OBJECTIVE MEASURES FOR REAL-TIME EVALUATION OF ADAPTIVE FEEDBACK CANCELLATION ALGORITHMS IN HEARING AIDS

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ABSTRACT

This paper discusses objective measures for real-time evaluation of feedback cancellation algorithms in hearing aids in the presence of non-white input signals. Based on these measures, the maximum stable gain, the sound quality and the time required to recover from instability are determined for several real-time frequency-domain adaptive feedback cancellation algorithms, including adaptive feedback cancellation with bandlimited adaptation and prediction-error method based adaptive feedback cancellation (PemAFC) with autoregressive modelling of the desired signal. Of all the evaluated methods, the PemAFC with adaptive desired signal model achieves the best feedback suppression for a tonal music signal and the shortest time to recover from instability.

1. INTRODUCTION

Acoustic feedback limits the maximum gain that can be applied in a hearing aid without making it unstable. A solution for the acoustic feedback problem is the use of feedback cancellation [7, 4]. Here, a model of the acoustic feedback path is identified, which is then used to estimate and remove the (unwanted) feedback signal from the microphone signal. Although feedback cancellation algorithms have become common in digital hearing aids, there is no standardized objective procedure available for evaluating them.

An often used performance measure to evaluate feedback cancellation algorithms is the maximum stable gain, i.e., the maximum gain that can be applied without rendering the system unstable [7, 4, 2, 8, 5]. To determine the MSG, the hearing aid gain is gradually increased until instability occurs. However, an objective criterion for detecting instability is lacking. Recently, Freed et al. [2] and Shin et al. [11] proposed objective criteria for instability based on the power concentration ratio and the hearing aid transfer function variation, respectively. However, these criteria only hold for a white noise input signal to the hearing aid. Adaptive feedback cancellation algorithms in particular encounter problems when the input signal is non-white e.g., a music signal [12, 14, 6]. Hence, the maximum stable gain determined for a white input signal is often an overestimation of the maximum stable gain that can be used in real-life scenarios. In addition to a high maximum stable gain, the feedback cancellation algorithm should preserve a good sound quality and quickly adjust to feedback path changes so that when the system becomes unstable, the duration of instability is minimal.

In this paper, objective criteria are investigated for detecting instability, the presence of oscillations or distortion in the case of spectrally coloured input signals. These measures are applied to evaluate the performance of several real-time frequency-domain adaptive feedback cancellation algorithms, including adaptive feedback cancellation with bandlimited adaptation and prediction-error method based adaptive feedback cancellation (PemAFC) with autoregressive modelling of the input signal. The maximum stable gain as well as the sound quality of the algorithms is determined for a speech signal and a tonal music signal. In addition, the time needed by the algorithms to recover from an unstable hearing aid condition is measured. The PemAFC with adaptive desired signal model achieves the best feedback suppression for tonal signals and the shortest time to recover from instability of the evaluated methods.

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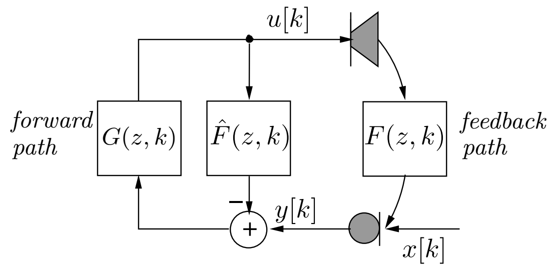

Figure 1: Adaptive feedback cancellation.

Notation

$P_u(f,k)$ is the short-term power spectral density (PSD) of the signal $u[k]$. $F(z,k)$ represents the transfer function of a discrete-time filter $\mathbf{f}[k]$ with time-varying coefficients $\mathbf{f}[k] = [\ f_0[k] \quad f_1[k] \quad \cdots \quad f_{L_F-1}\ [k]]^T$ and filter length L_F. Filtering $u[k]$ with $F(z,k)$ is denoted as $F(z,k)U(z)$ where $U(z)$ equals the z-transform of $u[k]$.

2. ADAPTIVE FEEDBACK CANCELLATION

2.1 Concept

Figure 1 depicts the basic set-up of an adaptive feedback canceller. The so-called *forward path* $G(z,k)$ represents the regular signal processing path of the hearing aid (typically including a frequency specific gain, compression and/or noise reduction). The *feedback path* between the receiver (loudspeaker) and the microphone is $F(z,k)$. The receiver and microphone signal are $u[k]$ and $y[k]$, respectively. The desired signal (i.e., without acoustic feedback) is $x[k]$.

