OBJECTIVE EVALUATION OF FEEDBACK REDUCTION TECHNIQUES IN HEARING AIDS

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ABSTRACT

This paper proposes an objective evaluation of feedback reduction techniques in four commercial hearing aids and one recent feedback cancellation algorithm. The added stable gain was determined for a speech and a music signal in two acoustic conditions. In addition, the reduction of feedback and oscillations at gain values below the maximum stable gain was measured. Added stable gains between 1 dB and 21 dB were found. Most feedback cancellers achieve worse feedback suppression for the music signal than for the speech signal. Constraining the feedback canceller based on an initial feedback path measurement improves the performance for music signals at the expense of a worse feedback suppression in the acoustic conditions that differ from the condition for which the initialization was performed.

1. INTRODUCTION

Acoustic feedback poses a major problem to hearing aid users. Because of the acoustical coupling between the microphone(s) and the receiver, the sound signal cannot be amplified sufficiently. Although feedback cancellation algorithms have become common in digital hearing aids, there is still no standardized objective procedure for evaluating them.

An often used criterion for assessing the feedback suppression is the maximum stable gain (MSG), i.e., the maximum gain that can be applied without rendering the system unstable [1, 2, 3, 5, 6]. Objective procedures for determining the MSG are, however, limited to the case of a white noise input signal. Adaptive feedback cancellation algorithms in particular encounter problems when the input signal is spectrally colored, e.g., a music signal [4, 10, 12]. In addition, existing evaluation procedures typically assume that the hearing aid behaves as a linear amplifier. This is rarely the case in practice due to non-linear processing such as dynamic range compression and due to the saturation of the receiver at high gains.

In this paper, the performance of the feedback reduction techniques in four recent commercial hearing aids is assessed for spectrally colored input signals. A comparison is made with the prediction error method based feedback canceller described in [12]. The evaluation is based on objective performance measures for detecting the presence of feedback and oscillations [11]. The measures compare the actual hearing aid output with the hearing aid output obtained in the absence of feedback (i.e., reference signal). In [11], the reference signal was obtained from the hearing aid microphone recording in the absence of feedback (i.e., with the receiver disconnected), amplified by the same gain function as when the feedback canceller is active. This procedure assumes access to the microphone signal. However, in black-box commercial hearing aids, only the hearing aid output can be measured, e.g., with the in-the-ear microphone of an artificial head. In this paper, a procedure for estimating the feedback-free reference signal in black-box hearing aids is proposed based on replacing an open fitting by a closed fitting, with appropriate compensation. Based on the objective performance measures, the added stable gain with the feedback cancellers is determined for a speech and a music signal in two acoustic conditions. In addition, the reduction of feedback and oscillations at gain settings below the maximum stable gain is assessed.

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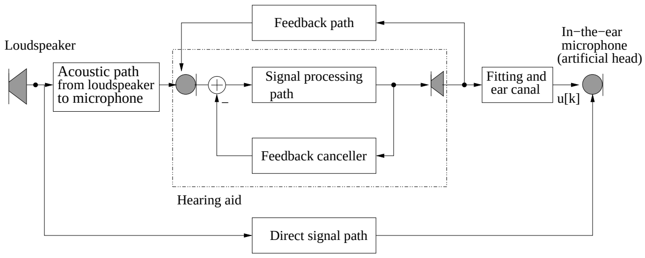

Figure 1: Block diagram of the set-up.

2. SET-UP AND HEARING AIDS

2.1 Set-up

The feedback evaluation was performed in a soundproof booth with a noise level of 20 dBA. The hearing aid was mounted on the left ear of a Cortex II artificial head using a Phonak Fit-and-Go open fitting. Signals were presented through a Fostex loudspeaker, positioned at 1 meter in front of the center of the head. The signal level equals 60 dBA, as measured at the center of the artificial head. Two test signals were used in the experiments: 17 seconds of real speech from the HINT (Hearing In Noise Test) database [8] and a 20 seconds opera fragment of ‘Der Hölle Rache’ from ’Die Zauberflöte of W.A. Mozart. The hearing aid output was recorded with the in-the-ear microphone of the Cortex II artificial head.

Two acoustic conditions were tested, referred to as ’Normal’ and ’Handset’. In the ’Handset’ condition, a handset was positioned on the left ear by means of a Velcro strap. In the ’Normal’ condition, there was no obstruction in the vicinity of the head.

