possible.

Further, binaural noise reduction can be achieved by extending monaural noise reduction techniques like those described in Section 3.3. The statistical model for the speech spectral coefficients can be extended to two dependent random variables, the left and the right spectral amplitude, forming a two-dimensional distribution. However, it has to be investigated whether the performance increase justifies the larger effort regarding computational requirements and the need for a wireless link.

In several cases, it is also possible to apply extended multimicrophone algorithms, for example, the TF-GSC outlined in the previous subsection, for binaural noise reduction. However, one problem for potential users is that such algorithms usually deliver only a monaural output signal so that the residual binaural hearing ability of the hearing impaired cannot be exploited.

2.3.3. *Directivity loss for low frequencies*

The effectiveness of a directional microphone might be reduced in the lower frequency range due to the vent of the ear mold, which is often necessary to reduce moisture build-up and the occlusion effect (occlusion effect: bad sound quality of the own voice if the ear canal is occluded). Sound passes through the vent in the ear canal, thus bypassing the hearing aid processing. A promising approach for future hearing aids is the use of active-noise-cancellation techniques, that is, to estimate the vent transmitted sound and to cancel it out by adding a phase inverted signal to the hearing aid receiver. One challenge will be to reliably estimate the transfer function from the hearing aid microphone through the vent in the ear canal. With this transfer function, the vent-transmitted sound can be calculated from the hearing aid microphone signal.

3. NOISE REDUCTION

Directional microphones, as described in the preceding section, are usually not applicable to small ear canal instruments for reasons of size constraints and the assumption of a free sound field which is not met inside the ear canal. Consequently, one-microphone noise reduction algorithms became an essential signal processing stage of today's high-end hearing aids. Due to the lack of spatial information, these approaches are based on the different signal characteristics of speech and noise. Usually, despite the fact that these methods may improve the SNR, they could not yet prove to enhance the speech intelligibility.

In the following, several noise reduction procedures will be described. The first method is also one of the early ones in the field. It decomposes the noisy signal into many subbands and applies a long-term smoothed attenuation to those subbands for which the average SNR is very low. The second Wiener-filter-based method applies a short-term attenuation to the subband signals and is thus able to enhance the SNR even for those signals for which the desired signal and the noise cover the same frequency range. The Ephraim-Malah-based approach, outlined in the third subsection, is comparable to the Wiener-filter-based approach, but exploits a more elaborated statistical model.

3.1. Long-term smoothed, modulation frequency-based noise reduction

The aim of this noise reduction method, which is one standard method for today's hearing aids, is to attenuate frequency components with very low SNR. To distinguish subbands which contain desired signal components from only noise subbands, the modulation frequency analysis can successfully be applied [22]. The modulation frequency analysis determines—generally speaking—the spectrum of the envelope of the respective subband signals. Not only *speech*, but also music exhibits much higher values of the modulation frequency around 4 Hz compared to pure noise, especially stationary noise. Thus, based on this value, a long-term attenuation can be determined to attenuate the subbands with a very low SNR [23]. The disadvantage of this method is that SNR enhancement is better achieved when the desired signal and noise components are located in different frequency ranges. This may reduce the subjectively observed noise reduction performance.

3.2. Wiener-filter-based, short-term smoothed noise reduction methods

The aim of these noise reduction procedures is to obtain significant noise reduction performance even for signals whose desired signal and noise components are located in the same frequency range.

Applying the Wiener-filter attenuation

$$H(l,k) = \frac{S_{ss}(l,k)}{S_{ss}(l,k) + S_{nn}(l,k)} = 1 - \frac{S_{nn}(l,k)}{S_{xx}(l,k)}, \qquad (1)$$

where l and k denote the time and frequency indices in many subbands and utilizing short-term estimates for the required power spectral densities $S_{ss}(l,k)$, $S_{nn}(l,k)$, and $S_{xx}(l,k)$ of speech, noise, and noisy speech, respectively, noticeable noise reduction can be obtained. In these cases, the filter coefficients $H(l,k)$ directly follow short-term fluctuations of the desired signal.

However, a high audio quality noise-reduced signal cannot be easily obtained with this method. The main reason is the nonoptimal estimation of power spectral densities which are required in (1). Here, especially the estimation of the noise power spectral density poses problems since the noise signal alone is not available.

In order to nevertheless obtain reliable estimates, well-known methods can be utilized. These are

(i) estimating the noise power spectral density in pauses of the desired signal which requires an algorithm to detect these pauses,
(ii) estimating the noise power spectral density with the minimum statistics method [24] or its modifications [25].

Both methods, however, exhibit a major disadvantage: they only provide long-term smoothed noise power estimates.