The adaptive feedback canceller produces an estimate $\hat{F}(z,k)U(z)$ of the feedback signal $F(z,k)U(z)$ and subtracts this estimate from the microphone signal $Y(z)$. As a result, the transfer function $C(z,k)$ from $x[k]$ to $u[k]$ equals

$$C(z,k) = \frac{G(z,k)}{1 - G(z,k)\left(F(z,k) - \hat{F}(z,k)\right)}. \tag{1}$$

Ideally, $\hat{F}(z,k) = F(z,k)$ such that $C(z,k)$ equals the desired hearing aid response $G(z,k)$.

2.2 Algorithms

The presence of the forward path $G(z,k)$ introduces a specific signal correlation between $x[k]$ and $u[k]$ when the desired signal $x[k]$ is spectrally colored. As a result, a standard continuous adaptation feedback canceller (CAF) fails to provide a reliable feedback path estimate $\hat{F}(z,k)$ [13]. Several approaches for improving the estimation accuracy of the CAF have been proposed in the literature, including adaptive feedback cancellation with bandlimited adaptation (BL-CAF) [1] and prediction-error method based adaptive feedback cancellation (PemAFC) [6, 14]. The BL-CAF restricts feedback cancellation to the frequency band that encompasses the unstable frequency (i.e., typically the higher frequencies) through bandlimited adaptation. The PemAFC reduces the correlation between $x[k]$ and $u[k]$ through prewhitening with an estimate of the inverse desired signal model.

In this paper, the following real-time partitioned-block frequency domain implementations (64-point FFT, block size of 32) are evaluated:

- CAF [13]
- BL-CAF using a high-pass filter with a cut-off frequency of 1 kHz [1]
- PemAFC using a fixed desired signal model (PemAFC-f). A first-order all-pole model $\frac{1}{1-0.9z^{-1}}$ is used. [6]
- PemAFC using a 20th-order adaptive all-pole desired signal model (PemAFC-a) [14]
- PemAFC-a with shadow filtering approach for improved tracking performance (PemAFC-as) [14]. The PemAFC-as combines a slowly adapting feedback canceller with a second fast canceller for improved tracking performance.

A detailed description of the algorithms can be found in [14].

In all algorithms, the step-size is normalized per frequency-bin according to the sum of the input power and the error power in the bin [1, 3]. The step-size constant $\mu = 0.01$ in CAF, BL-CAF, PemAFC-f and PemAFC-a. In the PemAFC-as, the step-size constant of the slowly adapting filter equals $0.01/\sqrt{10}$, the step-size constant of the fast adapting filter equals 0.1. The filter length of the feedback path estimate equals 64 taps.

To set a reference for the optimum feedback cancellation performance with a 64-taps FIR filter, the acoustic feedback path was also identified in quiet by sending white noise through the hearing aid receiver and disconnecting the forward path $G(z,k)$. This identified feedback path was then used to cancel the feedback. This static feedback canceller is referred to as 'Optimum'.

3. SET-UP

The feedback evaluation was performed in a soundproof booth. A Siemens Acuris behind-the-ear hearing aid was mounted on a Cortex II artificial head using a Phonak Fit-and-Go open fitting. The first microphone and the receiver of the left hearing aid were connected to a PC through an RME Hammerfall DSP Multiface II sound card. The processing was done at a sampling frequency $f_s = 16$ kHz. The forward path $G(z,k)$ consists of a flat frequency-gain response. Peak clipping was applied to $u[k]$ to keep the signals within the range of the DAC of the sound card. The total round-trip delay from the electrical microphone signal to the receiver signal equals 6.3 ms. This delay was modelled by adding a delay in the feedback cancellation path.

Signals were presented through a loudspeaker, positioned at 1 meter in front of the center of the head. Three test signals were used in the experiments: stationary speech-weighted noise, 20 seconds of real speech from the HINT database [10] and a 20 seconds opera fragment of 'Der Hölle Rache' from 'Die Zauberflöte of W.A. Mozart. The signal level equals 60 dB SPL, as measured near the entrance of the ear canal of the head. The level should be low enough to be able to determine the maximum stable gain: the maximum stable gain cannot be determined when the maximum output level of the sound card or the saturation level of the receiver is reached. On the other hand, it should be high enough such that it is well above the environmental and internal noise in the test room.