2.2 Feedback reduction techniques

Table 1 lists the feedback reduction systems that were assessed. Four commercial power behind-the-ear (BTE) hearing aids were evaluated, referred to as A, B, C, D. The hearing aids were acquired in December 2007 and were at that time, the most recent BTEs. In the mean time, newer devices with improved feedback reduction capabilities may have been developed by the manufacturers. In addition, a frequency-domain implementation of a prediction-error method based adaptive feedback canceller (PemAFC) was considered, referred to as system E. A detailed description of the PemAFC algorithm can be found in [12]. The PemAFC uses a 20th-order adaptive all-pole desired signal model to reduce correlation between the desired signal and the input to the feedback canceller. To improve its tracking performance, the PemAFC combines a slowly adapting feedback canceller with a second fast adapting feedback canceller. The PemAFC algorithm was implemented on a linux PC that is connected to the front microphone and receiver of a Siemens Acuris BTE hearing aid through an RME Hammerfall DSP Multiface II sound card. The processing was done at a sampling frequency $f_s = 16$ kHz. Peak clipping was applied to the receiver input to keep the signals within the range of the DAC of the sound card.

The feedback reduction systems in hearing aids A, B, C and D are all adaptive feedback cancellers. Hearing aid B combines feedback cancellation with a frequency-dependent gain limitation. Hearing aids A, B and D require a feedback path measurement during fitting (initialization). The feedback path measurement was done for the 'Normal' condition. After the initialization, the earmold and the hearing aid were reconnected to the head, as this will also be the case in practice. The frequency-dependent gain limitation in hearing aid B is based on the measured feedback path. The gain limitation is applied at all times, even when the feedback canceller is disabled. To assess the impact of the gain limitation, the feedback suppression performance of hearing aid B with the feedback canceller switched off was also determined when no feedback path measurement was performed. In this case, a standard feedback path for a closed fitting was used by the fitting software. Hearing aid A uses the measured feedback path to constrain the adaptive feedback canceller. Hearing aid D employs the feedback path measurement in music mode. The feedback canceller in hearing aid C and E do not require an initialization of the feedback path. As indicated by Table 1, hearing aids B, C and D have a special mode for listening to music. Hearing aid B and C reduce the adaptation speed of the feedback canceller, while hearing aid D employs a static feedback canceller, i.e., the initialized feedback path. To assess the processing delay of the hearing aids, the delay between the hearing aid output and the direct signal path component in the ear was measured as the difference in peak location of the direct path impulse response and the hearing aid path impulse response. Delays vary from 4.4 msec to 7.1 msec.

2.3 Hearing aid settings

The commercial hearing aids A, B, C, and D were programmed using NOAH software. To assess the performance of the feedback reduction system only, all the other signal processing features (such as directionality, noise reduction, compression, expansion, ..) were disabled to the extent that this was made possible by the manufacturer's fitting software. These signal processing features may have a positive or negative effect on the performance of the feedback canceller. Hence, the presented results may not reflect the feedback reduction performance of the overall hearing aid system as used by a hearing aid user. At high gains, compression could not be completely switched off and may thus also have an impact on the results. In addition, at high gains, the gain in certain frequency bins (typically the higher frequencies) is limited in some devices. In hearing aid B, the maximum programmable gain in each frequency bin is constrained based on the initial feedback path measurement. As a result, already at low gains, an increase in the overall hearing aid gain setting does not result in an increase of the gain at all frequencies. The maximum output power (MPO) of the hearing aids A,B, C and D was set as high as possible in order to maximize the maximum programmable gain. In system E, compression and gain limitation was not used.

The frequency-specific gain controls of the hearing aids were tuned such that the hearing aid output power spectral density (PSD) as closely as possible matched a reference output PSD for a multisine input signal with a uniform amplitude spectrum (60 dBA at the center of the head). As a reference, hearing aid A with a flat gain control over frequency was used. The hearing aid output PSDs were measured with the feedback canceller switched off at a gain

