Evaluation of feedback reduction techniques in hearing aids based on physical performance measures[a)]

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(Received 28 July 2009; revised 3 June 2010; accepted 7 June 2010)

This paper presents a physical evaluation of four feedback cancellation techniques in commercial hearing aids and two implementations of a recently developed feedback cancellation algorithm. Based on physical measures for detecting instability, oscillations and distortion, three performance aspects were measured: 1) the added stable gain compared to the hearing aid operating without feedback reduction for white noise as well as for spectrally colored input signals in two static acoustic conditions, 2) the amount of feedback, oscillations and distortion at gain values below the maximum stable gain, 3) the ability to track feedback path changes. Added stable gains between 3 dB and 26 dB were identified. Five of the six techniques achieve worse feedback reduction for a tonal opera input signal than for a speech input signal. Preventing the feedback canceller to drift away from an initial feedback path measurement results in improved performance for tonal signals at the expense of a worse feedback reduction in the acoustic conditions that differ from the condition for which the initialization was performed, as well as a worse tracking of feedback path changes. Repeated measures indicated that the reproducibility of the test set-up is crucial, in particular when the hearing aid operates close to instability. © *2010 Acoustical Society of America.*
[DOI: 10.1121/1.3458850]

PACS number(s): 43.66.Ts, 43.60.Mn [WMC] Pages: 1245–1261

I. INTRODUCTION

Acoustic feedback poses a major problem to hearing aid users. Because of the acoustic coupling (feedback) between the hearing aid receiver and the microphone(s), a closed-loop system is formed. The closed-loop system may become unstable when a large signal amplification (gain) is applied in the hearing aid. As a result, the sound signal often times cannot be amplified sufficiently. To reduce the acoustic feedback, feedback cancellation may be applied (see Fig. 1). 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(s) (Chi *et al.*, 2003; Greenberg *et al.*, 2000; Hellgren, 2002; Boukis *et al.*, 2007; Kates, 2003; Maxwell and Zurek, 1995; Siqueira and Alwan, 2000; Spriet *et al.*, 2005). The acoustic path between the receiver and the microphone(s) can vary significantly depending on the acoustic environment: reflecting surfaces close to the ear, such as a telephone handset or the palm of a hand, can temporarily reduce the feedback path attenuation by 10 to 20 dB and hence, cause instability (Hellgren *et al.*, 1999; Kates, 2001; Rafaely *et al.*, 2000; Stinson and Daigle, 2004). To deal with these feedback path changes, adaptive feedback cancellation techniques are typically used.

Although feedback reduction techniques have become common in digital hearing aids, there is still no standardized objective procedure for evaluating them. The major objectives of a feedback reduction technique are:

- The feedback reduction technique should provide a high maximum stable gain (MSG) compared to the hearing aid system without feedback reduction. The MSG is defined as the maximum gain that can be applied without rendering the system unstable.
- The feedback reduction technique should have a low susceptibility to tonal signals and preserve a good sound quality.
- The feedback reduction should be able to quickly adjust to feedback path changes so that when the system becomes unstable, the duration of instability is minimal.

For all three performance aspects, physical measures have been proposed in the literature. The main shortcoming of these measures is the limited applicability to specific input signals (such as a white noise input signal), linear hearing aid behavior and/or operation below instability. In practice, a hearing aid rarely acts as a linear amplifier due to non-linear processing such as dynamic range compression and due to the saturation of the receiver at high gains.

Physical procedures for determining the MSG are generally limited to the case of a white noise input signal (Freed and Soli, 2006; Merks *et al.*, 2006; Shin *et al.*, 2007) and/or linear hearing aid behavior (Shin *et al.*, 2007; Merks *et al.*,

[a)]Part of this work was presented at the European Signal Processing Conference (EUSIPCO), Glasgow, Scotland, August 2009.
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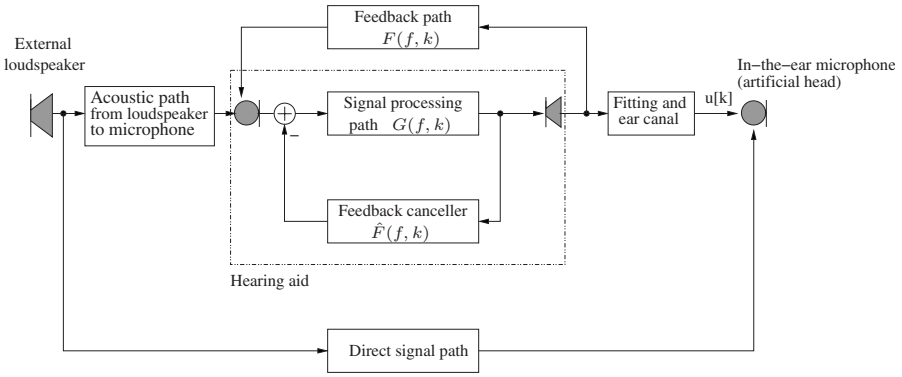


FIG. 1. Block diagram of the evaluation set-up. The components in the dashed-dotted square represent the hearing aid. The in-the-ear microphone of the artificial head captures the hearing aid output $u[k]$ as well as the sound that directly enters the ear through the vent (direct signal path).

