SIGNAL PROCESSING TECHNIQUES FOR A DSP HEARING AID

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ABSTRACT

The recent development of commercial hearing aids with digital signal processing (DSP) capabilities opens the door for an explosive growth in hearing aid sophistication. Current limitations on signal processing design due to hearing aid constraints, and the specialized needs of listeners with hearing loss, result in certain design specifications that any DSP hearing aid must meet. This paper will review the latest state-of-the-art with respect to current DSP hearing aids on the market, with an emphasis on the ReSound programmable DSP hearing aid, developed in a collaboration with GN Danavox and AudioLogic, Inc. Multiband compression, single-microphone processing, multi-microphone processing, and feedback reduction will be discussed.

1. INTRODUCTION

Until the past decade and a half, commercial hearing aid technology had developed little beyond simple linear amplification with peak clipping. Low power and small size requirements, combined with unclear scientific evidence for how to properly compensate for sensorineural hearing loss, kept the sophistication of hearing aid signal processing to a minimum. Today, the most sophisticated aids available cover a variety of nonlinear processing architectures for hearing loss compensation and some include simple noise reduction processing. They are small enough to fit completely inside the auditory canal, run on a 1.3V battery and draw less than 2mA for a battery life of over 100 hours.

While the introduction of hearing aids with DSP chips allow the application of sophisticated signal processing techniques to these devices, their capabilities are still limited by a processing power of a few MIPS due to the inherent constraints of a hearing aid platform. The following will attempt to detail the current state of the art in the hearing aid industry, with an emphasis on the technology being introduced with a programmable DSP system developed by a consortium formed by ReSound Corp., AudioLogic, Inc. and GN Danavox (the device hereafter being referred to as the RAD aid).

2. MULTIBAND COMPRESSION

Sensorineural hearing loss is primarily characterized by a loss of sensitivity to sounds that is more severe for low-level than high-level signals. In order for a hearing aid to restore loudness to normal levels, the gain supplied must vary with the level of the signal [10]. The gain must also adjust fast enough such that different amounts of gain is provided to successive phonemes of differing level. If the gain is reduced for a high-level vowel, for example, the gain must increase quickly so that a following consonant is audible to the hearing aid wearer and not forward-masked by the vowel. This level-dependent gain is referred to as fast-acting or syllabic compression and can be characterized by the compression ratio, the kneepoint level at which the gain transitions from linear to compression (usually set at a low level such that speech falls in the compression region), and the gain at the kneepoint.

Figure 1 shows the loudness curve for a hearing-impaired subject from which the gain necessary for normal loudness restoration can be calculated. The data was gathered using a loudness scaling procedure with half-octave bands of noise [1]. As indicated by the arrows, gains of 25-dB and 8-dB should be applied to stimulus levels of 50-dB SPL and 80-dB SPL, respectively.

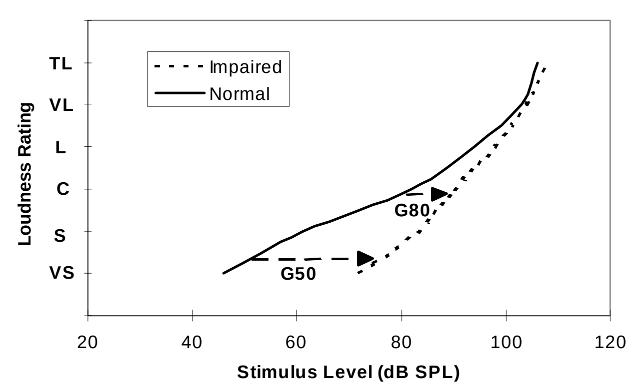

Figure 1. Typical loudness curves for both normal-hearing and hearing-impaired subjects. Loudness categories correspond to: Very Soft, Soft, Comfortable, Loud, Very Loud, Too Loud.

Because hearing loss varies with frequency, the gain and compression ratios must also vary with frequency. This is typically implemented by filtering the signal into different frequency bands and applying separate gain and compression to the signal in each band. One drawback to this system is illustrated in Figure 2, which shows the gain response of a three-band processor intended to provide equal gain at all frequencies, measured with both an 80-dB SPL tone-sweep and a broadband noise with 80-dB SPL in each band [2]. Identical I/O functions are used for each band so that equal gain should be applied across frequency for these signals. Because of the skirts of the filters used, the nonlinear processing near the crossover frequency between two bands provides more gain for narrowband signals than for broadband signals and violates the design requirement of a flat gain function. This effect is made worse as the number of bands increases since the number of crossover regions increases as well. Problems may also occur as the bandwidths of the filters approach the frequency spacing of a harmonic signal since the harmonics may fall near the transition regions between bands and the gain applied to each harmonic will depend on how many harmonics fall within any given band.