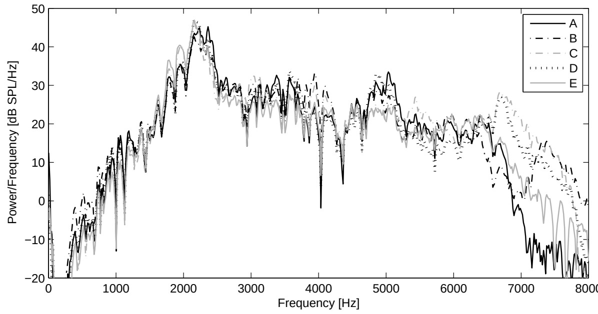

Figure 2: Hearing aid output power in dB SPL per frequency for a flat-spectrum input signal of 60 dBA.

of 18 dB or more below instability[1] such that it was not influenced by the presence of acoustic feedback or the feedback canceller. The hearing aid output was obtained as the difference between the in-the-ear microphone recording with the hearing aid switched on and the in-the-ear microphone recording with the hearing aid switched off, i.e., the direct signal path component. Figure 2 depicts the output PSD of the different hearing aids. Given the coarse controls for adjusting the frequency response, differences in the resulting output PSD of up to ± 10 dB could not be avoided. Above 7 kHz, even higher differences occurred because the frequency characteristic above 7 kHz was not always controllable.

3. OBJECTIVE EVALUATION PROCEDURE

The evaluation is based on objective measures for quantifying the amount of feedback and oscillations. To take into account spectral coloration of the input signal, the measures compare the actual hearing aid output $u[k]$ with the hearing aid output $r[k]$ that would be obtained in the absence of acoustic feedback (reference signal).

3.1 Feedback-free reference signal

As a reference signal, the hearing aid output (as measured by the in-the-ear microphone) at a gain far below instability is typically used. The gain difference between the actual hearing aid output and the low-gain output is then compensated for. This procedure, however, assumes that the hearing aid behaves as a linear system. This is rarely the case in practice due to non-linear processing such as dynamic range compression and frequency-dependent gain limitation. In addition, at high gains, the receiver of the hearing aid may become non-linear. In this paper, an alternative procedure for estimating the reference signal is proposed. The hearing aid output is recorded at the same gain as the actual hearing aid output but with a closed instead of an open fitting. For the closed fitting, a temporary foam earmold E-A-RTEMP 13A (EARtone) was used. Thanks to the closed fitting, the amount of feedback in the recording is minimal. The difference in frequency characteristic due to the closed fitting is compensated for by means of an FIR filter. The FIR filter is determined as the Wiener filter that estimates the hearing aid output with open fitting based on the output with closed fitting at a gain of 18 dB below instability with the feedback canceller switched off.

3.2 Performance measures

The amount of feedback is measured based on the intelligibility weighted feedback to desired signal ratio. In addition, the modifed transfer function variation criterion and power concentration ratio defined in [11] will be used to detect the presence of oscillations. Some segments of the speech and opera signal may be more prone to feedback and oscillations than other segments. Therefore, the performance measures will be computed using frames of 0.5 sec with an overlap of 80%. To assess the performance, the maximum measure over the whole speech and opera signal will be used.

[1] For B, instability in the absence of the gain limitation is meant.

	Algorithm type	Initialization	Music Mode	Delay
A	feedback canceller with constrained adaptation	yes	-	5.6 ms
B	feedback canceller and frequency-dependent gain limitation	yes	reduced adaptation speed	7.1 ms
C	feedback canceller	no	reduced adaptation speed	4.4 ms
D	feedback canceller	yes	static feedback canceller	4.7 ms
E	PEM-based feedback canceler	no	-	6.5 ms

Table 1: Evaluated feedback reduction sytems and properties. The delay is defined as the delay between the direct signal path component and the hearing aid output, as measured by the in-the-ear microphone of the artificial head.

3.2.1 *Intelligibility-weighted feedback to desired signal ratio*

To quantify the amount of feedback, the short-term intelligibility weighted feedback to desired signal ratio $\mathrm{FSR}(k)$ in the hearing aid output was computed as:

$$\mathrm{FSR}(k) = \sum_i I_{ERB,i} 10 \log_{10} \frac{\int_{f \in B_i} P_v(f,k) df}{\int_{f \in B_i} P_r(f,k) df}, \qquad (1)$$

with $P_v(f,k)$ the short-term PSD of the feedback signal $v[k] = u[k] - r[k]$ and $P_r(f,k)$ the short-term PSD of the reference signal $r[k]$. The weight $I_{ERB,i}$gives an equal weight to each auditory critical band B_i between 300 Hz and 6500 Hz, defined by the equivalent rectangular bandwidth (ERB) of auditory filters [7]. To limit the impact of environmental and internal noise on the measurements, signal frames are only included in the computation of the maximum of $\mathrm{FSR}(k)$ over the signal when the intelligibility weighted reference signal to environmental and internal noise ratio was 10 dB or higher. As an estimate for the environmental and internal noise, the in-the-ear microphone signal with the hearing aid at a gain setting of 6 dB below $\mathrm{MSG_{off}}$[2] and with the feedback canceller disabled was recorded in silence. The noise signal was rescaled to compensate for the gain difference between $\mathrm{MSG_{off}}$- 6 dB and the actual gain.