2006). In Freed and Soli (2006), Greenberg *et al.* (2000), Maxwell and Zurek (1995), Freed (2008), and Grimm and Hohmann (2006), the MSG is determined by gradually increasing the hearing aid gain until instability occurs or by gradually decreasing the gain until instability stops. For a white noise input signal, Freed and Soli (2006) and Shin *et al.* (2007) proposed physical criteria for instability based on the power concentration ratio and the hearing aid transfer function variation, respectively. In Gao and Soli (2000) and Merks *et al.* (2006), the loop gain is estimated through impulse response measurements from an external loudspeaker to the ear-canal of an artificial head at three hearing aid gain settings: a low gain, a high gain and a zero gain (with the hearing aid switched off). Adaptive feedback cancellation algorithms in particular encounter problems when the input signal is spectrally colored, e.g., a music signal (Hellgren, 2002; Spriet *et al.*, 2006; Siqueira and Alwan, 2000). Due to correlation between the input to the feedback cancellation algorithm and the hearing aid input signal, the feedback canceller erroneously attempts to cancel the hearing aid input signal instead of the acoustic feedback. As a result, the MSG determined for a white input signal is often an overestimation of the maximum gain that can be applied in real-life scenarios. In addition, the MSG is typically derived based on the hearing aid gain setting (Freed and Soli, 2006; Merks *et al.*, 2006). Some devices, however, automatically reduce their gain in order to avoid feedback. In this case, the gain setting will not correspond to the actually applied gain, which then indeed obscures the analysis.

In Freed and Soli (2006) and Merks *et al.* (2006), the susceptibility to tonal input signals was assessed. In Freed and Soli (2006), pure tones were presented to the hearing aid. Artifacts were detected by measuring the output power at the extraneous frequencies as a percentage of the power at the input frequencies. However, the method is limited to pure tone input signals. In Merks *et al.* (2006), the robustness to periodic input signals was assessed by measuring the occurrence of entrainment artifacts when presenting the hearing aid with music, machine noise, tonal and human sounds (child's voice, finger snapping). Entrainment is typically described as feedback after cessation of the sound: additional tones, warbling and echos. It occurs when the feedback canceller erroneously attempts to cancel a tonal input to the hearing aid. The entrainment is computed based on the difference between the short-term spectra of the hearing aid output with the feedback reduction technique disabled and enabled at a hearing aid gain of at least 5 dB below instability (with the feedback reduction technique disabled). With regard to sound quality, models have been proposed in the literature for assessing sound quality perceived by normal hearing (Huber and Kollmeier, 2006; Thiede *et al.*, 2000) as well as hearing impaired listeners (Beerends *et al.*, 2008; Kates and Arehart, 2009; Arehart *et al.*, 2007). These quality models compare the output signal with a clean reference signal. The models were mainly developed and evaluated for the assessment of quality degradation due to coding, additive noise and clipping distortions. In order to be applicable to the evaluation of feedback reduction systems, a feedback-free reference signal is required. In addition, the models should be insensitive to modifications in the hearing aid signal processing path caused by the feedback reduction technique (such as notch filtering or phase modifications) that are perceptually not disturbing.

Freed and Soli (2006) measured the oscillation time of the feedback reduction technique for a white noise input when a sudden change in the feedback path occurred. The feedback path change was introduced by manually placing a hat on the artificial head. Since the trajectory of the hat placement was not controlled, reproducibility was poor.

The main reason for the limited applicability of existing physical measures is the lack of an accurate estimation of a feedback-free reference signal, in particular, in the case of non-linear processing. In Spriet *et al.* (2008), the reference signal is obtained as the hearing aid microphone recording when the receiver is disconnected, amplified by the same gain function as if the feedback canceller were 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 is proposed for estimating the feedback-free reference signal in black-box hearing aids based on replacing an open fitting by a closed fitting, with appropriate compensation. Thanks to the reference signal, physical performance measures can be defined for spectrally colored input signals. This study focuses on performance measures for detecting feedback, oscillations and distortion (Spriet *et al.*, 2008, 2009). Once accurate physical measures

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are obtained, the application of quality models and the correlation with perceived quality can be studied. Based on the physical performance measures, four adaptive feedback cancellation based feedback reduction techniques as implemented in recent commercial hearing aids are evaluated for spectrally colored input signals. In addition, two prediction error method based adaptive feedback cancellers described in Spriet *et al.* (2006) are implemented on an experimental hearing aid platform and evaluated. Based on the physical performance measures, the added stable gain (ASG) with the feedback reduction is determined for four different input signals in two acoustic conditions. The ASG is determined based on the true hearing aid gain and not based on the hearing aid gain setting. In addition, the reduction of feedback, distortions and oscillations at several gains below the MSG is assessed. Finally, the ability of the adaptive feedback cancellers to track feedback path changes is investigated by means of a PC controlled motorized set-up.