4. MEASURES FOR DETECTING INSTABILITY, OSCILLATIONS AND SIGNAL DISTORTION

This section defines objective measures for detecting instability, oscillations and signal distortion in the hearing aid output, which are applicable to spectrally colored input signals. To take into account the spectral coloration of the input signal, the measures compare the actual receiver signal $u[k]$ with the receiver signal that would be obtained in the absence of acoustic feedback (or under ideal feedback cancellation). The latter is referred to as reference signal $r[k]$ and is obtained as the hearing aid microphone recording in the absence of feedback (i.e., $F(z,k) = 0$), amplified with the same gain function $G(z,k)$ as when the feedback canceller was running. The performance measures should not be affected by differences in the environment and internal noise $n[k]$ between the recordings of $u[k]$ and $r[k]$.

4.1 Detection of instability

4.1.1 *Receiver-to-reference signal energy ratio [dB]*

A simple method to detect instability is to track the short-term energy ratio $E(k)$ of the receiver-to-reference signal

$$E(k) = 10\log_{10}\frac{\sum_{i=k-L/2}^{k+L/2} u^2[i]}{\sum_{i=k-L/2}^{k+L/2} r^2[i]}. \tag{2}$$

$L+1$ is the window length used in the energy computation (e.g., using half-overlapping frames of 0.5 sec). Instability is said to occur if the energy ratio exceeds a certain threshold, e.g., 10 dB. The energy ratio $E(k)$ detects an increase in the output signal level caused by feedback. However, it does not give any information about the amount of residual feedback.

4.1.2 *Feedback-to-signal energy ratio [dB]*

To quantify the amount of feedback, the short-term feedback-to-signal energy ratio FSR(k) can be computed as:

$$FSR(k) = 10\log_{10}\frac{\sum_{i=k-L/2}^{k+L/2} (u[i]-r[i])^2}{\sum_{i=k-L/2}^{k+L/2} r^2[i]}. \tag{3}$$

To reduce the effect of differences in the environment and internal noise component $n[k]$ of $u[k]$ and $r[k]$, $FSR(k)$ is only computed when the short-term energy of the reference signal $r[k]$ exceeds the environment noise $n[k]$ by at least 10 dB.

4.2 Detection of oscillations

Even before instability occurs, oscillations can already be perceived. Freed et al. [2] and Shin et al. [11] defined objective criteria to detect oscillations, referred to as the power concentration ratio (PCR) and the hearing aid transfer function variation criterion (TVC), respectively. The proposed criteria are only defined for a white noise input signal. In addition, the PCR depends on the hearing aid frequency response. Below, these measures are modified so that they can be applied to spectrally coloured input signals and are robust to spectral peaks in the hearing aid frequency response. The measures are computed using half-overlapping frames of 0.5 sec.

4.2.1 *Transfer function variation criterion (TVC)*

In [11], the hearing aid transfer function is measured for increasing gain values by sending white noise through an external loudspeaker. The hearing aid transfer function in the initial stable stage acts as the reference hearing aid transfer function. This reference transfer function is multiplied by the gain difference between the increased gain and the stable gain. The difference in dB between the amplitude characteristic of the hearing aid transfer function and the gain compensated reference hearing aid transfer function is computed and is called transfer function variation function (TVF). The transfer function variation criterion (TVC) is then defined as

$$\mathrm{TVC} = \max_f(|TVF(f)|). \tag{4}$$

Instability is assumed if TVC $\geq$ 10 dB.

For spectrally coloured input signals, an estimate of TVF(f,k) is obtained as

$$\mathrm{TVF}(f,k) = 10\log_{10}\left(\frac{P_u(f,k)}{P_r(f,k)}\right), \tag{5}$$

which may then be used in (4).

To avoid erroneous results caused by differences in the environment noise component $n[k]$ of $u[k]$ and $r[k]$, the PSDs $P_u(f,k)$ and $P_r(f,k)$ are constrained:

$$\begin{aligned} P_u(f,k) &= \max(P_u(f,k), \alpha P_n(f)), \\ P_r(f,k) &= \max(P_r(f,k), \alpha P_n(f)), \end{aligned} \tag{6}$$

where $P_n(f)$ is the long-term PSD of $n[k]$ and $\alpha > 1$.