3.2.2 *Transfer function variation criterion (TVC)*

To detect the presence of oscillations, the difference $\mathrm{TVF}(f,k)$ in amplitude characteristic between the actual hearing aid transfer function and the desired hearing aid transfer function (i.e., in the absence of feedback) is estimated as

$$\mathrm{TVF}(f,k) = 10 \log_{10} \left(\frac{\max(P_u(f,k), \alpha P_n(f))}{\max(P_r(f,k), \alpha P_n(f))} \right). \qquad (2)$$

This difference is referred to as the transfer function variation function [9, 11]. 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 by $\alpha P_n(f)$ where $P_n(f)$ is the long-term PSD of the internal and environmental noise $n[k]$ and $\alpha > 1$. The largest peak or dip in the transfer function variation function

$$\mathrm{TVC(k)} = \max_f(|\mathrm{TVC}(f,k)|), \qquad (3)$$

referred to as transfer function variation criterion (TVC), is then used to assess the presence of oscillations.

3.2.3 *Power concentration ratio (PCR)*

In [1], 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 measure, however, assumes that the input to the hearing aid is white. In order to be applicable to spectrally colored input signals, a modified measure based on the PCR was proposed in [11]:

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

$$\Delta\mathrm{PCR}(k) = \mathrm{PCR}_u(k) - \mathrm{PCR}_r(k) \qquad (4)$$

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

The TVC and PCR measures are computed on the frequency range from 500 Hz to 6500Hz. Outside this frequency range, the hearing aid output is low (cf. Figure 2) and hence, susceptible to noise.

3.3 Added stable gain

The added stable gain (ASG) is defined as the difference in the maximum stable gain (MSG) with the feedback canceller enabled ($\mathrm{MSG_{on}}$) and the MSG with the feedback canceller disabled ($\mathrm{MSG_{off}}$). The ASG was determined following two procedures, one (ascending protocol) in which the gain is gradually increased through the manufacturer's fitting software until instability occurs and one (descending protocol) in which the gain is gradually decreased until instability disappears. The step size of the gain control was 1 dB for A, B, C and E and 2 dB for D. At each gain setting, the hearing aid output was recorded and the maximum short-term feedback to desired signal ratio $\max_k\{\mathrm{FSR}(k)\}$ over the whole signal segment was computed. To avoid adaptation effects, the signal was presented once before the recording was made. The maximum gain setting for which $\max_k\{\mathrm{FSR}(k)\}$ remained below 0 dB was determined. Alternative criteria for instability can be found in [11]. To compensate for a possible frequency-dependent attenuation of the hearing aid output by the hearing aid (e.g., hearing aid B), the actual gain at the maximum gain setting was computed as the average gain between 500 Hz and 6500 Hz of the reference signal at the maximum gain setting compared to a reference gain setting.

4. RESULTS

4.1 Added stable gain

Figure 3 depicts the ASG of the feedback cancellation algorithms for the speech and the opera signal under the two acoustic conditions. The up-pointing triangulars show the ASG according to the ascending protocol. The down-pointing triangulars show the ASG according to the descending protocol. For hearing aids A, C, D and E, retest data are also shown. For the opera signal, the ASG with the music mode of hearing aids B, C and D is depicted too (referred to as B-m, C-m, D-m). For hearing aid B, the ASG that is obtained with the gain limitation only is depicted with squares. This ASG was determined as the difference between $\mathrm{MSG_{off}}$ with feedback path initialization and $\mathrm{MSG_{off}}$ without feedback path initialization.

