ON SUBBAND ADAPTIVE MODELING OF COMPRESSION HEARING AIDS

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ABSTRACT

The current generation of digital hearing aids perform amplitude compression in multiple channels. Resulting non-linear processing has the potential to create distortion components which reduce their effectiveness and user satisfaction. The performance of these devices is typically evaluated using subjective test procedures. While these approaches are preferred, they are time consuming. Objective electroacoustic measurements of speech quality are attractive but require effective modeling of these devices. Subband adaptive modeling architectures have been found to facilitate these measurements. In this paper, it is shown that optimal modeling occurs only when the number of channels in the subband adaptive model matches the number of hearing aid channels. A technique based on combined sinusoidal and broadband noise excitation is then exploited to identify the number of channels in the hearing aid to be used in the optimal subband adaptive model.

1. INTRODUCTION

Accurate electroacoustic characterization of hearing aids is critical in patient fitting, assessment, and design of these devices. With ever increasing complexity and use of advanced digital signal processing techniques, hearing aids must be evaluated using real world stimuli like speech and music. The present collection of electroacoustic test measures, more specifically ANSI S3.22 [1], employ pure-tone and/or broadband excitation signals not fully representative of critical natural speech characteristics. Hearing aid performance elicited under these natural conditions is important for validating the basic operation, manufacturing specifications, and overall device quality.

A majority of current generation digital hearing aids perform amplitude compression in multiple channels to facilitate effective mapping of the wide dynamic range acoustic signals to the reduced audibility range of hearing impaired individuals [2]. Resulting non-linear processing has the potential to create distortion components which, when combined with the inherent device noise, reduce the effectiveness and overall user satisfaction. Typically, the performance of hearing aids is evaluated through subjective speech intelligibility and quality tests [3]. While the subjective tests are preferred for their face-validity, they are time consuming and expensive. Objective electroacoustic measurements of speech quality are therefore attractive for device evaluation and characterization.

Objective measures of speech quality that have been successfully used in quantifying speech coder performance (such as the Perceptual Evaluation of Speech Quality (PESQ) measure [4] and the Measuring Normalizing Blocks (MNBs) [5]) can be adapted for evaluating hearing aid speech quality. In both PESQ and MNB techniques, the distance between the input and output is computed using a set of perceptual criteria. Since hearing aids by design shape the frequency spectrum of the input signal (in relation to the hearing loss profile), PESQ and MNB cannot be computed directly on the hearing input and output signals. An adaptive model is instead first used to estimate the time-varying frequency response of the hearing aid, and the distance parameters are computed between the output of the adaptive model and the hearing aid output [5]. Results show that with this adaptive model, the objective measures correlate well with perceptual judgments of hearing aid sound quality, both by individuals with normal hearing and by individuals with hearing impairment [5].

Earlier work [6] has shown that subband adaptive models outperform full band adaptive models in accurately modeling the dynamic behavior of multi-channel compression hearing aids. However, the modeling performance of the subband adaptive structures was found to be suboptimal when there is a difference between the number of channels in the adaptive model and the hearing aid. In this paper, channel-offset modeling is used as a precursor to validate the impact on subband adaptive modeling when the number of model channels is different from the number of hearing aid channels. In addition, a technique based on biased sinusoidal plus broadband noise excitation, originally developed by Kates [7], is extended to identify the number of channels in a given digital hearing aid which is used in developing the optimal subband adaptive model.

2. ADAPTIVE SUBBAND MODELING

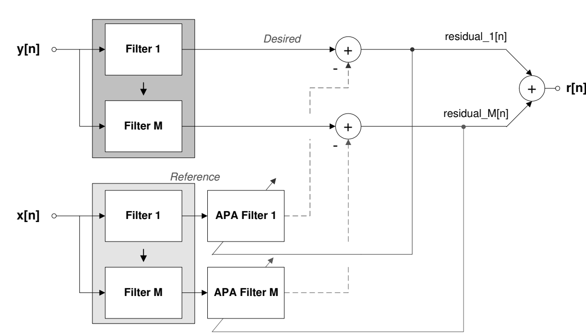

Fig. 1. Subband Adaptive Architecture

Due to the presence of multiple channels, a full band adaptive model will not adequately characterize the performance of a multichannel amplitude compression hearing aid. Figure 1 illustrates the subband architecture considered in this paper.