The TVC detects the largest peak or dip in the transfer function variation function. However, it does not take into account the signal power $P_u(f,k)$ at the detected oscillation frequency. For non-white input signals, not all frequencies are equally excited. As a result, the detected oscillation frequency may be masked by non-critical frequencies with more power.

4.2.2 *Power concentration ratio (PCR)*

In [2], the power concentration ratio, i.e., the degree to which a large amount of power is concentrated at a small number of frequencies in the hearing aid output, is introduced for detecting oscillations. The hearing aid output is recorded while presenting a white noise signal to the hearing aid. The PCR is defined as the ratio of the power in the five frequency bins with the highest power to the total power. Oscillation is defined based on a threshold PCR value (e.g., 0.5).

The PCR assumes that any strong peaks in the output PSD are due to acoustic feedback. However, spectral peaks can already be present in the stable hearing aid system (e.g., resonances in the hearing aid response). In addition, the PSD of a non-white signal such as speech and music may have strong spectral peaks and hence, a large PCR, making the measure inappropriate for colored input signals. Therefore, a modified measure based on the PCR is proposed:

1. First, the oscillation frequencies f_c are detected as the frequencies where the transfer function variation $TVF(f)$ (cf. 5) equals or exceeds 6 dB. This avoids that resonances in the hearing aid response are falsely identified as oscillation frequencies. The fraction of the total power $P_u(f)$ of $u[k]$ that is located at the five (or less) strongest oscillation frequencies is computed and referred to as PCR$_u(k)$.
2. To reduce the PCR dependency on the input signal PSD and the hearing aid response, the fraction of the total power $P_r(f)$ of the reference signal $r[k]$ that is located at the detected oscillation frequencies f_c is also computed and is referred to as PCR$_r$.
3. The difference ΔPCR

$$\Delta\mathrm{PCR} = \mathrm{PCR}_u - \mathrm{PCR}_r \tag{7}$$

is then used as a measure for the presence of oscillations.

4.3 Frequency-weighted log-spectral signal distortion

To assess the distortion of the receiver signal $u[k]$, the frequency-weighted log-spectral signal distortion SD(k) is defined as

$$\mathrm{SD}(k) = \sqrt{\int_{f_l}^{f_u} w_{ERB}(f)\left(10\log_{10}\frac{P_u(f,k)}{P_r(f,k)}\right)^2 df}. \tag{8}$$

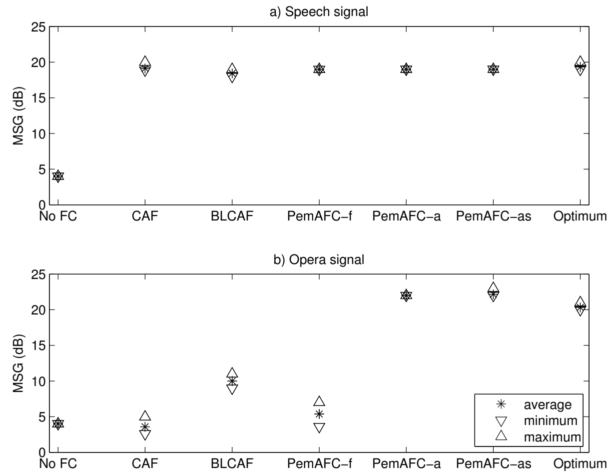

Figure 2: MSG of the feedback cancellation algorithms compared to no feedback cancellation ('No FC') and the optimum achievable MSG ('Optimum'). a) Speech signal; b) Opera signal.

The short-term PSD $P_u(f,k)$ and $P_r(f,k)$ are computed using frames of 16 ms. They are constrained as in (6) so that only the frequencies bins with an energy above the environment and internal noise level are taken into account. The frequency-weighting factor $w_{ERB}(f)$ gives equal weight to each auditory critical band between $f_l = 300$Hz and $f_u = 6500$Hz, defined by the equivalent rectangular bandwidth (ERB) of auditory filters [9]. Both the mean and maximum distortion SD(k) in the signal $u[k]$ will be used to assess the distortion. Signal segments are only included in the computation of the mean and the maximum if the energy of the reference signal $r[k]$ is at least 10 dB above the environment and internal noise level.

5. EVALUATION PROCEDURE AND RESULTS

Based on the measures of Section 4, the maximum stable gain, the sound quality and the time to recover from instability of the feedback cancellation algorithms is assessed.