Due to the gain limitation, instability was never reached with hearing aid B in the 'Normal' condition, even when the feedback canceller was switched off. As a result, the ASG that is offered by the feedback canceller could not be determined. This is indicated with the double facing arrow. In addition, the depicted ASG that is obtained with the gain limitation only is a lower bound (which is indicated by the upward-pointing arrow). In the 'Handset' condition, gain limitation occurred at $\mathrm{MSG_{on}}$ (except for the music mode):

[2]For B, $\mathrm{MSG_{off}}$ in the absence of the gain limitation is meant.

the maximum added gain without limitation equals 7 dB. For hearing aids A and D, compression occurred at MSG_{on} in the 'Normal' condition. In addition, the gain was limited at certain frequencies because the maximum output or the maximum hearing aid gain was reached.

For the 'Normal' condition, an ASG between 12 dB and 20 dB is achieved for the speech signal. The ASG of C, D and E is not seriously affected by the presence of the handset. The ASG of hearing aid A drops from 20 dB for the 'Normal' condition to 5-7 dB for the 'Handset' condition. Due to the constrained adaptation, only small deviations from the feedback path in the 'Normal' condition can be modelled by the feedback canceller in hearing aid A, explaining the worse feedback suppresion for the 'Handset' condition.

For the opera signal, the ASG ranges from 1 dB to 21 dB for the 'Normal' condition and from 1 dB to 18 dB for the 'Handset' condition. Hearing aids C and D generally achieve a lower ASG for the opera signal than for the speech signal. Except for hearing aid D in the 'Normal' condition, the music mode of hearing aids B, C and D does not increase the ASG. Thanks to the constrained adaptation, hearing aid A achieves a high ASG for the opera signal in the 'Normal' condition. Hearing aid E still achieves a high ASG for the opera signal. However, oscillations already occur at gain settings below MSG_{on} (see Section 4.2).

The difference between test and retest data and the ascending and descending protocol are in general limited to 1 dB a 2 dB, which corresponds to the step size of the gain control. For the opera signal, larger differences sometimes occur. For the adaptive feedback cancellers, the internal and environmental noise may result in a small change in the filter coefficients of the feedback canceller. However, when the adaptive feedback canceller operates close to instability, a small change in filter coefficients may result in a large change in the feedback signal level and hence, $FSR(k)$. Tbe larger differences between test and retest data and the ascending and descending protocol for C-m and E may be due to a small gain margin at intermediate gain settings. For the static feedback canceller D-m, the difference in ASG between test and retest in the 'Normal' condition is 6 dB. A small deviation between the initialized and the actual feedback path may have a big impact on the feedback suppression performance. After initialization of the feedback canceller, the earmold was reconnected to the hearing aid, which resulted in a small change in feedback path. To illustrate the optimal performance of D-m, the feedback canceller was re-initialized after the earmold was reconnected (referred to as ideal static filter). In this case an ASG of 18 dB was obtained.

4.2 Performance at gains between MSG_{off} and MSG_{on}

Table 2 and Table 3 depict the maximum $FSR(k)$ and the maximum $TVC(k)$ for the speech signal and the opera signal, respectively, at gains between MSG_{off} and MSG_{on} (in steps of 6 dB) for the 'Handset' condition. The performance measures of the ascending and the descending protocol were averaged. For hearing aids A, C, D and E, both test and re-test data are provided.

For the speech signal, all feedback cancellers achieve good performance below MSG_{on}: the amount of feedback and oscillations is reduced compared to the hearing aid output with feedback canceller switched off at MSG_{off}. For the tonal opera signal, performance of hearing aids B, C, D and E degrades and oscillations are present after feedback cancellation, even at gains below MSG_{on}. For hearing aid D, performance at MSG_{off} is improved by the music mode. However, the ASG is worse due to the static feedback canceller that is not optimal for the 'Handset' condition. For hearing aid B and C, no improvement can be observed by the music mode for the opera signal in the 'Handset' condition.

Test and retest data are consistent with each other. Differences in the sound pressure level between test and retest data at MSG_{off} were due to small variations in the positioning of the handset. In addition, it should be noted that close to instability, a small difference in the estimated feedback path may result in a big difference in the hearing aid output and hence, the performance measures. This explains why the difference between test and retest data is larger at gains close to instability.