II. SET-UP AND HEARING AIDS

A. Set-up

The feedback reduction evaluation was performed in a soundproof booth with a background noise level of 20 dBA. Figure 1 depicts the set-up. All hearing aid devices were mounted on the left ear of a steerable Cortex II artificial head. To maximize feedback, a Phonak Fit-and-Go open fitting was used. The Phonak Fit-and-Go consists of a standard tubing with a retention hook to hold the tubing in the ear (Chung, 2004). The artificial head was mounted by means of a planetary gear head (Bayside PX60–010–002) to a brushless DC servo-motor (Animatics SmartMotor 2315 D) to allow rotation around its axis that is controlled by a PC through an RS232 cable. Signals were presented through a Fostex loudspeaker, positioned at 1 m in front of the center of the head. The signal level was set at 60 dBA, as measured at the center position of the artificial head in absentia. Four test signals were used in the experiments: 10 s of stationary white noise and 10 s of speech weighted noise from the HINT (Hearing in Noise Test) database (Nilsson *et al.*, 1994), 17 s of male speech from the HINT database and a 20 s opera fragment of 'Der Hölle Rache' from 'Die Zauberflöte' of W.A. Mozart, sung by a soprano. The signal at the in-the-ear microphone of the Cortex II artificial head was amplified and recorded with an RME Hammerfall DSP Multiface II sound card at a sampling frequency of 32 kHz. The



FIG. 2. (Color online) Handset condition.

in-the-ear signal consists of two components: the hearing aid output $u[k]$ and the sound that directly enters the ear through the vent of the earmold (i.e., the direct signal path component). The latter is measured by recording the in-the-ear microphone signal with the hearing aid switched off. The hearing aid output $u[k]$ can be computed by subtracting the in-the-ear microphone signal with the hearing aid switched off from the signal with the hearing aid switched on, since the acquisition of both signals was synchronized with the presentation of the same test signal.

Two static acoustic conditions were tested, referred to as 'Normal' and 'Handset'. In the 'Handset' condition, a handset was positioned on the left ear of the artificial head by means of a Velcro strap (see Fig. 2). The position of the Velcro strap and the handset was marked on the artificial head to reduce variations in the positioning of the handset. In the 'Normal' condition, there was no obstruction in the vicinity of the head. In addition, a third condition with a dynamic feedback path was tested to assess tracking performance (cf. Sec. III C 4).

B. Feedback reduction techniques

Table I lists the feedback reduction systems that were evaluated. Four commercial power behind-the-ear (BTE) hearing aids were evaluated, referred to as hearing aids A, B, C, D. The hearing aids were acquired in December 2007 and were, at that time, the most recent BTEs on the market. In the meantime, newer devices with improved feedback reduction capabilities may have been developed by the manufacturers. The BTEs evaluated here all have adaptive feedback cancellation based feedback reduction (see Fig. 1). In addi-

TABLE I. Evaluated feedback reduction systems 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.

	Manufacturer	Model	Algorithm type	Initialization	Music mode	Delay (ms)
A	GN Resound	Azure AZ80-DVI	Feedback canceller initialized based on a feedback path measurement	Yes	—	5.6
B	Phonak	Savia Art 411 dSZ	Feedback canceller and frequency-dependent gain limitation	Yes	Reduced adaptation speed	7.1
C	Siemens	Centra HP	Feedback canceller with automatic speed control	No	Reduced adaptation speed	4.4
D	Starkey	Destiny 400	Feedback canceller	Yes	Static feedback canceller	4.7
E	K.U. Leuven	N/A	PEM-based feedback canceler (PemAFC) (E)	No	—	6.5
			PemAFC combining a fast and slowly adaptive filter (E-s)	No	—	6.5

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tion, two frequency-domain implementations of a prediction error method based adaptive feedback canceller (PemAFC) were considered, referred to as system E and E-s. A detailed description of the PemAFC algorithms can be found in Spriet *et al.* (2006). The PemAFC in system E uses a 20th-order adaptive all-pole desired signal model to reduce correlation between the so-called desired signal and the input to the feedback canceller. To improve its tracking performance, the PemAFC in system E-s 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 the 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 input signal of the receiver to keep the signal within the range of the DAC of the sound card. The processing complexity of the PemAFC algorithms exceeds the complexity of the feedback reduction techniques in the commercial hearing aids. This should be kept in mind when comparing their performance to that of the commercial hearing aids.