However, for power spectral density estimation of the noisy signal, which can easily be obtained by smoothing the subband input signal power, short-term smoothing has to be applied in order that the Wiener-filter gains can follow short-term fluctuations of the desired signal.

Calculating the Wiener-filter gain with differently smoothed power spectral density estimates causes the well-known musical tones phenomenon [26].

To avoid this unpleasant noise, a large number of procedures have been investigated of which the most widely used are

(i) overestimating the noise power spectral density estimates,
(ii) lower-limiting the Wiener-filter values to a minimum, the so-called *spectral floor*.

With the overestimation of the noise power spectral density, short-time fluctuations of the noise no more provoke a random "opening" of the Wiener-filter coefficients—the cause of musical tones.

However, this overestimation reduces the audio quality of the desired signal since especially low-power signal components are more strongly attenuated or vanish due to the overestimation. Limiting the noise reduction to the spectral floor reduces this problem but, unfortunately, also reduces the overall noise reduction performance. Nevertheless, this reduced noise reduction performance is generally preferred against strong audio quality distortion. More sophisticated methods utilize, that is, speech characteristics [27] or masking properties [28] of the ear, to limit the Wiener attenuation and thus reduce the signal distortion without compromising the noise reduction effect too much.

3.3. Ephraim-Malah-based, short-term smoothed noise reduction methods

An alternative approach to the above outlined Wiener-based noise reduction procedures is the MMSE spectrum amplitude estimator which was initially proposed by Ephraim and Malah [29]. The single-channel noise reduction framework estimates the background noise, for example, by the minimum statistics approach. The task of the speech estimator block is to derive the speech spectrum given the observed noisy spectral coefficients which result from a DFT transform of an input signal block.

For the determination of the filter weights, the knowledge of the distribution of the real and imaginary parts of

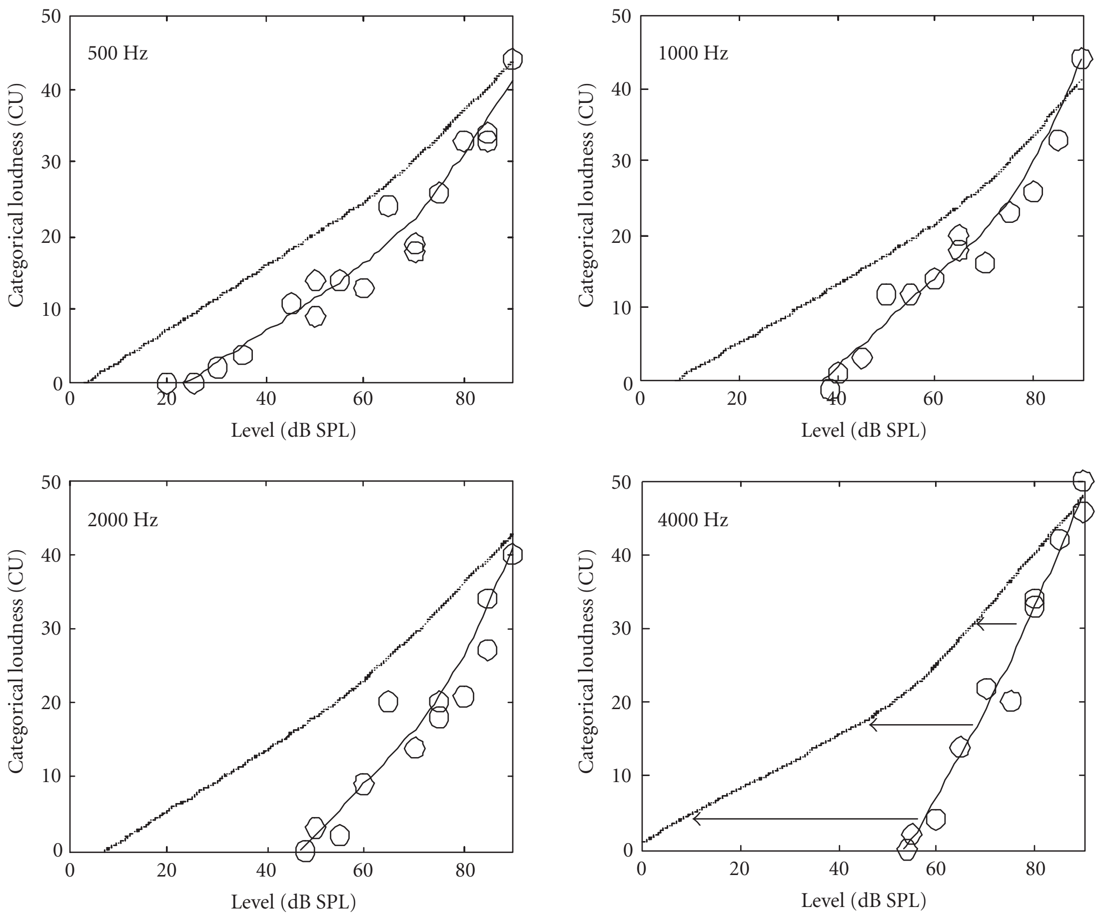


Figure 7: Loudness as a function of level for a hearing-impaired listener (circles) and normal listeners (dashed line).

