Improving speech intelligibility in hearing aids. Part I: Signal processing algorithms

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Abstract

The improvement of speech intelligibility in hearing aids is a traditional problem that still remains open and unsolved. Modern devices may include signal processing algorithms to improve intelligibility: automatic gain control, automatic environmental classification or speech enhancement. However, the design of such algorithms is strongly restricted by some engineering constraints caused by the reduced dimensions of hearing aid devices. In this paper, we discuss the application of state-of-the-art signal processing algorithms to improve speech intelligibility in digital hearing aids, with particular emphasis on speech enhancement algorithms. Different alternatives for both monaural and binaural speech enhancement have been considered, arguing whether they are suitable to be implemented in a commercial hearing aid or not.

Index Terms: Speech enhancement, hearing aids, speech intelligibility.

1. Introduction

Hearing aids are electronic devices worn by hearing-impaired people ideally to improve the reduced intelligibility caused by hearing loss. Despite the fact that traditional devices may improve speech quality or hearing comfort, their capability to improve speech intelligibility has been largely discussed. Simple devices often produce amplified noises when the user is in a multi-source environment (e.g. a crowded bar). Besides the automatic gain control (AGC) system, modern devices include some type of enhancement schema to overcome this limitation, for instance, directional microphones or speech enhancement algorithms. However, in addition to the problems found by current speech enhancement algorithms when improving intelligibility, their application in hearing aids entails three main additional problems: hearing-impaired listeners have greater susceptibility to the distortions introduced by signal processing algorithms, the small size of hearing devices limits the number of microphones assembled in the device, and the reduced life of the current batteries constrains the computational cost of the implemented algorithms.

Nowadays, the hearing aids market is still dominated by monaural systems, which gradually have included several microphones in each device. Nevertheless, the integration of wireless communications in high-end devices has motivated the growing interest in binaural speech enhancement systems. Binaural systems theoretically allow the user to listen more realistically by keeping the spatial information. Unfortunately, the wireless data transmission required between devices originates two new problems: the increment of power consumption and the need of synchronization between devices.

The objective of this paper is to discuss the application of state-of-the-art signal processing algorithms to improve speech intelligibility in digital hearing aids, with particular emphasis on speech enhancement algorithms. The remaining of this paper is organized as follows. Section II describes the hearing impairment problem and the different auditory problems faced by hearing impaired people. Section III gives an overview of signal processing algorithms in hearing aids and quantifies the available computational resources in such devices. Section IV provides a thorough review of the state-of-the-art of speech

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enhancement algorithms, evaluating its application to improve speech intelligibility in monaural and binaural hearing aids. Finally, section V ends with a summary of the conclusions obtained in this study.

2. Hearing impairment

The number of people with hearing loss is increasing at an alarming rate not only because of the aging of the world's population, but also because of the growing exposure to noise in daily life. Some figures confirming these facts are, for instance, that about one-third of Americans between the ages of 65 and 74, and about half the people who are 85 and older, have important hearing loss [1]. Or that about 16% of adult Europeans have hearing problems strong enough to adversely affect their daily life. The royal national institute for deaf people (RNID) has reported that there are 8.7 million deaf and hard of hearing people in the UK, and that just one in four hearing-impaired Britons owns a hearing aid [2]. All these facts compel scientists and engineers to enhance hearing aids in the effort of making them more accessible for people, especially the elderly.

Hearing loss is commonly represented by an audiogram, which shows the auditory threshold in logarithmic units (dB) of the sound pressure level (SPL) for standardized frequencies measured by an audiometer. Hearing impairment implies larger thresholds than normal hearing but the level of loss among frequencies is not uniform and depends on each person. The degree of hearing loss is usually defined as the average hearing loss measured at a particular octave-band, and the level of loss is usually classified into mild (up to 40 dB), moderate (from 40 to 60 dB) and severe (over 60 dB). For hearing-impaired people suffering from mild to moderate hearing loss, a hearing aid is helpful, but in the case of severe hearing loss, the use of hearing aids is of little benefit, and some other solutions such as cochlear implants may be considered. Additionally, hearing loss can be unilateral, but in most cases it is bilateral, which means that both ears are affected with either the same or different degree of loss.