The hearing aid output and input sequences, *y[n]* and *x[n]*, are filtered using uniform, M-channel analysis filter banks [8]. The resulting subband output sets, $y_0[n]$, ..., $y_{M-1}[n]$ and $x_0[n]$, ..., $x_{M-1}[n]$, form the desired and reference sequences required by the affine adaptive filter blocks (APA Filter 1, ... , APA Filter M), respectively.

Each adaptive filter was implemented as a Finite-Impulse Response (FIR) filter whose coefficients were updated using the complex affine projection algorithm. Let $\mathbf{W}_i(k) = [W_i^0(k), W_i^1(k), ..., W_i^{N-1}(k)\ ]^T$ be the N^{th} order vector representing the adaptive filter in the i^{th} subband. The coefficient update algorithm can be summarized as:

$$\mathbf{e}_i(\mathrm{k}) = \mathbf{y}_i(\mathrm{k}) - \mathbf{X}_i^H(\mathrm{k})\mathbf{W}_i^*(\mathrm{k})$$

$$\Phi_i(\mathrm{k}) = [\mathbf{X}_i^H(\mathrm{k})\mathbf{W}_i(\mathrm{k}) + \delta\mathbf{I}]^{-1}$$

$$\mathbf{W}_i(\mathrm{k+1}) = \mathbf{W}_i(\mathrm{k}) + \mu\mathbf{X}_i(\mathrm{k})\Phi_i(\mathrm{k})\mathbf{e}_i^*(\mathrm{k})$$

where δ is the power-bias term and μ is the adaptation constant. The projection order is kept low due to the non-stationary characteristics of the speech sequence.

It is assumed that the residuals account for the non-linear effects of distortion and inherent noise produced by the hearing aid.

3. METHOD

3.1. Simulated Hearing Aid

In order to investigate the effect of channel mismatch on modeling performance, a multichannel compression hearing aid was simulated, as shown in Figure 2.

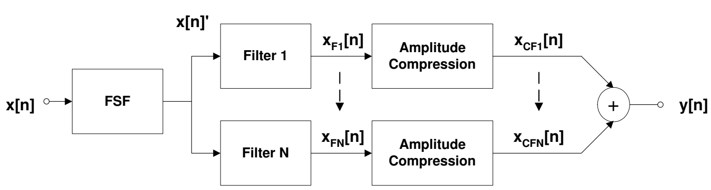

Fig. 2. Simulated Hearing Aid

The speech excitation sequence, x[n], is filtered with a minimum mean-squared error designed finite-impulse response frequency shaping filter, FSF. This filter provides gain compensation for a typical moderate-to-severe hearing loss audiogram. Gain compensation can also be realized by corresponding gain changes in each of the subband channels. The resulting full band sequence feeds a uniform, N-channel filter bank. This filter bank is designed using a Kaiser-window based cosine-modulated technique [8]. Each resulting subband sequence is processed using discrete amplitude compression modules [9]. Compression parameters for each channel (including compression threshold, compression ratio, attack time, and release time) can be independently set and are meant for illustrative purposes only. The resulting full band, amplitude compressed sequence, y[n], is the sum the of individual channel sequences, x_{CF1}[n], ... , x_{CFN}[n].

3.2. Channel-Offset Modeling

Channel-offset modeling is used to validate the impact on subband adaptive modeling when the number of channels is different from the number of hearing aid channels. It is based on a maximized effectual signal-to-distortion ratio (SDR).

An 8 kHz re-sampled HINT sentence (”The front yard was pretty”) is applied to the simulated hearing aid. The processed and original sequences become the y[n] and x[n] inputs to the channel-offset processing structure. The number of analysis channels, M, is incremented from one to a final value larger than the number of channels in the hearing aid. For each M-channel set, an SDR value is determined. A plot of the number of filter bank channels along the abscissa with corresponding SDR values along the ordinate allows visual indication of the impact of a channel-offset.

3.3. Bias-Tone with Broadband Excitation

As noted by Kates, hearing aid processing can be ascertained by observing how a frequency-swept bias-tone modifies the frequency response of a hearing aid to broadband noise [7].

The swept-tone functions to bias the hearing aid into a nonlinear processing state while the broadband noise is used to extract the respective frequency magnitude response. Due to the linear nature of the underlying process, hearing aids using linear processing will not record gain changes.