the speech and noise components is required. They are often assumed as Gaussian [29]. This assumption holds for many noise signals in everyday acoustic environments, but it is not exactly true for speech. A performance investigation for the application in hearing aids can be found, for example, in [30]. Other spectral amplitude estimators for speech can be formulated using super-Gaussian statistical modeling of the speech DFT coefficients [31, 32, 33]. Noise reduction algorithms based on this modified estimator outperform the classical approaches using the Gaussian assumption and are a trend for future hearing aids. The noise reduction effect can be increased at an equal target signal distortion level. A computationally efficient realization has been published [33] which allows a parameterization of the probability density function for speech spectral amplitudes so that an implementation in hearing aids is feasible in the near future.

4. MULTIBAND COMPRESSION

Whereas most signal processing algorithms in hearing aids can also be useful for normal hearing (e.g., noise reduction in telecommunications), multiband compression directly addresses the individual hearing loss. A phenomenon typically observed in sensorineaural hearing loss is "recruitment" [34], which can be measured by categorical loudness scaling procedures (e.g., "Würzburger Hörfeld" [35]) and also could be demonstrated in physiological measurements of basilar membrane velocity [36]. Figure 7 shows the growth of loudness as a function of level for a typical hearing-impaired listener in comparison to the normal hearing reference.

With increasing frequency, the level difference between normal and hearing-impaired listeners for soft sounds ($<$ 10 CU; CU = categorical loudness unit) increases, whereas curves cross at high levels. The arrows in the right bottom graph indicate the necessary level-dependent gain to achieve the same loudness perception at 4 kHz for normal and hearing-impaired listeners. Thus, this measurement directly calls for the need of a frequency specific and level dependent gain—if loudness will be restored to normal. Since more gain is needed for low input levels than for high input levels, the resulting input-output curves of an appropriate automatic gain control (AGC) system have a compressive characteristic.

Restoration of loudness—often also called "loudness normalization"—has been shown, both theoretically [37] and empirically [38], to be capable of also restoring temporal and spectral resolution (as measured by masking patterns) to normal. However, despite many years of research related to loudness normalization [34, 39], the benefits of this approach are difficult to prove [40]. Thus, over the years, many alternative rationales and design goals have been developed resulting in a large variety of AGC systems.

4.1. State of the art

Practically every modern hearing aid employs some form of AGC. The first stage of a multiband AGC is a spectral analysis. In order to restore loudness, this spectral analysis should be similar to the human auditory system (for details see [41]). Therefore, often nonuniform filterbanks are used: constant bandwidth of about 100 Hz up to 500 Hz and approximately 1/3-octave filters above 500 Hz. In each channel the envelope is extracted as input to the nonlinear input-output function.

Depending on the time constants used for envelope extraction, different rationales can be realized. With very slow attack and release times (several seconds), the gain is adjusted to varying listening environments. These systems are often referred to as *automatic volume control* (AVC), whereas systems with fast time constants (several milliseconds) are called "syllabic compression" as they are able to adjust the gain for vowels and consonants within a syllable. For loudness normalization (also of time varying sounds), gains must be adjusted quasi-instantaneously, that is, the gains follow the magnitude of the complex bandpass signals. Moreover, combinations of both slow and fast time constants ("dual compression") have been developed [42].

To avoid a flattening of the spectral structure of speech signals—which is regarded to be important for speech intelligibility—neighboring channels are coupled or the control signal is calculated as a weighted sum of narrowband and broadband level [42]. The input-output function (see component in Figure 8) calculates a time-varying gain which is multiplied by the bandpass signal or the magnitude of the complex bandpass signal prior to the spectral resynthesis stage. There are many rationales to determine the frequency-specific input-output functions from an individual audiogram, for example, loudness restoration (see above), restoration of audibility (DSL i/o [43]), or optimization of speech intelligibility without exceeding normal loudness (NAL-NL1 [44]). The optimum rationale usually depends on many variables like hearing loss, age, hearing aid experience, and actual acoustical situation.

Whereas the above-mentioned AGC systems branch off the control signal before the multiplication of bandpass signal by nonlinear gain ("AGC-i"), output controlled systems ("AGC-o") get the control signal afterwards. AGC-o is often used to ensure that the maximum comfortable level is not exceeded and is thus typically implemented subsequent to an AGC-i. Recently, an AGC-o system has been proposed which is based on percentile levels and keeps the output not only below a maximum level but also above a minimum level in order to optimize audibility [45].