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 Phonak Fit-and-Go and the hearing aid were removed and reconnected to the head, as it will also be the case in practice. The feedback canceller in hearing aid C and in system E and E-s do not require any initialization of the feedback path. Hearing aid A uses the measured feedback path as a starting point for the adaptive feedback canceller. When used in combination with a slow adaptation constant, its behavior is similar to using constrained adaptation (Kates, 1999, 2003), i.e., temporarily large variations from the measured feedback path will not be tracked. Hearing aid B combines feedback cancellation with a frequency-dependent gain limitation. The fitting philosophy of Phonak is to not enable over-critical gain during fitting ('Normal' condition), i.e., gain above the MSG with the feedback canceller disabled, in order to prevent feedback occurring in situations when an object is approached to the ear. The frequency-dependent gain limitation 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 reduction performance of hearing aid B with the feedback canceller disabled was also determined when no feedback path measurement was performed ('No Init'). In this case, a standard feedback path for a closed fitting is used by the fitting software. In hearing aid C, the adaptation speed of the feedback canceller is selected based on an analysis of the input signal. As indicated by Table I, hearing aids B, C and D have a special mode for music signals. In music mode, hearing aid B and C reduce the adaptation speed of the feedback canceller, while hearing aid D replaces the adaptive feedback canceller with a static one, i.e., the initialized feedback path. To assess the average processing delay (over frequency) 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 the maximum 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.

C. Hearing aid settings

Hearing aids A, B, C, and D were programmed using NOAH software from HIMSA (the Hearing Instrument Manufacturer's Software Association). To assess the performance of the feedback reduction system only, all other signal processing features (such as directionality, noise reduction, compression, and 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 actual feedback reduction performance of the overall hearing aid system. At high gains, compression cannot 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, even 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 hearing aids A, B, C and D was set as high as possible in order to maximize the maximum programmable gain. In system E and E-s, there is no compression and no gain limitation.

The frequency-specific gain controls of the hearing aids were tuned such that the hearing aid output power spectral density (PSD) optimally matched a reference output PSD for a multi-sine 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 reason for choosing a flat gain control over frequency for hearing aid A was to maximize the range of hearing aid gain settings with equal gain adjustments across frequencies while minimizing the amount of feedback at the lowest gain setting within that range. The hearing aid output PSDs were measured with the feedback canceller disabled at a gain setting of 18 dB below instability[1] such that the measurement was not influenced by the presence of acoustic feedback or the feedback canceller. The maximum stable gain setting (MSGS) with the feedback canceller disabled will be referred to as $\mathrm{MSGS}_{\mathrm{off}}$.[2] Figure 3 depicts the output PSDs of the different hearing aids: the output PSDs of hearing aids B, C, D and E were scaled such that they have the same average level between 500 Hz and 6500 Hz as hearing aid A. 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 6.5 kHz, even higher differences occurred because the frequency characteristic was not always controllable in this frequency range.

III. METHODS

The evaluation is based on physical measures for quantifying the amount of feedback, oscillations and distortion.

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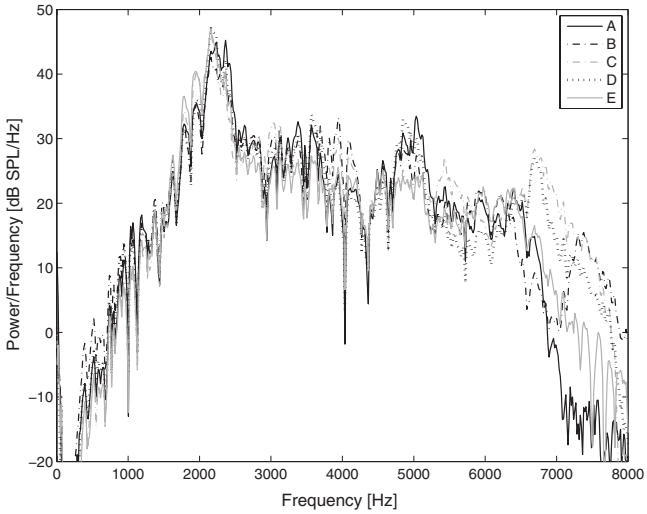


FIG. 3. Hearing aid output power in dB SPL per frequency for a flat-spectrum input signal of 60 dBA.