A test signal is constructed by adding a broadband noise sequence at a -30 dB power level to a swept sinusoid at a -10 dB power level. Two differences from Kates original work include the use of unshaped broadband noise and extension of the final swept frequency to 9500 Hz. Responses of the simulated hearing aid to this test signal, the broadband noise alone, and the swept-bias tone alone are recorded. A two-channel adaptive noise cancellation system, shown in Figure 3, processes these three sequences.

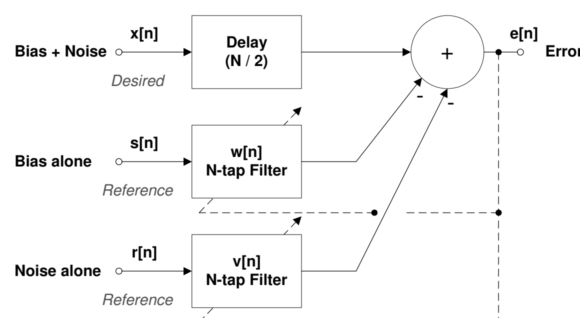

Fig. 3. Two Channel Adaptive System

The bias-and-noise response, x[n], is applied to an N/2 delay block, forming the desired sequence. The bias-alone response, s[n], is applied to w[n], an N-tap adaptive filter. The noise-alone response, r[n], is applied to v[n], also an N-tap adaptive filter. Adaptive filter w[n] removes the bias-tone component from the desired response and adaptive filter v[n] removes the noise component. The tap-weights of v[n] are extracted at periodic intervals and used to determine the frequency magnitude response of the hearing aid. A gain-threshold is used to transform these frequency magnitude responses into corresponding two-level representations and are plotted as a function of the sweep frequency. The resulting pattern is correlated against ideal reference patterns.

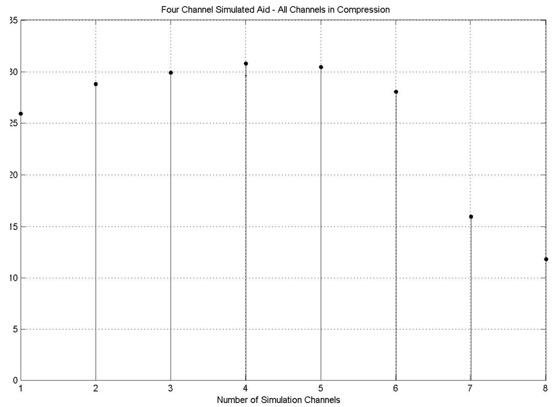

Fig. 4. Four Channel Model

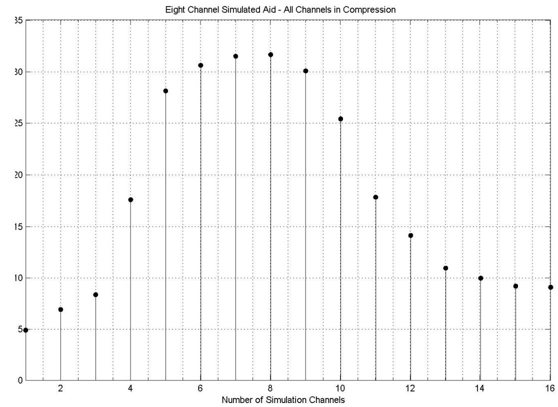

Fig. 5. Eight Channel Model

4. SIMULATION RESULTS

Channel-offset modeling and the bias-tone with broadband excitation were considered for four and eight channel hearing aid models.

4.1. Channel-Offset Modeling

Table 1 shows the compression parameters for each channel in the simulated four channel compression hearing aid.

Compression Ratio	4:1
Compression Threshold	-10 dB below peak
Attack Time	5 msec.
Release Time	10 msec.

Table 1. Channel Compression Parameters

Active detection of the peak sample value in the filtered subband sequences is used to dynamically set the compression threshold 10 dB lower to ensure that compression is active. Figure 4 illustrates the effectual SDR value as a function of the number of simulation channels, M. As can be seen, the largest SDR value occurs when the number of channels in the adaptive model is equal to 4.

The four-channel compression parameter set is used for the eight-channel model as well. Figure 5 illustrates the eight-channel results, which again shows that the largest SDR value occurs when the number of channels in the adaptive model is 8. Similar results were obtained when the number of compression channels are increased to 16 and 32.

Thus the maximum effectual SDR occurs when the number of channels matches the number of simulated hearing aid channels. The SDR values on either side of the optimal number of channels are within 1 to 2 dB of the optimal value. Further increases in offset result in lower SDR values, indicating that under- or over-modeling the number of compression channels in the hearing aid will result in degraded modeling performance.