4.2. Future trends

A possibility to cope with situation-dependent fitting rationales is to control the AGC parameters (e.g., attack and release time, input-output function) by the classifier. In a situation where speech intelligibility is most important, for example, a conversation in a crowded restaurant, the appropriate parameters for realizing NAL-NL1 are loaded, whereas when listening to music a setting with optimized sound quality is activated. A wireless link between hearing aids might be beneficial to synchronize the settings on both sides in order to avoid localization problems.

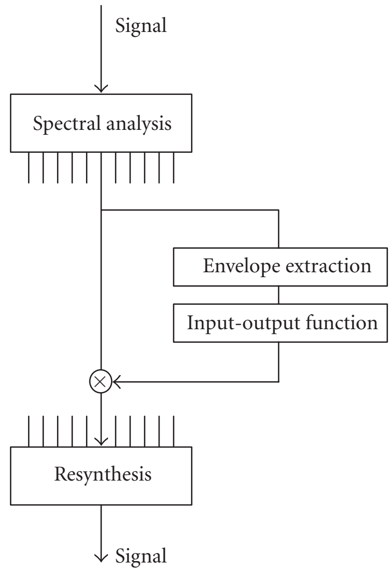


FIGURE 8: Signal-flow for multiband AGC processing.

Another promising scenario is to implement psychoacoustic models (e.g., speech intelligibility, loudness, pleasantness) and use them for a continuous and situation-dependent constrained optimization of the AGC parameters or directly of the time-varying gain. The latter can be realized by estimating the spectra of noise, speech, and the composite signal block by block, similar to the Wiener-filter approach. The speech and noise spectra are used to calculate speech intelligibility (e.g., according to the SII [46]), whereas the overall spectrum is used to determine the current loudness (e.g., according to [37]). Then the channel gains are optimized for each block with the goal to maximize speech intelligibility and the constraint that the aided loudness for the individual hearing-impaired listener does not exceed the unaided loudness for a normal listener. In this case, the hearing aid setting is not optimized for the average male speaker in a quiet surrounding (as is done with NAL-NL1), but for the individual speaker in the given acoustical situation.

5. FEEDBACK SUPPRESSION

Acoustic feedback ("whistling") is a major problem when fitting hearing aids because it limits the maximum amplification. Feedback describes the situation when output signal components are fed back to the hearing aid microphone and are again amplified. In cases where the hearing aid amplification is larger than the attenuation of the feedback path,

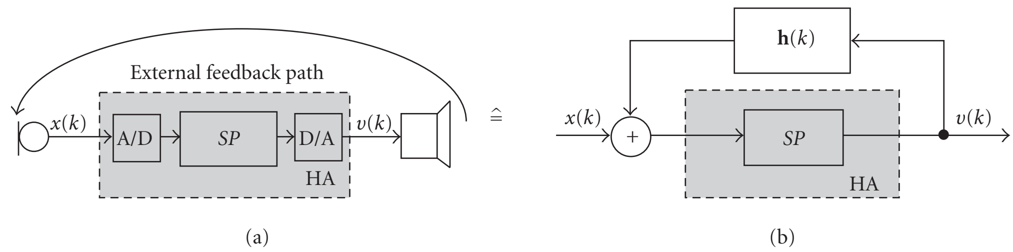


Figure 9: (a) The acoustic coupling between the hearing aid output and its microphone is shown and (b) the corresponding signal model where the acoustic path is modelled as a FIR filter with impulse response $\mathbf{h}(k)$. (HA denotes hearing aid.)

and the feedback signal is in phase, instabilities occur and whistling is provoked. The feedback path describes the frequency response of the acoustic coupling between the receiver and the microphones as depicted in Figure 9.

As described in Section 2.3.3, the occlusion effect can be effectively reduced by ear mold venting. However, increasing the vent diameter automatically increases the feedback risk and lowers the achievable amplification.

Typical hearing aid feedback paths are depicted in Figure 10. Here, one can observe that generally the paths exhibit a bandpass characteristic with the highest amount of coupling at frequency components between 1 and 5 kHz. The typical length of feedback paths which has to be modelled is approximately 64 coefficients for a sampling rate of 20 kHz. The current feedback path is highly dependent on many parameters of which the four most important are

(i) the type of the hearing aid: behind-the-ear (BTE) or in-the-ear (ITE),
(ii) the vent size,
(iii) obstacles around the hearing aid (hands, hats, telephone receivers),
(iv) the physical fit in the ear canal and leaks from jaw movements.