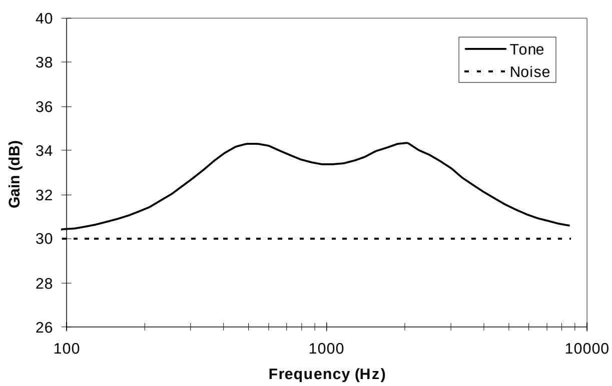

Figure 2. Gain response of a 3-band compressor, with the gain and compression ratio set at 30 dB and 3:1, respectively, for the stimulus level used. The response was measured with both a tone sweep and broadband noise.

This problem is solved by the sufficient overlapping of bands, such that the gain at any given frequency is determined by multiple bands [5]. The bandpass filtering can be treated as a sampling of the power spectrum and thus a minimum sampling rate is required to eliminate aliasing in the autocorrelation domain. By overlapping the bands, the crossover effect is not only reduced but problems resulting from, for example, the gain varying with the number of harmonics within a band is also reduced.

In the RAD aid, the multiband compressor that compensates for the listener's hearing loss is composed of 14 significantly-overlapping bands that minimize the problems discussed above. The gain and power calculations are performed in the frequency domain by using shift-and-sum FFT processing. This technique allows other FFT-based algorithms to be performed in parallel with the compressor and saves processing cycles. It is important that the overall delay of the hearing aid is on the order of milliseconds in order to preserve synchrony with visual cues and to keep the perception of the listener's own voice in synchrony with their speech production. This necessitates short frame lengths in the FFT structure.

3. SINGLE-MICROPHONE PROCESSING

Hearing-impaired listeners have a more difficult time understanding speech in noise than normal-hearing listeners and may require a speech-to-noise ratio (SNR) up to 10 dB higher than normal for the same recognition performance[6]. Many techniques exist for improving the SNR in a single-microphone system, but none have been shown to improve speech intelligibility for human listeners [4]. In fact, some algorithms increase the SNR while reducing intelligibility. As such, single-microphone hearing aid techniques for improving speech in noise are designed to improve subjective measures of benefit, referred to in the field as sound comfort.

Slow-acting automatic gain control (AGC) systems have been implemented in commercial hearing aids which reduce the level of sounds in frequency regions with high levels of noise and little to no levels of speech. This processing is performed both in regions with and without hearing loss. While many have suggested that this technique reduces upward spread of masking in the presence of high-level low-frequency noise, making high-frequency speech cues more audible, it has not been shown to have any benefit towards improved intelligibility [9]. Thus, improved comfort is the goal of this technique.

The signal envelopes in different frequency regions provide a useful statistic for estimating the presence of speech and the level of noise in each band. If the statistics of the envelope in a band is similar to that which is expected for speech in quiet, then no attenuation in that band is applied. As the envelope statistics approach that of noise, whether mechanical noise or speech babble, the gain in that band is reduced. This technique is embodied in the RAD aid using the envelopes in each of the 14 bands of the compressor. Using a large number of bands with this technique maximizes the ability to take advantage of spectral mismatches between the noise and speech. The different time constants of the syllabic compressor and noise reduction system minimize the interaction of the two AGCs such that they do not counteract each other. This noise reduction technique also addresses the common hearing aid problem of audible microphone noise in quiet environments when the compressor gain is maximum.

Speech understanding in noise is exacerbated by auditory systems with damaged by outer hair cells since the resulting loss of lateral suppression and broader auditory filters reduces perceived spectral contrast (e.g., [8]). Place cues that rely on spectral envelope shape become less distinguishable because of the smoothed spectrum in the perceptual domain. Attempts to sharpen the spectrum with single-microphone processing techniques, however, have had little success [7].

4. MULTI-MICROPHONE PROCESSING

Array processing can be used with hearing aids in order to provide direction-dependent gain. The design typically assumes that the target signal is in front of the hearing aid wearer, and the directional system can be designed to maximize the attenuation from all other directions. Because of cosmetic constraints, arrays with several microphones that span a useful distance and are separate from the body of the hearing aid have not been feasible as widely-acceptable commercial products. Two omni-directional microphones mounted on a hearing aid, however, can provide a significant amount of directionality that results in improved speech understanding in noisy environments.

. The size of behind-the-ear cases on which the microphones must be mounted limits the maximum separation between microphones to approximately 1.5 cm; in-the-ear cases limit microphone separation to approximately 1 cm. The small distance between microphones prevents the application of standard beamforming--passing the signal unchanged from the direction of the beam by adding the microphone signals together--since useful directionality only occurs above 6 kHz at these separation distances. Directionality is achieved by subtracting the back microphone from the front one, steering the direction of the response null by adding a delay to the back microphone. With this processing, the on-axis (frontal) frequency response has a highpass characteristic, rolling off at 6-dB/octave, which provides a tinny characteristic to the sound and may make low-frequency signals inaudible. The RAD aid, which has two microphones, achieves the back-microphone delay with an all-pass filter and uses digital filtering to compensate for the reduced low-frequencies.