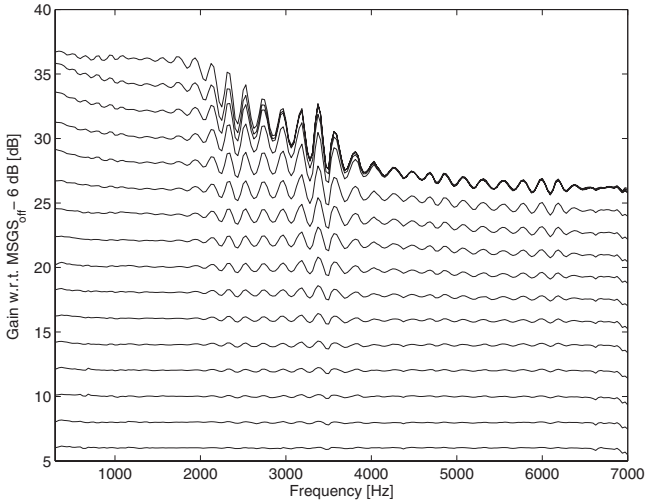


FIG. 4. Gain (as a function of frequency) with respect to $MSG_{off}-6$ dB for increasing gain settings from MSG_{off} to the maximum gain setting in steps of 2 dB for hearing aid A with the closed fitting and the white noise signal in the normal condition.

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 that would be obtained in the same acoustic scenario but then in the absence of acoustic feedback (reference signal $r[k]$).

A. Feedback-free reference signal

In black-box hearing aid systems where there is no access to the hearing aid microphone and receiver signal, the hearing aid output (as measured by the in-the-ear microphone) at a gain far below instability and with the feedback canceller disabled is typically used as a reference signal (Merks *et al.*, 2006; Shin *et al.*, 2007). 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 nonlinear processing such as dynamic range compression and frequency-dependent gain limitation. In addition, at high gains, the hearing aid receiver may become non-linear. In this paper, an alternative procedure for estimating the reference signal is proposed that does not assume linear system behavior. The hearing aid output with the feedback canceller disabled is recorded at the same gain as the actual hearing aid output but with a closed instead of an open fitting (Spriet *et al.*, 2009). For the closed fitting, a temporary foam earmold E-A-RTEMP 13A (EARtone) was used. The difference in frequency characteristic of the ear canal due to the closed fitting is compensated for by means of a finite impulse response (FIR) filter. The FIR filter is determined as the least-squares filter that estimates the hearing aid output with open fitting based on the output with closed fitting at a gain setting of 18 dB below $MSGS_{off}$. For the identification of the FIR filter, a white noise input signal with a presentation level of 70 dBA was used such that the hearing aid output is sufficiently above the environmental and internal noise level. With the closed fitting, the amount of feedback in the recording is minimal. As an illustration, Fig. 4 shows the gain (as a function of frequency) of hearing aid A with the closed fitting with respect to a gain setting of 6 dB below $MSGS_{off}$ for increasing gain settings (in steps of 2 dB) from $MSGS_{off}$ to the maximum gain setting. The results are shown for the white noise signal in the 'Normal' condition. The presence of feedback is limited: at the highest gain setting, only small oscillations of 2 dB occur around 3.5 kHz.

B. Performance measures

This section defines physical measures for detecting instability, oscillations and signal distortion in the hearing aid output, which are applicable to spectrally colored input signals. For non-stationary signals (such as the speech and the opera signal), some segments may be more prone to feedback and oscillations than other segments. Therefore, the performance measures were computed using frames of 0.5 s with an overlap of 80% between adjacent frames. To assess the worst case performance, the maximum value of the performance measure over the whole speech and opera signal was then used. The signals $u[k]$ and $r[k]$ were downsampled to 16 kHz before computing the performance measures.

The performance measures should not be affected by differences in the environmental and internal noise $n[k]$ between the recordings of $u[k]$ and $r[k]$. To estimate the environmental and internal noise, the in-the-ear microphone signal was recorded in silence with the hearing aid gain setting set to 6 dB below $MSGS_{off}$ [3] and with the feedback canceller disabled. The noise signal was rescaled to compensate for the gain difference between $MSGS_{off}-6$ dB and the actual gain setting. To reduce the effect of differences in the environmental and internal noise component in $u[k]$ and $r[k]$, the performance measures were only computed when the short-term energy of the reference signal $r[k]$ exceeded the environmental and internal noise energy by at least 10 dB.

The short term PSD of a signal $u[k]$ is defined as $P_u(f,k)$ where k is the time index and f is the frequency index. The short-term PSD is computed using Welch's method. Each frame of 0.5 s is segmented in half-

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overlapping sections of 256 samples at 16 kHz. Each section is filtered with a Von Hann window before being transformed to the FFT-domain. The short-term PSD is obtained as the averaged square amplitude of the FFTs.