The first two parameters are static whereas the third is highly time-varying during the operation of the hearing aid. In Figure 11, the variance of the feedback paths can be observed in response to changes in the above given parameters.

Corresponding to the time-dependent or static parameters, fixed and dynamic measures are utilized in today's hearing aids to avoid feedback.

A static method is to measure the normal feedback path (without obstacles) once after the hearing aid has been fitted. Limiting the gain of the hearing aid so that the closed-loop gain is smaller than one for all frequency components generally can prevent feedback.

Nevertheless, a totally feedback-free performance of the hearing aid can usually not be obtained without additional measures, especially when the closed-loop gain of the hearing aid in normal situations is close to one. Reflection obstacles such as a hand may then provoke feedback. To avoid this, dynamic methods are necessary for cancelling feedback adaptively when it appears.

For these dynamic measures, two methods are widely spread.

(1) Selectively attenuating the frequency components for which feedback occurs is utilized in today's hearing aids. This method is normally efficient to avoid feedback. However, it is equivalent to a narrowband hearing aid gain reduction.

(2) Another method is the feedback compensation method where the feedback path is modelled with an internal filter in parallel to the feedback path and which subtracts the feedback signal. Thus, the hearing aid gain is not affected by this method. Additionally, it even allows hearing aid gain settings with closed-loop gains larger than one. This method is currently becoming state of the art for hearing aids.

5.1. Feedback cancellation: dynamic and selective attenuation of feedback components

An effective and selective attenuation of feedback components can be reached by notch filters. These notch filters are generally characterized by three parameters: the notch frequency, the notch width, and the notch depth. It is most important to choose the appropriate notch frequency, that is, when feedback occurs, the feedback frequency has to be determined fast and precisely.

Different methods, in the time and frequency domains, are applicable for the estimation of the feedback frequency. These are comparable to methods which can also be found for pitch frequency estimation [47]. These methods are, for example, the zero-crossing rate, the autocorrelation function and the linear predictive analysis. Most important is the fast reaction to feedback but also to apply the notch filters only where and as long as necessary in order to minimize the negative effect of the reduced hearing aid gain.

5.2. Feedback compensation

The reduced hearing aid gain can be totally avoided by the compensation approach. Here, see Figure 12, a filter is internally put in parallel to the external acoustic feedback path. The output of the filter models the feedback signal.

The challenge of this approach is to properly estimate the external feedback path with an adaptive filter. This is hard to realize due to the correlation of the input signal and the signal which is acoustically fed back to the microphones. For

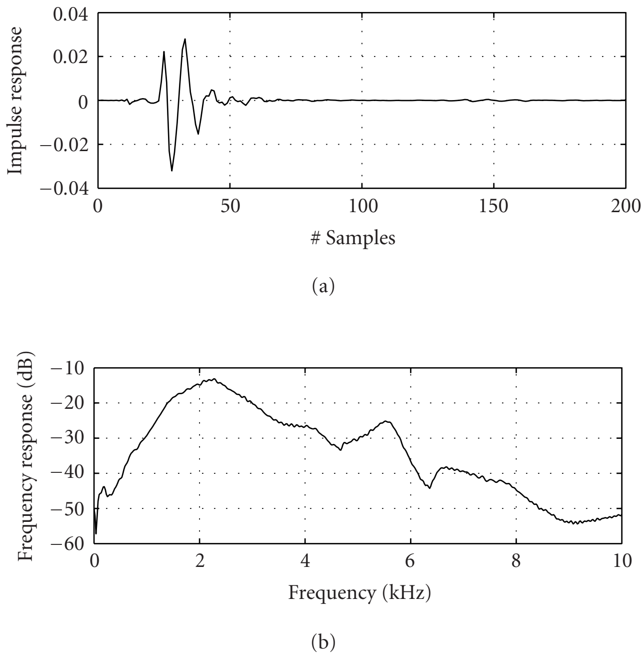


Figure 10: (a) Impulse and (b) frequency responses of a typical hearing aid feedback path sampled at 20 kHz.

reliable estimates of the feedback path, the adaptation has to be controlled by sophisticated methods.

Adaptive algorithms generally estimate the filter coefficients, based on an optimization criterion. The criterion which is very often utilized is the minimization of the mean square error signal, that is, the signal after the subtraction of the adaptive filter's output signal.

In this case, the adaptive filter coefficients converge towards a biased coefficient vector provoked by the correlation of input and output signals [48]. This bias causes a distortion of the hearing aid output and has to be avoided.

Thus, the main objective for enhancing the adaptation should be to reduce this correlation. Here, different methods exist [49]:

(i) decorrelating the input signal with fast-adaptive decorrelation filters,
(ii) delaying the output signal, or
(iii) putting a nonlinear processing unit before the output stage of the hearing aid.