5.1 Maximum stable gain (MSG)

The maximum stable gain (MSG) of the feedback cancellation algorithms was measured by increasing the forward path gain $|G(z,k)|$ in steps of 1 dB per signal length until instability occurred. The MSG of the algorithms is compared with the MSG without feedback cancellation ('No FC') and the optimum MSG based on a fixed identified 64-taps feedback path ('Optimum'). The latter reflects the optimum MSG that can be achieved by the algorithms. For each gain setting, the maximum short-term energy ratio $\max_k\{E(k)\}$ over the whole signal segment was computed. The maximum gain setting for which $\max_k\{E(k)\}$ remains below 10 dB was determined. To compensate for a possible attenuation of the hearing aid output by the feedback cancellation algorithm, the MSG is defined as the minimum of the maximum gain setting and the actual gain achieved at this maximum gain setting. The actual gain was computed as the average energy ratio of $u[k]$ to the reference signal $r[k]$ without amplification (i.e., $|G(z,k)| = 1$). To check reproducibility, eight repeated measurements were made.

Figure 2 depicts the MSG for two different input signals, i.e., the speech signal and the opera signal. The average MSG values over the eight repeated measurements are depicted with stars (*), the minimum MSG values are depicted with down-pointing triangles (∇), the maximum MSG values are depicted with up-pointing triangles ($\triangle$). For the speech signal, all algorithms achieve the same MSG of 19 dB, i.e., compared to 'No FC' an added stable gain of 15 dB is provided. For the speech signal, the delay of 6.3 ms sufficiently decorrelates $x[k]$ and $u[k]$ so that a good performance is

Algorithm	Gain $\|G(z,k)\|$	$E(k)$ mean	$E(k)$ max	$FSR(k)$ mean	$FSR(k)$ max	$TVC(k)$ mean	$TVC(k)$ max	$\Delta PCR(k)$ mean	$\Delta PCR(k)$ max	$SD(k)$ mean	$SD(k)$ max
No FC	4.0	1.2	7.6	-7.2	6.6	9.1	23.5	0.13	0.74	3.2	10.2
CAF	4.0	-0.4	1.0	-11.2	-8.2	2.2	4.0	0.00	0.00	1.9	4.5
	14.0	-0.4	1.1	-10.6	-7.5	2.3	4.2	0.00	0.00	2.1	5.3
	19.0	-0.2	2.3	-9.3	-1.5	3.0	11.5	0.01	0.35	2.4	10.6
BLCAF	4.0	-0.1	0.5	-14.0	-8.8	2.5	4.1	0.00	0.00	1.9	5.1
	14.0	0.4	1.8	-9.4	-3.5	3.5	10.3	0.00	0.07	2.5	6.5
	18.0	0.5	2.1	-8.8	-2.4	3.7	9.4	0.01	0.31	2.8	9.4
PemAFC-f	4.0	-0.4	0.7	-11.1	-6.9	2.3	4.1	0.00	0.00	1.9	4.8
	14.0	-0.4	0.7	-10.9	-6.3	2.3	4.2	0.00	0.00	2.0	5.0
	19.0	-0.1	3.6	-9.6	1.1	3.1	13.2	0.01	0.45	2.4	13.2
PemAFC-a	4.0	-0.3	0.8	-12.2	-8.5	2.0	3.5	0.00	0.00	1.8	4.4
	14.0	-0.3	0.9	-12.0	-8.5	1.9	3.5	0.00	0.00	1.9	4.4
	19.0	-0.0	3.9	-9.9	1.7	3.0	14.2	0.01	0.50	2.3	11.5
PemAFC-as	4.0	-0.4	0.9	-12.3	-9.6	2.0	3.6	0.00	0.00	1.8	4.4
	14.0	-0.3	0.9	-11.9	-9.2	2.0	3.5	0.00	0.00	1.8	4.6
	19.0	-0.1	2.3	-10.5	-1.3	2.7	11.5	0.01	0.36	2.2	10.6
Optimum	4.0	-0.0	0.0	-23.6	-14.1	0.4	1.0	0.00	0.00	0.9	1.7
	14.0	0.0	0.3	-20.4	-12.3	0.8	2.5	0.00	0.00	1.1	3.7
	19.0	0.2	3.9	-15.9	0.6	2.4	13.2	0.01	0.44	1.6	11.3

Table 1: Distortion measures at MSG$_{off}$, MSG$_{off}+10dB$ and MSG$_{on}$ for the speech signal.