Hearing-impaired people face a variety of different auditory problems that reduce their ability of understanding. These problems are described below [3].

- **Decreased level of audibility**
  Depending on the level of hearing loss, a person will hear some sounds but miss some other sounds. In general, the high-frequency components of speech are weaker than the low-frequency components, and hearing loss of elderly people is higher at high frequencies. Consequently, hearing-impaired people tend to miss high-frequency information, basically consonants. This fact leads to miss essential parts of some phonemes reducing the intelligibility.

- **Reduced dynamic range**
  The dynamic range of the auditory system is defined as the level difference between the auditory threshold and the discomfort threshold (i.e. threshold of pain). For hearing-impaired people, the auditory threshold is increased in comparison to normal hearing people, hence the dynamic range is reduced. In order to avoid exceeding the discomfort threshold, hearing aids must amplify weak sounds more than intense sounds.

- **Reduced frequency resolution**
  Frequency resolution gradually decreases as the degree of hearing loss increases, and hearing-impaired people find difficult to distinguish between sounds of different frequencies simultaneously. This is due to the loss of sensitivity of the hair cells of the cochlea, which decreases the ability of discriminating frequencies.

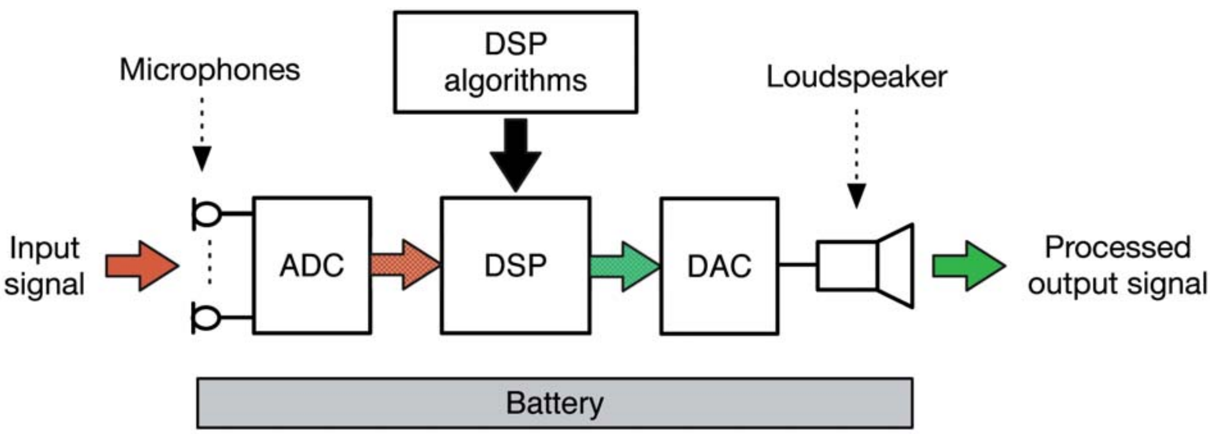

Figure 1. A simplified scheme of the typical structure of a digital hearing aid.

- **Decreased temporal resolution**
  In general, weaker sounds are sometimes masked by intense sounds that immediately precede or follow them, which decreases the chances of intelligibility. In addition, the ability to hear weak sounds during short-time slots gradually decreases as the degree of hearing loss increases, and hearing-impaired people usually experience decreased temporal resolution, which involves that the speech intelligibility perceived by them is further decreased.

All the aforementioned problems combined together cause significant reduction in the speech intelligibility perceived by hearing-impaired people. The first two problems are commonly approached by a multiband compression algorithm, which applies a frequency and signal level dependent gain customized for each person. The intelligibility decrease originated by the reduction in the temporal and frequency resolutions can be compensated by speech enhancement algorithms.