However, none of these methods is a straightforward solution to the given problem, since many problems occur while implementing the proposals. Here, future hearing aids still offer room for improvements.

Additionally, the filter adaptation speed may be explicitly lowered for highly correlated input signals, such as speech or tonal excitation in general, and raised whenever feedback occurs. The distinction between feedback and tonal signals, however, cannot easily be obtained. A solution approach will be shown in the next section.

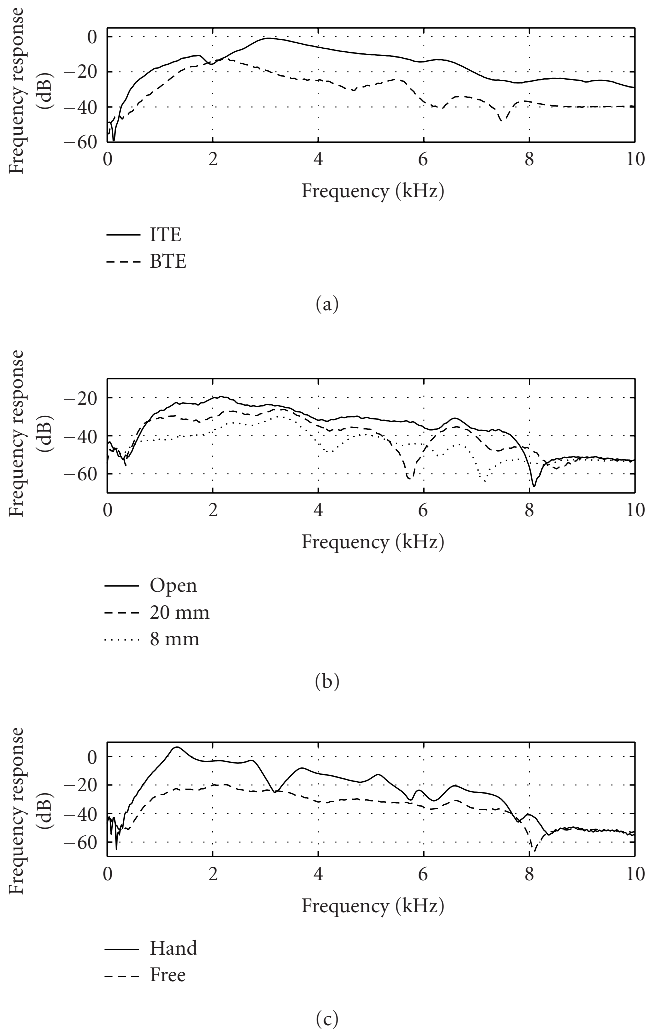


Figure 11: Typical feedback paths for different types of (a) hearing aids, (b) different vent sizes, and (c) obstacles, that is, a hand near the hearing aid compared to the normal situation.

5.3. Future trends

Alternative and future approaches may benefit from the fact that hearing-impaired individuals generally utilize hearing aids on both sides of the head. Thus, the robustness against sinusoidal or narrowband input signals can be improved. One promising approach is the binaural oscillation detector depicted in Figure 13. The basic idea is that oscillations detected by one hearing aid can only be caused by feedback if the hearing aid on the other side did not detect oscillations of exactly the same frequency. Obviously, this approach makes use of the head shadow effect and needs a data link between both hearing aids.

6. CLASSIFICATION

Hearing aid users encounter a lot of different hearing situations in everyday life, for example, conversation in quiet or

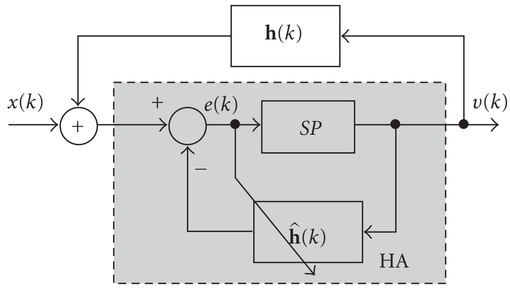


Figure 12: General setup of a feedback cancellation system with *SP* modeling the hearing aid signal processing, $\mathbf{h}(k)$ the external feedback path, $\hat{\mathbf{h}}(k)$ the adaptive filter.