Algorithm	Gain $\|G(z,k)\|$	$E(k)$ mean	$E(k)$ max	$FSR(k)$ mean	$FSR(k)$ max	$TVC(k)$ mean	$TVC(k)$ max	$\Delta PCR(k)$ mean	$\Delta PCR(k)$ max	$SD(k)$ mean	$SD(k)$ max
No FC	4.0	1.7	7.8	-6.6	6.8	8.6	18.6	0.13	0.59	2.8	6.5
CAF	4.0	-0.3	6.0	-7.3	5.0	6.0	29.4	0.03	0.51	3.2	10.3
BLCAF	4.0	-0.2	3.8	-6.5	1.2	5.5	18.9	0.02	0.40	3.1	7.1
	10.0	0.7	6.4	-3.9	5.0	9.6	27.4	0.08	0.66	4.3	10.7
PemAFC-f	4.0	-0.5	2.2	-7.7	0.7	5.5	25.7	0.02	0.35	3.1	9.2
	5.0	-0.0	10.2	-7.1	9.8	6.4	35.6	0.03	0.63	3.3	14.5
PemAFC-a	4.0	-0.3	3.4	-6.8	-1.0	5.2	11.7	0.02	0.46	3.1	7.1
	14.0	-0.1	5.1	-5.8	2.3	6.9	17.0	0.03	0.60	3.4	8.5
	22.0	0.2	4.7	-4.8	2.9	9.9	25.3	0.04	0.58	3.9	11.1
PemAFC-as	4.0	-0.4	2.8	-7.5	-1.5	4.7	9.3	0.01	0.34	2.9	7.3
	14.0	-0.3	2.3	-6.9	-2.1	5.6	12.9	0.02	0.29	3.0	7.6
	22.0	-0.0	2.6	-5.9	-1.0	8.4	21.5	0.02	0.35	3.6	8.0
Optimum	4.0	-0.0	0.1	-22.6	-10.6	0.6	2.1	0.00	0.00	0.9	1.7
	14.0	-0.0	0.2	-20.5	-11.2	2.0	13.3	0.00	0.00	1.1	3.4
	20.0	0.2	1.4	-14.6	-4.4	5.5	19.8	0.00	0.01	1.8	6.4

Table 2: Distortion measures at MSG$_{off}$, MSG$_{off}+10dB$ and MSG$_{on}$ for the opera signal.

achieved by the CAF. The MSG equals the optimum MSG that can be achieved with a 64-taps adaptive filter. This indicates that the MSG is limited by non-adaptation-related aspects such as the presence of non-linearities and reverberation and undermodelling. For the tonal opera signal, the PemAFC-a and PemAFC-as clearly outperform the CAF, the BL-CAF and the PemAFC-f thanks to the prewhitening with the adaptive desired signal model. Repeated measurements show good reproducibility for the speech signal. For the opera signal and the CAF, BL-CAF and PemAFC-f, the variance over the measurements is larger (due to a small stability margin at low gain settings). The optimum feedback canceller achieves a 2 dB lower ASG than the PemAFC-a and the PemAFC-as. The optimum feedback canceller was determined for a fixed hearing aid gain and by sending white noise through the hearing aid receiver. The lower ASG of the optimum feedback canceller may be due to differences in the PSD of the receiver signal and/or non-linearities in the hearing aid.

5.2 Sound quality

A high MSG is only valuable if the feedback canceller also preserves a good sound quality [8]. Assessment of the sound quality is thus necessary. Table 1 and Table 2 show the performance measures of Section 4 for the speech signal and the opera signal, respectively, at three gain settings[1]: MSG$_{off}$, MSG$_{off}+10$ dB and MSG$_{on}$. MSG$_{off}$ is the MSG without feedback canceller (i.e., 4 dB) while MSG$_{on}$ is the MSG with feedback canceller.

For the speech signal, all algorithms achieve good performance for MSG$_{off}$ and MSG$_{off}+10$ dB. The algorithms achieve less distortion compared to 'No FC'. The differences between the feedback cancellation algorithms are small. At MSG$_{on}$, the maximum of all the performance measures significantly increase while the mean performance measures show a much smaller increase. This is due to the fact that some segments of the non-stationary speech signal exhibit more distortion and oscillations than other speech segments.