4.2. Bias Tone with Broadband Excitation

Table 2 shows the compression parameters for each channel in the simulated four channel compression hearing aid used in this set of experiments.

Compression Ratio	4:1
Compression Threshold	-30 dB
Attack Time	5 msec.
Release Time	10 msec.

Table 2. Channel Compression Parameters

The power level of the swept-bias tone is set at -7 dB and broadband noise power level is set 30 dB lower to -37 dB. These values are selected to accommodate the amplitude compression working threshold range from 0 dB to 96 dB. As noted by Kates, the number of coefficients used by the two-channel adaptive filter system represents a compromise between the rate of convergence and frequency resolution. The number of coefficients was set to 32 to accommodate more than two channels.

Figure 6 illustrates the identification pattern with all four channels in compression. It can be seen that four channels are present with their respective bandwidths being approximately uniform. Interpretation of this result and its classification is facilitated using an idealized system identification pattern. Figure 7 illustrates the ideal pattern for a four-channel system using amplitude compression.

Compression parameters identical to the four-channel model are used in the eight-channel model. Figure 8 illustrates the eight-channel results, while the ideal pattern for an eight channel compression hearing aid with all channels in active compression is shown in Figure 9. From these graphs, it is evident that the combined bias tone plus broadband excitation correctly identifies the number of active compression channels. Thus this technique can serve as a front-end to the subband adaptive modeling paradigm for determining the number and bandwidths of the subbands.

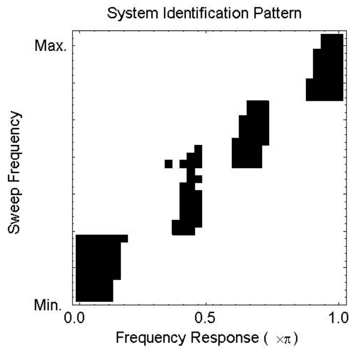

Fig. 6. Identification Pattern with All 4 Channels in Compression

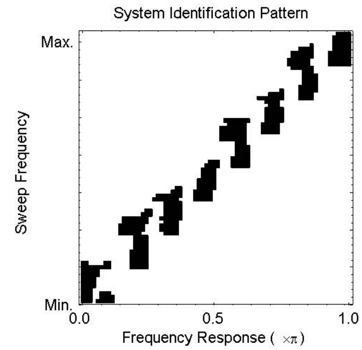

Fig. 8. Identification Pattern with All 8 Channels in Compression

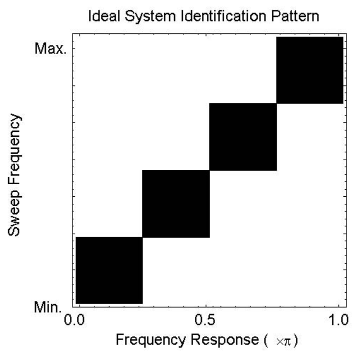

Fig. 7. Ideal Four Channel Identification Pattern

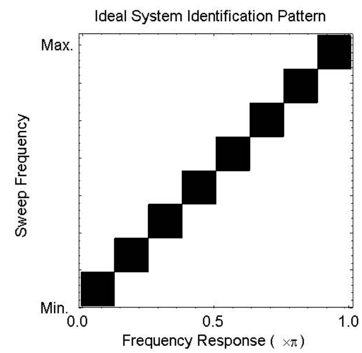

Fig. 9. Ideal Eight Channel Identification Pattern

5. CONCLUSIONS

Distortion in hearing aids reduces their effectiveness and user satisfaction. While subjective tests of speech intelligibility and quality are preferred, they are time-consuming and expensive. Objective measures of speech quality are therefore attractive. Because these objective measures rely upon how accurate an adaptive model of a hearing aid is, knowledge of the exact number of channels is critical.

The results of channel-offset modeling presented in this paper validated the need to ensure that the number of channels in the subband adaptive model is equal to the number of hearing aid channels. The combined sinusoidal plus broadband noise excitation test developed by Kates [7] was successfully extended to predict the number of processing bands and associated bandwidth for simulated four and eight channel hearing aids. This system can therefore be used as a front-end for developing the optimal subband adaptive model for a given multichannel hearing aid.

6. ACKNOWLEDGEMENTS

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7. REFERENCES

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