in noise, telephone calls, being in a theater or in road traffic noise. They expect real benefits from a hearing aid in each of the mentioned situations. As was shown in the previous part of this paper, modern digital hearing aids provide multiple signal processing algorithms and possible parameter settings, for example, concerning directivity, noise reduction, and dynamic compression. This portfolio of algorithms is expected to still grow with increasing IC computational power. Single algorithms and their multitude of possible parameter settings are mostly working in a situation-specific way, that is, these algorithms are beneficial in certain hearing situations whereas they have no or even negative impact in other situations. For example, noise reduction algorithms as described in Section 3 reduce stationary background noise efficiently, whereas they may have some negative influence on the sound of music and should therefore be disabled in such situations. Even if the optimal signal processing algorithm for any relevant situation would be available, the problem to activate it reliably in the current specific hearing situation remains. A promising solution for this problem is to use a classification system, which can be understood as a superordinate, intelligent algorithm that continuously analyzes the hearing situation and automatically enables the optimal hearing aid setting. The alternative would be a great number of situation-specific hearing aid programs, which have to be chosen manually. However, this approach would certainly overextend the mental and motor abilities of many hearing aid users, especially for the small ITE devices, and therefore, seems not to be a very attractive alternative [50].

6.1. Basic structure of monaural classification

Figure 1 shows the basic structure of a digital hearing aid with a superordinated classification system controlling the different signal processing blocks like directional microphone, noise reduction, shaping of the frequency response, and dynamic compression. Classification systems consist of different functional stages.

As a first step, "features" are extracted from the microphone signal. "Features" are certain properties of the signals, whose magnitude is as different as possible for selected situation classes like "speech in quiet," "(speech in) noise," or "music" and can therefore be used to distinguish between situation classes. In literature several spectral and temporal features have been proposed, mostly in the context of separation of "speech in quiet" and "speech in noise": profile and temporal changes of the frequency spectrum [51, 52, 53], statistical distribution of signal amplitudes [54], or analysis of modulation frequencies [55].

To illustrate the principle of feature extraction, Figure 14 shows the extraction of a modulation feature from three different signals belonging to the classes "speech in quiet," "speech in noise," and "music." The fluctuations of the signal envelope which are calculated by taking the absolute value and lowpass filtering are called "modulation." Typical for speech are strong modulations in the range of 1–4 Hz. The magnitude curves of this feature for the three examples, as depicted in Figure 14, show that values of this feature are obviously higher for "speech in quiet" than for the other signals. Consequently, the modulation feature allows to separate "speech in quiet" from "speech in noise" and "music," whereas separation of "speech in noise" and "music" is not possible due to similar feature values. Therefore, most applications of classification techniques require the simultaneous evaluation of a larger number of features to ensure sufficient decision reliability. The assignment of feature values and their combinations to the different classes can be achieved with standard approaches like the Bayes classifier [55] or neural networks [52]. These algorithms learn the necessary a priori knowledge about the relationship between feature values and situation classes in appropriate training procedures, which have to be based on large and representative databases of everyday life signals.

The adaption of the hearing aid signal processing to the detected listening situation is divided into two parts as shown in Figure 1. The block "selection of algorithm and parameters" contains an "action matrix" describing which of the settings for the algorithms and parameters are optimal in each situation. The definition of the action matrix is based on detailed knowledge of the properties of the particular algorithms in the different situations. Extensive investigations and tests are the base for this knowledge. Every time the detected situation class is updated, the next block generates "on/off"-control signals for each hearing aid algorithm. Sudden "off/on"-switching of signal processing components like the directional microphone are considered as irritating and unpleasant. Thus, appropriate fading mechanisms which realize a gliding smooth transition from one state of operation to another are advantageous. In many cases, this can easily be achieved by lowpass filtering of the control signals. Figure 15 illustrates the fading from omnidirectional to directional microphone mode.

6.2. Future trends

Using multimicrophone signals is the most important step from classification based on the statistical information of one microphone signal towards a future sound scene classification [56]. Typical situations where single-signal-based classification systems fail are, for example, listening to music from the car radio while driving or conversing in a cafe with background music. To classify these situations correctly so that

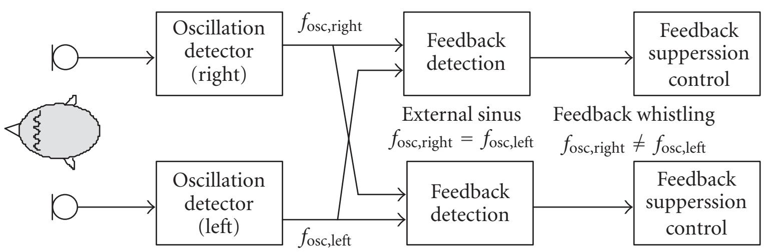


Figure 13: Binaural oscillation detection for feedback suppression.