3. Signal processing in digital hearing aids

The introduction of the digital signal processor (DSP) in hearing aids opened a new era where these devices offer their users a greater flexibility to compensate for their hearing loss, providing a more natural sound quality than the previous analog hearing aids. The typical structure of a digital hearing aid is shown in figure 1. The device comprises the next elements:

- A single or multiple microphones that convert the acoustic signal into an electric signal.
- An analog-to-digital converter (ADC) to transform the continuous electric signal (analog) into a digital signal.
- A DSP, the main part of the device, which includes signal processing algorithms for different purposes.
- A digital-to-analog converter (DAC) to reconvert the digital processed signal into an analog signal.
- A tiny loudspeaker that produces the output acoustic signal from the processed analog electric signal.
- A small battery to supply power to the previous electronic devices.

The fact that hearing loss does not only result in sound attenuation, but also in distortions that lead to a reduction in speech intelligibility, motivates that modern digital hearing aids include a variety of signal processing algorithms for different purposes:

- A multiband compression algorithm to compensate hearing loss and fit the output level into the dynamic range. The gain is automatically adjusted by the AGC system according to the individual hearing loss and the input level.
- Speech enhancement algorithms that aim to improve the speech intelligibility provided by hearing aids in different noisy environments.
- Automatic environmental classification in order to adapt the amplification or processing program to different listening conditions (e.g. a quiet room, a conference hall, a noisy street, etc.). Both the multiband compression and speech enhancement algorithms can be adjusted depending on the environment.
- Acoustic feedback cancellation to prevent the instability of the device due to the acoustic feedback that appears when part of the amplified output signal produced by the hearing aid returns through the external auditory canal and enters again the device, thus being again amplified.

Figure 2 shows the different signal processing blocks in a state-of-the-art hearing aid. All these algorithms must be implemented in the DSP embedded in the hearing aid. Un-

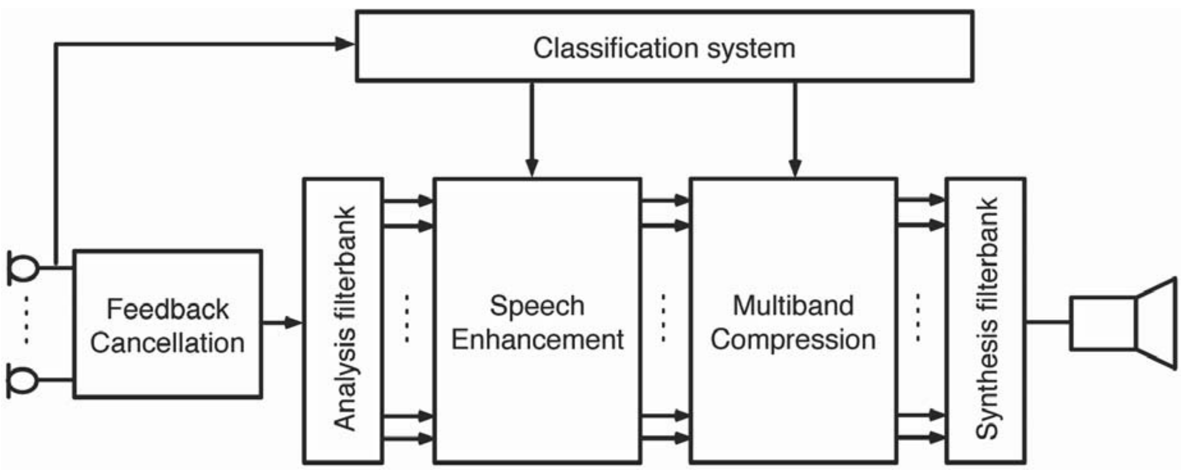

Figure 2. Signal processing algorithms in a state-of-the-art hearing aid.

The introduction of the digital signal processor (DSP) in hearing aids opened a new era where these devices offer their users a greater flexibility to compensate for their hearing loss, providing a more natural sound quality.

fortunately, the computational capability and the memory available in the DSP of such devices are highly restricted: the processor is forced to work at low-clock frequencies in order to minimize the power consumption and thus to maximize the battery life. The current batteries available for hearing aids and the expectation of a minimum battery life of one week entail that the DSPs found in state-of-the-art commercial devices have on-chip processors with a selective clock speed that usually goes from 5.12 MHz down to 1.28 MHz, which is a relative low speed in comparison to the current DSPs that can be used in other applications. For instance, in the special case of a processor with a clock speed of 5.12 MHz (5 MIPS), and working with a sampling rate of 16 kHz, analysis window of 128 samples with 50% of overlap, and 65 frequency bands, the number of instructions available to process each frequency band of a frame is 308. These instructions are shared between the aforementioned signal processing algorithms included in the device. The time-frequency analysis is based on a DFT filter bank and usually implemented in a specific processor, hence it does not imply any extra consumption of computational resources.