	FC	Gain	dB SPL	FSR	TVC	ΔPCR
A	off	MSG_{off}	75/ 77	-5.7/ -6.1	26/ 16	0.8/ 0.4
	on	MSG_{off}	75/ 77	-7.4/ -8.7	6/ 6	0.0/ 0.0
	on	MSG_{on}	80/ 83	-3.8/ -4.8	28/ 18	0.8/ 0.4
B	off	MSG_{off}	77	-6.0	15	0.2
	on	MSG_{off}	77	-10.6	5	0.0
	on	+6	84	-8.6	6	0.0
	on	MSG_{on}	90	-3.1	32	0.7
C	off	MSG_{off}	76/ 75	-6.5/ -2.5	18/ 37	0.7/ 1.0
	on	MSG_{off}	76/ 75	-8.9/ -8.1	7/ 8	0.1/ 0.1
	on	+6	82/ 81	-7.3/ -6.1	8/ 9	0.3/ 0.3
	on	MSG_{on}	87/ 85	-2.0/ -3.0	22/ 16	0.6/ 0.5
D	off	MSG_{off}	75/ 76	-5.5/ -4.7	13/ 18	0.5/ 0.7
	on	MSG_{off}	75/ 76	-7.0/ -7.1	5/ 5	0.0/ 0.0
	on	+6	81/ 82	-6.4/ -6.4	6/ 7	0.0/ 0.0
	on	+12	87/ 88	-5.3/ -5.0	8/ 12	0.1/ 0.2
	on	MSG_{on}	93/ 92	-1.9/ -2.8	24/ 22	0.6/ 0.6
E	off	MSG_{off}	73/ 73	-5.0/ -5.3	19/ 19	0.7/ 0.7
	on	MSG_{off}	73/ 73	-7.1/ -7.9	5/ 5	0.0/ 0.0
	on	+6	79/ 80	-7.4/ -8.2	5/ 4	0.0/ 0.0
	on	+12	84/ 85	-7.5/ -8.3	5/ 5	0.0/ 0.0
	on	MSG_{on}	90/ 90	-5.6/ -6.4	11/ 11	0.5/ 0.5

Table 2: Sound pressure level (dB), $\max_k\{FSR(k)\}$ (dB), $\max_k\{TVC(k)\}$ (-) and $\max_k\{\Delta PCR(k)\}$ (-) at different gains between MSG_{off} and MSG_{on} for the speech signal in the 'Handset' condition (test/retest).

5. CONCLUSIONS

In this paper, the performance of the feedback reduction techniques in four commercial hearing aids and one recent feedback cancellation technique was assessed based on an objective procedure. The ASG was determined for a speech and an opera signal in two acoustic conditions. In addition, the reduction of feedback and oscillations at gain values below the maximum stable gain was assessed. For the speech signal, the ASG ranges from 12 dB to 20 dB for the 'Normal' condition and from 5 dB to 18 dB for the 'Handset' condition. Hearing aids B, C, D and E achieve worse feedback suppression for the tonal opera input signal than for the speech input signal: even at gains below MSG_{on}, oscillations occur. The music mode of B and C do not result in an improved performance for the opera signal. Constraining the adaptive feedback canceller based on a feedback path measurement results in improved performance for tonal signals at the expense of a worse feedback suppression in the acoustic conditions that differ from the condition for which the initialization was performed.

6. ACKNOWLEDGMENT

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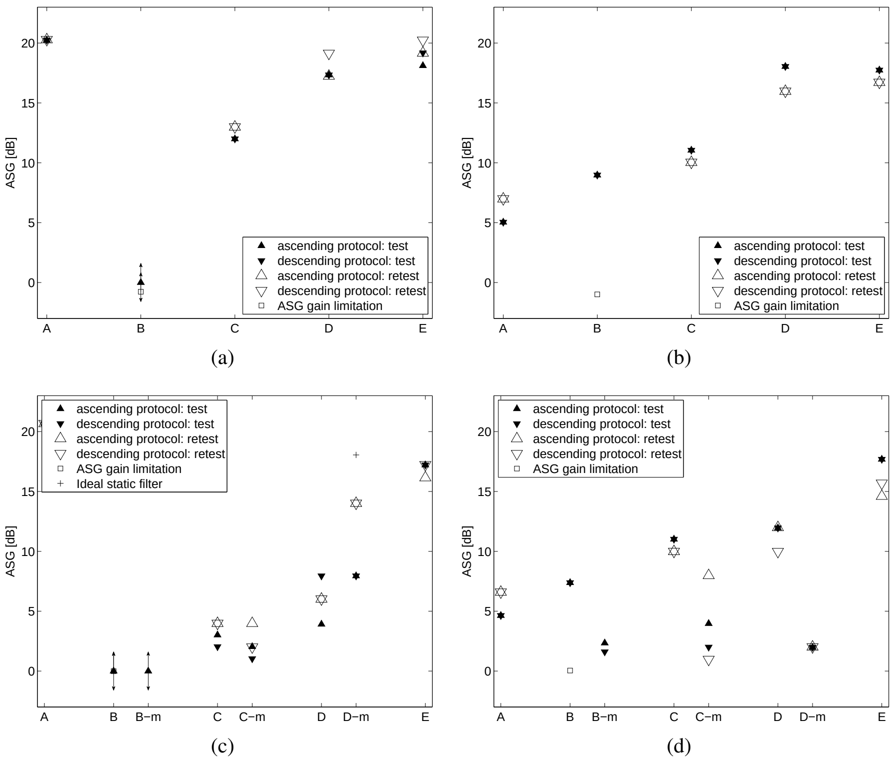