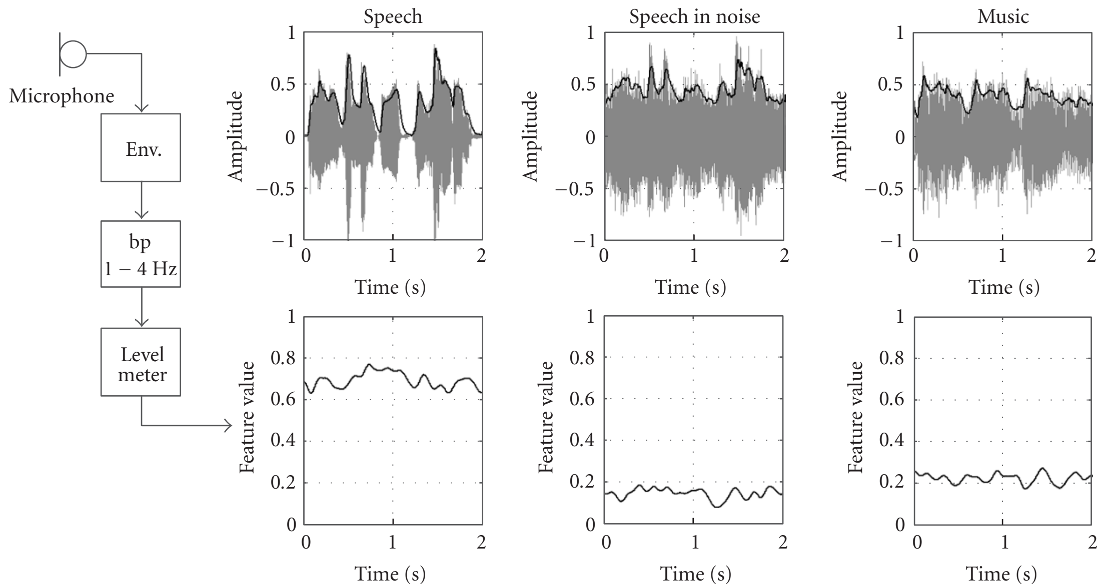


Figure 14: Example for the calculation of a modulation feature.

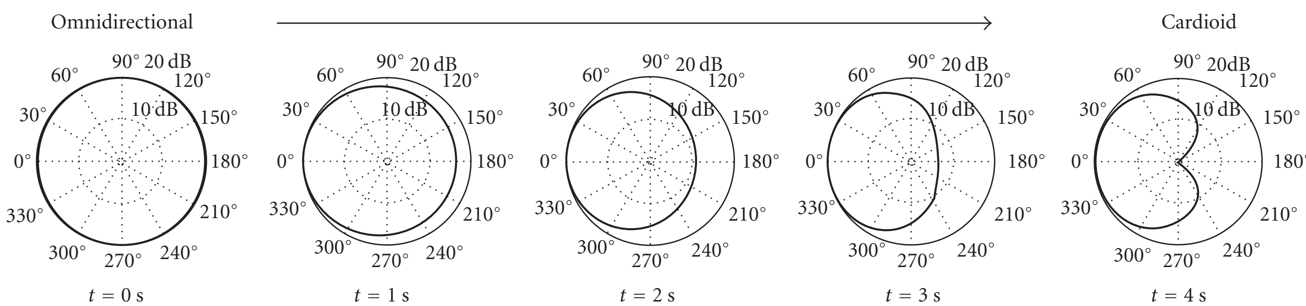


Figure 15: Fading from omnidirectional to directional microphone mode.

the algorithms can take advantage of the result requires information about the sound incidence direction, and the number, distance, and type of sound sources in the room. This information can be derived from future multimicrophone localization and classification algorithms. Methods known from the computational auditory scene analysis (CASA) [57] can be used to further develop today's classification systems. For example, simultaneous speech sources in noisy environment can be recognized by pitch tracking [58].

7. SUMMARY

The development of hearing aids covers a wide range of different signal processing components. They are mainly motivated by audiological questions. This paper focuses on algorithms dealing with the compensation of the recruitment phenomenon, the improvement of speech intelligibility, and the enhancement of comfort while using the hearing aid in everyday life.

As one important component of hearing aids, the directional microphone and its effect on the improvement of speech intelligibility is discussed. Directional microphones of different complexities like first-order differential arrays, second-order arrays, and adaptive beamformers are discussed.

One component which mainly focuses on the improvement of comfort is the noise reduction unit. Algorithms of different complexities with different amounts of statistical a priori knowledge concerning the computed signal and different speeds of reaction are described. Noise reduction algorithms which exploit the binaural wireless link of future high-end digital hearing aids are discussed as well.

A significant unit in hearing aids is the AGC which compensates the recruitment phenomenon. This paper discusses state-of-the-art systems and future trends.

Another important aspect is the feedback phenomenon which may occur at high levels of amplification in the hearing aid. This paper presents approaches to reduce feedback by means of different feedback suppression units.

Finally, the ability of modern hearing aids to detect different hearing situations and to properly control the interaction of all involved algorithms is discussed.

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