The next section is focused on providing a review of the state of the art of the application of speech enhancement algorithms in hearing aids, which among the algorithms included in hearing aids, are those directly included to increase speech intelligibility.

4. Speech enhancement algorithms for hearing aids

Imagine an elderly grandmother who wears a hearing aid in one or both ears. She is in a room where her family is celebrating her birthday. There are so many talks, music, the TV, and background noise mixing with each other that the old lady cannot understand what her grandson is telling her. The solution to this problem would be that the hearing aids themselves were able to enhance only the voice of the grandson separately from the rest of the sounds without interest. The inclusion of speech enhancement algorithms in modern devices aims to solve this problem. However, the design and implementation of this type of algorithms in digital hearing aids is strongly limited by some engineering constraints, which are not present in other speech enhancement applications such as hands-free devices or automatic speech recognition (ASR) systems. As mentioned before, there are several signal processing algorithms running simultaneously in the DSP of modern digital hearing aids, trying to solve different problems. These algorithms demand a significant part of the computational power of the device, and at the same time, electrical power. Bearing in mind the limited power of the processor, the computational cost of the algorithms used for speech enhancement must be very low, taking only a small part of the available computational resources.

Many hearing-impaired people have bilateral hearing loss and they are forced to wear two devices. Often, when hearing aids are worn at both ears, these devices operate independently. However, there is a new trend of binaural hearing aids that connects both devices in order to exchange information between them. Binaural hearing provides considerable benefits over using only one ear, due to the fact that the nature of the human auditory system is binaural. Humans are able to separate and selectively attend to individual sound sources in a cluttered acoustical environment taking advantages of the so-called spatial cues. Hence, it is fundamental that the hearing aid system preserves these cues, which notably increments the ability to localize sounds and consequently improves speech intelligibility. This obviously requires a communication link between both hearing devices. The simplest solution would be to connect them using a wire. However, most users do not like this approach because of the non-aesthetic aspect of the wire linking both hearing aids from one ear to the other. This enforces to use a wireless link between both devices, what unavoidably increases the power consumption and, consequently, reduces the battery life, one of the most important limiting factors for implementing signal processing algorithms on digital hearing aids. Roughly speaking, the current technology demands as much power to communicate both hearing aids as that required for the signal processing on a monaural device [3]. The reduction of the data rate helps cut down the power consumption, but it is done at the expenses of bringing down the performance of the enhancement algorithms.

Directional microphones have been used in hearing aids for over 25 years and have proved to significantly increase speech intelligibility in various noisy environments [4]. However, they are usually not applicable to small in-the-canal (ITC) devices for reasons of size and the assumption of a free sound field which is not met inside the ear canal. Nevertheless, directional microphones are not under the scope of this paper. A comprehensive review is given in [5].

Besides directional microphones, modern hearing aids include one or several omnidirectional microphones combined with speech enhancement algorithms to improve intelligibility. Noise reduction and sound source separation are two different approaches that may be applied to enhance speech. An exhaustive review of single-channel and multichannel speech enhancement algorithms can be found in [6]. The implementation of algorithms for speech enhancement in hearing aids presents particular challenges:

- The requirement of real-time processing limits the processing delay to few milliseconds, which in turns limits the algorithmic complexity. A delay between the direct sound and the amplified sound may be perceived as degraded sound quality or the perception of an echo.

- The reduced battery life limits the clock speed of the processor, which also limits the computational capability of the device.

- The number of microphones in multichannel systems is reduced due to the dimensions of the device. For instance, considering ITC devices, which have an ample role in the market, their shape can be approximated by a cylinder 1.5 cm in length and 1 cm in diameter. Common omnidirectional microphones placed in these devices have a diameter of 0.25 cm, hence, to be realistic, there cannot be more than 4 microphones assembled in each device.