1. Detection of acoustic feedback

a. Output-to-reference signal energy ratio. A simple method to detect instability is to track the short-term output-to-reference signal energy ratio $E(k)$ (Freed, 2008)

$$E(k) = 10\log_{10}\frac{\sum_{i=k-L/2}^{k+L/2-1} u^2[i]}{\sum_{i=k-L/2}^{k+L/2-1} r^2[i]}. \quad (1)$$

L is the window length used in the energy computation. L was set to 8000 samples at 16 kHz. Instability is said to occur if the energy ratio exceeds a certain threshold, e.g., 6 dB. The energy ratio $E(k)$ detects an increase in the output signal level caused by feedback. However, below instability, it does not give any information about the amount of residual feedback.

b. Feedback-to-reference signal energy ratio. To quantify the amount of feedback, the short-term feedback-to-reference signal energy ratio $\mathrm{FSR}(k)$ is computed as (Grimm and Hohmann, 2006; Spriet *et al.*, 2008):

$$\mathrm{FSR}(k) = 10\log_{10}\frac{\sum_{i=k-L/2}^{k+L/2-1} (u[i]-r[i])^2}{\sum_{i=k-L/2}^{k+L/2-1} r^2[i]}, \quad (2)$$

with L again set to 8000 samples.

In addition, the feedback-to-reference signal energy ratio is analyzed in critical bands (Moore, 2003) by means of the short-term intelligibility weighted feedback-to-reference signal energy ratio $\mathrm{FSR}_{\mathrm{intellig}}(k)$:

$$\mathrm{FSR}_{\mathrm{intellig}}(k) = \sum_i w[i]\mathrm{FSR}_i, \quad (3)$$

with FSR_i the feedback-to-reference signal energy ratio in the ith auditory critical band:

$$\mathrm{FSR}_i = 10\log_{10}\frac{\max\left(\int_{f\in B_i} P_v(f,k)df, \beta\int_{f\in B_i} P_n(f)df\right)}{\max\left(\int_{f\in B_i} P_r(f,k)df, \beta\int_{f\in B_i} P_n(f)df\right)}. \quad (4)$$

$P_v(f,k)$ is the short-term PSD of the feedback signal $v[k]=u[k]-r[k]$, $P_r(f,k)$ is the short-term PSD of $r[k]$ and $P_n(f)$ is the long-term PSD of $n[k]$. To limit the impact of environmental and internal noise on the measurements, the feedback and the reference signal energy within each band are lower bounded by β times the noise energy (with $\beta=2$).

The critical bands were defined according to the ANSI S.3.5–1997 standard (17 equally contributing bands between 300 Hz and 6400 Hz) (ANSI S3.5, 1997). The weight $w[i]$ equals 0.06 for each auditory critical band B_i between 300 Hz and 6400 Hz. Since the bandwidth of the auditory critical bands increases with frequency, a narrowband component in a high frequency band gets a lower weight than a narrowband component in a low frequency band. As a result, a high frequency oscillation is weighted less in $\mathrm{FSR}_{\mathrm{intellig}}$ than a low frequency oscillation. In addition, oscillations above 6400 Hz will not be detected.

2. Detection of oscillations

Even before instability occurs, oscillations can already be perceived. Shin *et al.* (2007) and Freed and Soli (2006) defined physical criteria to detect oscillations, referred to as the hearing aid transfer function variation criterion (TVC) and the power concentration ratio (PCR), respectively. These 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 colored input signals and are robust to spectral peaks in the hearing aid frequency response (Spriet *et al.*, 2008, 2009).

a. Transfer function variation criterion (TVC). In Shin *et al.* (2007), the hearing aid transfer function is measured for increasing gain values and a white noise input signal. The hearing aid transfer function in the initial stable condition 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 referred to as the transfer function variation function (TVF). The transfer function variation criterion (TVC) is then defined as

$$\mathrm{TVC}(k) = \max_f(|\mathrm{TVF}(f,k)|). \quad (5)$$

For spectrally colored input signals, an estimate of $\mathrm{TVF}(f,k)$ is obtained as

$$\mathrm{TVF}(f,k) = 10\log_{10}\left(\frac{P_u(f,k)}{P_r(f,k)}\right), \quad (6)$$

which may then be used in (5) (Spriet *et al.*, 2008, 2009).

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

$$P_u(f,k) = \max(P_u(f,k), \alpha P_n(f)),$$

$$P_r(f,k) = \max(P_r(f,k), \alpha P_n(f)), \quad (7)$$

where $P_n(f)$ is the long-term PSD of the internal and environmental noise $n[k]$ and $\alpha>1$. Here, $\alpha=10$ and hence only frequencies where the energy exceeds the noise by at least 10 dB are considered.