Because of the location of the microphones, a significant amount of directionality is provided simply from the headshadow effect. Figure 3 shows the polar patterns of a two-microphone directional system and a single omni-directional microphone on a behind-the-ear aid worn on a mannequin. The improvement in speech intelligibility is approximately 5 dB for a speech signal in front of the listener and diffuse background noise, meaning that signal-to-noise ratio is 5-dB less than that needed for the same intelligibility with an omni-directional microphone. To take into account the importance of different frequency regions to speech intelligibility, the articulation index can be incorporated with a directivity index to obtain an overall measure of the estimated benefit from the directionality of an aid with respect to understanding speech in noise [3].

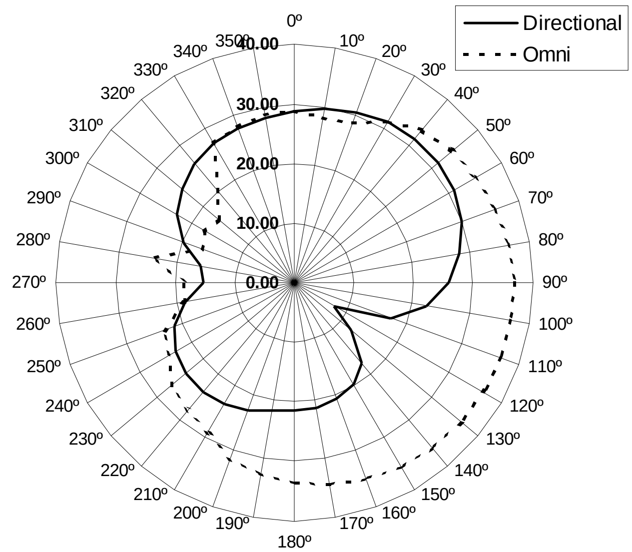

Figure 3. Polar plots for a behind-the-ear hearing aid with an omni-microphone and a two-microphone directional array, measured with the aid on a mannequin.

Directionality becomes less effective when there are phase and amplitude mismatches between the two microphones on a directional aid, and effect of the mismatch is more significant as the distance between microphones decreases. Thus, the responses of the two microphones in a directional aid must be as closely matched as possible. The RAD aid has two A/D converters such that the output of each microphone is sampled and digitized, the rear microphone signal is delayed, and the rear signal is subtracted from the front signal. By subtracting the two microphones digitally instead of electrically before the A/Ds, digital filters can be used to match the two microphones such that their amplitude and phase responses are as close as possible, as is done in the RAD aid. This also allows techniques for tracking any changes in their responses over time and adaptively compensating for these changes.

Noise cancellation can also be attempted by substituting the rear omni-directional microphone with a directional microphone that is directed behind the listener. An adaptive filter is applied to the signal from the directional microphone to minimize the error signal obtained when subtracting the filtered rear signal from the front signal. Laboratory experiments have shown that maximum benefit is achieved when the adaptive filter is only updated when no speech (target) signal is detected from the on-axis direction [11]. This ensures that that the directional microphone is only measuring noise and the target signal is not canceled.

5. FEEDBACK

Feedback is a significant problem for many hearing aid wearers. Leaks around an ear mold and through the vent in the ear mold can cause feedback, which may be made worse by the presence of an object brought near the aid such as a phone receiver, hat or hand. Because vents are often used to prevent an occlusion effect of the listener's own speech sounding boomy, the amount of gain that an aid provides may be limited to below that necessary for proper hearing loss compensation due to the feedback path through the vent or leaky earmold.

One solution that many manufacturers have used is to simply reduce the hearing aid gain in a broad frequency band that includes the feedback signal. Notch filters have also been used to solve this problem. Both of these solutions have the drawback that they reduced the gain in the problem frequency region, making any speech cues that may exist in that reduced frequency region inaudible to the hearing aid wearer. The RAD aid uses digital processing for adaptive feedback cancellation whereby the feedback path between the receiver and microphone is estimated and a filtered version of the signal to the receiver is subtracted from the output of the microphone. A brief, internally-generated noise burst is used to obtain an initial estimate of the feedback path and the parameters of an adaptive filter used as a part of the feedback cancellation is calculated using correlation techniques. During the normal functioning of the aid, the receiver input is passed through the digital filter and then subtracted from the microphone output to cancel the feedback signal. The filter is also adapted during the normal use of the aid, without any internally-generated noise probe, in order to track both sudden and slowly-varying changes to the feedback path such as the placement of a phone near the ear or a loosening ear mold, respectively.

6. CONCLUSIONS

DSP chips allow for significant improvements to hearing aid processing, but care must be taken that the specialized needs of the hearing-impaired users are accounted for. Specifically, any processing such as noise reduction or directionality must not counter the effect of the hearing-loss compensation processing. While improved speech intelligibility is one of the primary goals of a hearing aid, algorithms which improve subjective characteristics such as sound quality are also goals for signal processing.

7. REFERENCES

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