- The number of frequency bands used for the analysis of the input signal is relatively small (usually 64-128 bands).

- Hearing-impaired listeners have greater susceptibility to interference from background noise than normal listeners. They typically require a signal-to-interference ratio (SIR) that is 5-10 dB higher than a normal hearing person in order to achieve the same level of speech understanding [7].

Bearing in mind the above limitations, the remainder of this section discusses the suitability of the different speech enhancement approaches for their implementation in hearing aids. The algorithms are divided into single-channel and multichannel algorithms.

4.1 Single-channel speech enhancement

Single-channel noise reduction in hearing aids is even more challenging than in the general case. Traditional single-channel noise reduction algorithms tend to reduce noise introducing distortions in the signal. Impaired listeners are more sensitive to speech distortions than normal listeners. Consequently, the effect that these distortions have on intelligibility can be minimized for normal listeners but it is magnified for hearing-impaired listeners. Among the single-channel noise reduction algorithms existing in the literature [6], those based on the Wiener filter [8] and the Ephraim-Malah-based approach [9] have been traditionally implemented in hearing aids. Unfortunately, these methods may improve the signal-to-noise ratio (SNR), but they could not yet prove to enhance the speech intelligibility [10]. One of the main reasons is that listeners are more influenced by speech distortions than by background noise. Despite their limitations, single-channel noise reduction systems are still implemented in modern hearing aids.

Besides directional microphones, modern hearing aids include one or several omnidirectional microphones combined with speech enhancement algorithms to improve intelligibility.

Single-channel algorithms for speech separation are dominated by the computational auditory scene analysis (CASA) approach [11]. The separation of sound sources in CASA systems is normally achieved by identifying and grouping spectro-temporal regions in the mixture belonging to the same source, which originates time-frequency binary masks. The application of CASA to single-channel noise reduction consists in generating time-frequency masks to weight the different time-frequency regions, emphasizing regions dominated by the target speech and suppressing regions dominated by noise. However, the proposed methods are either too complex or the performance is too limited to be directly applicable to practical hearing systems. These algorithms typically involve complex operations for feature extraction, segregation and grouping, which makes a real-time implementation difficult. In addition, the performance of such algorithms is not good enough for the implementation in a hearing aid [12]. Nevertheless, the application of time-frequency masking is a promising approach, as long as the mask computation is relatively simple. A conceptually and computationally simpler procedure than the original CASA approach is to estimate the mask using machine learning techniques to identify time-frequency points as either speech-dominated or noise-dominated [13].

Finally, an alternative to improve intelligibility in single-channel devices is the design of enhancement algorithms that optimize objective measures correlated with intelligibility. The work in [14] examines different objective measures for predicting the intelligibility of speech in noisy conditions for normal listeners. Although those measurements may probably correlate with the intelligibility perceived by hearing impaired listeners, they are not completely designed for them. For instance, the hearing-aid speech quality index (HASQI) [15] has been designed to measure intelligibility in both normal-hearing and hearing impaired listeners.

4.2 Multichannel speech enhancement

Recently, high-end hearing aids including multiple microphones have demonstrated to provide reasonable improvements in intelligibility and listening comfort. Multichannel source separation algorithms, such as those based on the independent component analysis (ICA) [16] or clustering [17], have reduced application in hearing aids due to their complexity. Hence, multichannel speech enhancement in hearing aids has been dominated by noise reduction techniques, which is achieved by means of spatial filtering (i.e. beamforming).

Beamforming techniques achieve speech enhancement by using the principle of spatial filtering provided by a microphone array, normally composed of omnidirectional microphones, and assuming that the target source and the unwanted sources are physically separated in space. Spatial filtering aims to boost the signal coming from a determined direction, attenuating the interfering signals coming from different directions. In theory, a microphone array allows reducing the noise without distorting much the speech signal, in opposition to single-channel enhancement algorithms, which usually introduce distortions. Beamforming techniques can be broadly grouped into data-independent (fixed) and data-dependent (adaptive). Data-independent techniques use fixed parameters during the processing of the input signal. On the other hand, data-dependent techniques update their parameters constantly depending on the input signal, adapting to changing noise conditions. Both fixed and adaptive beamformers have been successfully implemented in modern hearing aids. Usually, the array is steered towards the front (look) direction, but in some cases the filters are designed to suppress the interferences coming from the back direction.