From Fig. 1, it can be shown that the transfer function variation function $\mathrm{TVF}(f,k)$ corresponds to an estimate of

$$10\log_{10}\left(\left|\frac{1}{1-G(f,k)(F(f,k)-\hat{F}(f,k))}\right|^2\right), \quad (8)$$

with $G(f,k)$ the frequency response of the hearing aid signal processing path and $F(f,k)$ and $\hat{F}(f,k)$ the frequency response of the feedback path and the feedback canceller, respectively Hence, $\mathrm{TVF}(f,k)$ is related to the loop gain $G(f,k)(F(f,k)-\hat{F}(f,k))$. For example, for a loop gain of 0.9 (which is close to instability), the TVC equals 20 dB.

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The TVC detects the largest peak or dip in the transfer function variation function. However, it does not take into account the input excitation power 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.

b. Power concentration ratio. In Freed and Soli (2006), the power concentration ratio is introduced for detecting oscillations and defined as the degree to which a large amount of power is concentrated at a small number of frequencies in the hearing aid output. The measure, however, assumes that the hearing aid input signal is white. For spectrally colored input signals, two modified measures based on the PCR are presented here.

The first measure is referred to as the *difference* $\Delta\mathrm{PCR}(k)$ in power concentration ratio between the hearing aid output and the reference signal (Spriet *et al.*, 2008, 2009).

1. First, the oscillation frequencies f_c are detected as the frequencies where the transfer function variation function $\mathrm{TVF}(f,k)$ (see 6) is at least 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 dependence 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 $\Delta\mathrm{PCR}(k)$

$$\Delta\mathrm{PCR}(k) = \mathrm{PCR}_u(k) - \mathrm{PCR}_r(k) \tag{9}$$

is then used as a measure for the presence of oscillations. Caution should be used when the power concentration $\mathrm{PCR}_r(k)$ of the reference signal approaches one (i.e., when the reference signal has most of its energy at the oscillation frequencies). In this case, $\Delta\mathrm{PCR}(k)$ approaches zero, even when oscillations occur.

It is to be noted that oscillation frequencies with a large signal power $P_u(f,k)$ result in a larger difference $\Delta\mathrm{PCR}(k)$ than oscillation frequencies with a low signal power. As a result, weak oscillations may not always be detected by $\Delta\mathrm{PCR}(k)$.

A second measure is referred to as the *normalized power concentration ratio* $\mathrm{PCR}_n(k)$ and computes the PCR of Freed and Soli (2006) on the normalized PSD $P_u(f,k)/P_r(f,k)$. First, the oscillation frequencies are determined based on the transfer function variation function (see step 1 above). The fraction of the normalized power $P_u(f,k)/P_r(f,k)$ that is located at the five (or less) strongest oscillation frequencies is referred to as $\mathrm{PCR}_n(k)$. The normalized power concentration ratio does not take into account the input excitation power of the oscillation frequencies: all frequencies are equally weighted.

The modified TVC and PCR measures are computed on the frequency range from 500 Hz to 6500Hz. Outside this frequency range, the hearing aid output power for an open fitting is low (see Fig. 3) and hence, susceptible to noise.

3. Detection of signal distortion

To assess the distortion of the hearing aid output $u[k]$, the frequency-weighted log-spectral signal distortion $\mathrm{SD}(k)$ is defined as (Leijon, 2005)

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

The weighting function $w(f)$ gives equal weight to each auditory critical band between f_l=300 Hz and f_h=6400 Hz. The function $w(f)$ was derived through linear interpolation of the ratio of the critical band weights $w[i]$ (see (3)) and the critical band bandwidths and was normalized such that

$$\int_{f_l}^{f_h} w(f)df = 1.$$

The short-term PSD $P_u(f,k)$ and $P_r(f,k)$ are constrained as in (7) so that only frequency bins with energy level above the environmental and internal noise level are taken into account.

C. Evaluation procedure

1. Measurement procedure

For each hearing aid and acoustic scenario (i.e., 'Normal' or 'Handset'), the following measurement procedure was adopted.

1. Use the open fitting.
2. Present the white noise signal at 60 dBA. Determine $\mathrm{MSGS}_{\mathrm{off}}$ as the maximum gain setting for which no instability occurs, i.e., $\max_k\{E(k)\}<6$ dB (see Sec. III C 2).
3. Present the white noise signal at 70 dBA. Record the in-the-ear microphone signal with the hearing aid switched off and the in-the-ear microphone signal with the hearing aid switched on and the feedback canceller disabled at a gain setting of 18 dB below $\mathrm{MSGS}_{\mathrm{off}}$ (see Sec. III A).
4. Repeat the following procedure for each test signal. Present the test signal at 60 dBA. Record the in-the-ear microphone signal with the hearing aid switched off. Record the in-the-ear microphone signal with the hearing aid switched on and the feedback canceller disabled for increasing gain settings from $\mathrm{MSGS}_{\mathrm{off}}-6$ dB to the maximum gain setting and for decreasing gain settings from the maximum gain setting to $\mathrm{MSGS}_{\mathrm{off}}-6$ dB. Repeat the same measurements but now with the feedback canceller enabled.
5. Replace the open fitting with the closed fitting. Repeat step 3.
6. Repeat the following procedure for each test signal. Present the test signal at 60 dBA. Record the in-the-ear microphone signal with the hearing aid switched off. Record the in-the-ear microphone signal with the hearing aid switched on and the feedback canceller disabled for increasing gain settings from $\mathrm{MSGS}_{\mathrm{off}}-6$ dB to the maximum gain setting.