Figure 3: ASG of the different feedback cancellers for the speech and the opera signal in the 'Normal' condition (a-c) and the 'Handset' condition (b-d). Up-pointing triangulars show the ASG of the ascending protocol; down-pointing triangulars show the ASG of the descending protocol. A double-pointing arrow indicates that ASG could not be determined. An up-pointing arrow indicates that only a lower bound of the ASG could be measured.

	FC	Gain	dB SPL	FSR	TVC	ΔPCR
A	off	MSG_{off}	74/ 79	-4.6/ -4.9	10/ 12	0.2/ 0.3
	on	MSG_{off}	74/ 79	-5.9/ -6.2	7/ 6	0.0/ 0.0
	on	MSG_{on}	79/ 85	-2.5/ -3.3	42/ 30	0.9/ 0.3
B	off	MSG_{off}	78	-1.4	30	0.8
	on	MSG_{off}	78	-5.6	6	0.0
	on	+6	88	-2.5	13	0.2
	on	MSG_{on}	91	-1.2	29	0.5
B-m	on	MSG_{off}	78	-3.9	15	0.4
	on	MSG_{on}	80	-1.7	27	0.6
C	off	MSG_{off}	77/ 75	-6.0/ -4.9	15/ 41	0.6/ 1.0
	on	MSG_{off}	77/ 75	-6.8/ -6.4	7/ 7	0.0/ 0.0
	on	+6	84/ 81	-4.5/ -4.3	13/ 11	0.4/ 0.3
	on	MSG_{on}	88/ 85	-1.6/ -1.5	20/ 21	0.7/ 0.6
C-m	on	MSG_{off}	77/ 75	-6.6/ -6.3	10/ 9	0.1/ 0.1
	on	MSG_{on}	79/ 76	-4.6/ -6.1	30/ 8	0.6/ 0.1
D	off	MSG_{off}	76/ 76	-5.4/ -5.0	10/ 13	0.3/ 0.5
	on	MSG_{off}	76/ 76	-5.0/ -5.5	15/ 18	0.2/ 0.3
	on	+6	82/ 82	-4.5/ -4.4	20/ 22	0.5/ 0.3
	on	MSG_{on}	88/ 86	-2.4/ -2.5	32/ 30	0.6/ 0.6
D-m	on	MSG_{off}	76/ 76	-5.8/ -5.9	6/ 7	0.0/ 0.0
	on	MSG_{on}	78/ 78	-5.0/ -5.1	11/ 12	0.3/ 0.3
E	off	MSG_{off}	74/ 76	-4.8/ -3.7	14/ 23	0.6/ 0.8
	on	MSG_{off}	74/ 76	-3.8/ -3.7	8/ 10	0.0/ 0.4
	on	+6	80/ 82	-4.1/ -3.9	12/ 12	0.4/ 0.4
	on	+12	86/ 87	-4.1/ -3.3	12/ 8	0.4/ 0.4
	on	MSG_{on}	90/ 91	-3.9/ -3.7	11/ 10	0.4/ 0.4

Table 3: Sound pressure level (dB), $\max_k\{\mathrm{FSR}(k)\}$ (dB), $\max_k\{\mathrm{TVC}(k)\}$ (-) and $\max_k\{\Delta\mathrm{PCR}(k)\}$ (-) at different gains between MSG_{off} and MSG_{on} for the opera signal in the 'Handset' condition (test/retest).

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