The required processing time of fixed beamformers is relatively low, as long as the filter coefficients that satisfy the design constraints can be previously computed and easily included as constant values in the embedded algorithm. An additional advantage is that fixed beamformers are more robust than adaptive beamformers to minor steering errors and reflections correlated with the desired signal. However, their performance is reduced when rejecting directional interferences. There are several works that analyse the effects of the array geometry and the number of microphones for several types of fixed beamforming techniques, evaluating the intelligibility improvement introduced for hearing-impaired subjects. Some remarkable works are [18], [19], [20], and [22]. A common and affordable approach is the use of independent small endfire arrays, often integrated into behind-the-ear devices, with low microphone distances of around 1-2 cm [22]. The use of external larger arrays has been also proposed, for instance, with microphones placed in eyeglasses [23], but this solution is not comfortable for hearing aid users.

Adaptive beamforming requires higher computational capability and it is more sensitive to steering direction errors, but it has better performance rejecting interferences. However, the evaluation of the performance is highly influenced by the acoustic environment, which makes the measurement of the benefit obtained over fixed beamforming difficult. An example that uses the so-called minimum variance distortionless response (MVDR) filter [24] is found in [25], where the filter is used to implement a multichannel Wiener filter (MWF). One promising approach is the application of a generalized sidelobe canceller (GSC) structure [26]. Some GSC-based algorithms for hearing aids are [27], [28], [29], and [30].

4.3 Binaural speech enhancement

Hearing loss usually affects to both ears and the hearing-impaired person is forced to wear a hearing device in each side. Bilateral systems perform independent processing in the left and right hearing aids, which originates that the spatial cues are distorted, decreasing the localization ability of the user. A recent trend motivated by the availability of wireless data links between the right and left hearing aids is the design of binaural speech enhancement algorithms, which are a special case of multichannel algorithms where the speech is enhanced by combining the information from both ears.

Binaural systems work with dual-channel input-output signal, although more than one microphone could be placed in each device. The main advantage of binaural processing is the availability of spatial cues that can be used to separate sounds. The interaural time differences (ITD) and interaural level differences (ILD) are two of the most important spatial cues for the estimation of the source azimuth angle, which is the main priority for hearing aid users. However, these cues must be preserved in the binaural output in order to maintain the original spatial information. A simple example of binaural noise reduction is found in [31], where the ILD and ITD estimates are compared with a reference value for the frontal direction.

Binaural fixed beamformers have low computational complexity, but they only preserve the spatial cues of speech (i.e. the target signal). The work in [32] designs a dual-channel superdirective beamformer and obtains the binaural output signal by applying adaptive spectral weights to the beamformer input channels. The spectral weights are computed from the monaural output of the beamformer. The desired signal is passed unfiltered. The performance is further increased applying a MWF.

Some examples of binaural adaptive beamforming based on the GSC structure are [33], [34], which use two-microphone subband adaptive GSC-like structure to adaptively cancel out interfering sources. The binaural MWF produces the minimum mean square error (MMSE) estimate of the speech components in both ears, preserving the spatial cues of speech, although the noise cues may be distorted. The work in [35] introduces a binaural extension of a monaural multichannel noise reduction algorithm for hearing aids based on Wiener filtering. The algorithm preserves the ITD cues of the filtered speech. The work is extended in [36] to preserve both the ILD and ITD cues of speech and noise. Beamforming combined with CASA techniques allows preserving the binaural cues of speech and noise, but the use of time-frequency masking introduces some distortions. In [37], a binaural adaptive beamformer is trained to form a null in the front direction. A single time-frequency mask is then calculated comparing the responses from the front cardioid and the back cardioid. The binary masking algorithm is very simple, making feasible its implementation in a hearing aid. The system is designed using real measurements obtained from a KEMAR manikin. In [38] an adaptation of the MVDR beamformer is combined with monaural CASA attributes. The simultaneous and temporal grouping steps are performed with clustering and Kalman filtering. Finally, the analysis of the robustness of different binaural speech enhancement systems in hearing aids is carried out in [39], using objective perceptual quality measures.