For the opera signal, all algorithms exhibit more distortion and oscillations than for the speech signal: at MSG$_{off}$ and MSG$_{off}+10$ dB, the maximum performance measures are considerably higher than for the speech signal. The PemAFC-a and PemAFC-as outperform the CAF, BLCAF and PemAFC-f.

In general, the performance measures are consistent with each

[1] The measures for MSG$_{off}$+10 dB are only shown when MSG$_{off}+$ 10 dB does not exceed MSG$_{on}$.

Algorithm	$\|G(z,k)\| = 9dB$		$\|G(z,k)\| = 12dB$		$\|G(z,k)\| = 15dB$		$\|G(z,k)\| = 18dB$	
	min	max	min	max	min	max	min	max
CAF	1.17	1.59	2.41	19.87	3.88	>20.00	3.98	>20.00
BLCAF	0.54	0.58	2.37	2.67	3.16	4.33	4.14	8.35
PemAFC-f	0.41	0.55	1.13	2.85	2.29	>20.00	2.11	>20.00
PemAFC-a	0.41	0.42	0.89	0.94	1.24	1.43	1.68	2.02
PemAFC-as	0.23	0.24	0.28	0.30	0.47	0.54	0.61	0.82

Table 3: Oscillation time of the feedback cancellation algorithms.

other. For some conditions (e.g., the optimum feedback canceller for the opera signal at a gain of 14 dB) a high TVC is observed while the other measures are low. This may be due to the fact that the TVC does not take into account the amount of signal power at the detected spectral peak.

5.3 Time required to recover from instability

A third objective of a feedback canceller is fast tracking of changes in the acoustic feedback path, in particular when these changes lead to instability. In the previous experiments, the feedback cancellers were adapted starting from a stable hearing aid system: the forward path gain $G(z,k)$ was increased from a gain below MSG_{off} to an unstable gain. This experiment measures the ability of the adaptive feedback cancellers to recover from instability when starting from an unstable condition. This situation may for example occur when a hand or mobile phone is suddenly put close to the ear.

The forward path gain $G(z,k)$ was set above MSG_{off} and the filter coefficients of the feedback cancellers were initialized with zeros such that the initial hearing aid system was unstable. Then, the receiver signal $u[k]$ was recorded during 20 seconds while the adaptive feedback canceller was running and a stationary speech weighted noise was presented. In addition, the reference signal $r[k]$ in the absence of feedback was recorded. The time to recover from instability (called oscillation time) was determined as the time required for the short-term receiver-to-reference signal energy ratio $E(k)$ (using windows of 100 ms) to drop below a threshold of 10 dB. To achieve good temporal resolution, the energy ratio $E(k)$ was calculated using windows of 50-ms duration with 80% overlap.

Table 3 depicts the minimum and maximum oscillation time (in seconds) of ten repeated measurements for 4 different gain settings $|G(z,k)|$, i.e., $|G| = 9$ dB, $|G| = 12$ dB, $|G| = 15$ dB and $|G| = 18$ dB. For all gains, PemAFC-a and PemAFC-as have a smaller oscillation time than the other algorithms. In addition, the variance over the ten measurements is considerably smaller. PemAFC-as outperforms PemAFC-a thanks to the combination of a slowly and a fast adapting feedback cancellation filter. CAF and PemAFC-f do not always succeed to recover from instability: this is reflected in the oscillation times >20 s. At instability, the adaptation error power and input signal power of the CAF, BL-CAF and PemAFC-f are large at the oscillation frequencies. Due to the frequency-dependent step-size normalization, the algorithms are too slow at the oscillation frequencies to quickly reduce the adaptation error, causing problems to the filter adaptation. In PemAFC-a and PemAFC-as, the adaptive AR model whitens the adaptation error and hence reduces the strong oscillation frequencies in the adaptation error, improving the adaptation of the feedback canceller.

6. CONCLUSIONS

Objective measures were discussed for a real-time evaluation of feedback cancellation algorithms in the presence of spectrally colored input signals. The measures were applied to determine the maximum stable gain, the sound quality and the time to recover from instability of several adaptive feedback cancellation algorithms. All evaluated methods achieve worse feedback cancellation performance for a tonal music (opera) input signal than for a speech input signal. Among the evaluated methods, the PemAFC with adaptive desired signal model achieves the best feedback cancellation performance for the music signal and the shortest time to recover from instability.

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