One problem associated to the design of spatial filters in hearing aids is the next. The signals that arrive at each microphone of the array are affected by the well-known head shadow effect, which implies that the impinging signals not only differ in time differences, which depend on the relative position between the source and the microphone, but it also undergo amplitude distortions. This effect must be considered in the computation of the filter coefficients. The fact that this effect is highly dependent on a person causes the design of an array customized for a subject to need a correct measurement of such effect, which is not practical in real scenarios. The lack of information about the head of the hearing aid user causes directivity reduction and distortions. Many beamforming-based systems for hearing aids proposed in the literature neglect this problem. Some examples are [40], [22]. In other cases, the head shadow effect is considered in the design, assuming that it has been measured or modelled somehow, for instance in [32], [33], [39]. The work in [41] proposes different approaches to optimize the filter coefficients in case of unknown information about the head of the user. The methods aim at maximizing the average array gain while minimizing the average distortions, using a design dataset of head measurements. The authors also compare different microphone array configurations, including monaural and binaural arrays.

Binaural hearing aids require the exchange of information between the left and the right devices. Due to aesthetic reasons, the best solution is the use of a wireless link for data transmission, which notably increases the power consumption, one of the main limitations in these devices. This fact opens a new area of research: how to reduce the amount of information transmitted (bit-rate) without altering the performance of the enhancement system. One of the first related works is [42], which evaluates the gain provided by collaborating hearing aids as a function of the communication rate, using an information theoretic approach. In [43] the authors evaluate the decrement of noise reduction achieved by a binaural MWF when reducing the bandwidth of the transmission link. The work in [44] proposes two approaches to reduce data transmission. The first approach is to transmit only an estimation of the undesired signal at a determined bit rate, and the second approach is to transmit the complete received signal at the determined bit rate. The second schema transmits more information, but it requires higher transmission rate. Furthermore, the authors evaluate the transmission of only the low-frequency components. Unfortunately, the performance of the algorithms in [42], [44] is notably reduced when the transmission rate decreases (e.g. lower than 16 kbps). An additional problem associated to the use of beamforming techniques for wireless-communicated binaural hearing aids is the following. The output of the beamformer is obtained by combining a weighted version of the input channels from both devices. If one or several speech signals have been quantized and transmitted to the other device, the beamforming output is directly affected by quantization noise.

A recent trend motivated by the availability of wireless data links between the right and left hearing aids is the design of binaural speech enhancement algorithms.

The works in [45], [46] present a novel approach for the design of energy-efficient speech enhancement algorithms with low computational cost for wireless-communicated binaural hearing aids. Speech enhancement is achieved by means of source separation, combining time-frequency masking and supervised machine learning. In order to increment the energy efficiency of the wireless-communicated binaural hearing aids, it is proposed to quantize some of the parameters to be transmitted, avoiding the transmission of others found to be unnecessary. The number of quantization bits assigned to each parameter is computed by means of evolutionary computation techniques aiming at finding a balance between low bit rate and good speech enhancement.

5. Conclusions

The improvement of speech intelligibility in hearing aids is a very complex task due to the fact that hearing impaired people are more affected by speech distortions than normal listeners and that this type of devices presents important engineering limitations. This paper provides an overview of signal processing algorithms included in hearing aids as well as a discussion about which speech enhancement algorithms are suitable to be implemented in such devices.

Regarding single-channel speech enhancement, the only alternative to improve intelligibility is the design of algorithms that optimize objective measures correlated with speech intelligibility rather than with speech quality. In the case of multichannel speech enhancement, spatial filtering is a suitable solution, as long as the filter coefficients are calculated with low computational cost (e.g. fixed beamforming), which obtains good levels of intelligibility improvement. In the case of binaural systems, different binaural beamforming techniques have been proposed, but their performance is notably decreased by quantization noise and when the transmission rate is reduced in order to maximize the battery life. An alternative to increase the energy-efficiency in binaural speech enhancement systems has been proposed.

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