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2. Added stable gain

The added stable gain (ASG) is defined as the difference between the maximum stable gain with the feedback canceller enabled (MSG_{on}) and the maximum stable gain with the feedback canceller disabled (MSG_{off}):

$$ASG = MSG_{on} - MSG_{off}. \quad (11)$$

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 (Freed and Soli, 2006). The step size of the gain control was 1 dB for A, B, C, and E/E-s and 2 dB for D. At each gain setting, the hearing aid output was computed 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 path component). To avoid convergence effects, the signal was presented once before each recording was made. In this paper, the output-to-reference signal energy ratio is used as a criterion for instability. For each gain setting, the maximum short-term output-to-reference signal energy ratio $\max_k\{E(k)\}$ over the whole signal segment was computed. The maximum gain setting for which $\max_k\{E(k)\}$ remained below 6 dB was determined (i.e., MSGS). The gain at MSGS 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 (i.e., a gain setting of 6 dB below MSG_{off}).

To compare the output level of the different hearing aids, the sound pressure level of the reference signals with the different hearing aids at MSG_{off} was computed.

3. Performance measures at gains MSG_{off}, MSG_{off} +6 dB and MSG_{on}

A high ASG does not necessarily guarantee a high feedback suppression and no signal distortion at all gain settings: at gain settings with a small stability margin, oscillations and distortion may already be perceived. Therefore, the maximum amount of feedback ($E(k)$, $FSR(k)$ and $FSR_{intellig}(k)$), oscillations ($TVC(k)$, $\Delta PCR(k)$ and $PCR_n(k)$) and distortion ($SD(k)$) for the speech and the opera signal was also determined at MSG_{off}, MSG_{on} and at the intermediate gain setting that corresponded to an actual gain of MSG_{off}+6 dB. The intermediate gain was only considered when smaller than MSG_{on}. The performance measures of the ascending and the descending protocol were averaged.

4. Tracking performance

In addition to a high feedback reduction in static conditions, the feedback reduction algorithm should be able to quickly adjust to feedback path changes so that the possible duration of instability is minimal. To check the tracking performance of the feedback cancellers, a dynamic feedback path was created by turning the head in such a way that the left ear approaches an object. An office lamp was positioned close to the left ear of the artificial head (see Fig. 5). In the start position, the left ear of the artificial head was turned away from the lamp by 60 degrees (clockwise rotation). 50 s of stationary HINT noise was presented with a level of 60 dBA at the center of the head. After 20 s of signal presentation, the artificial head was rotated by −60 degrees using the Animatics SmartMotor. The maximum velocity and maximum acceleration of the rotation were equal to 100 deg/s and 144 degrees/sec^2. The in-the-ear microphone signal was recorded with the feedback canceller enabled. To compute the hearing aid output only, the direct path contribution was measured as the in-the-ear signal with the hearing aid switched off and subtracted from the in-the-ear microphone signal. The hearing aid gain setting was fixed to 6 dB above $MSGS_{off}$ in the end position. Note that this does not necessarily mean that the actual gain of the hearing aid was 6 dB above MSG_{off}. For A, C, D, E and E-s, the 6 dB increase in gain setting indeed corresponds to a 6 dB increase in actual gain. However, in hearing aid B, the increase in actual gain is smaller at high frequencies due to the gain limitation.



FIG. 5. Set-up of the tracking experiment. An office lamp is positioned close to the left ear of the artificial head.

The performance measures $E(k)$, $FSR(k)$, $FSR_{intellig}(k)$, $TVC(k)$, $\Delta PCR(k)$ and $PCR_n(k)$ were computed from the hearing aid output signals using overlapping windows of 0.2 s with an overlap of 80%. A window length of 0.2 s instead of 0.5 s was used in order to increase the temporal resolution. This, however, reduces the accuracy of the PSD estimates in the performance measures. The duration of instability is defined as the time difference between the middle of the last and the first window where the performance measure exceeds a certain threshold. Hence, it is determined by values of the performance measures above the threshold of instability. The thresholds were set sufficiently high such that the duration was not affected by the reduced accuracy of the PSD estimates. Table II indicates the threshold values that

TABLE II. Thresholds of instability that are used to detect instability in the tracking experiment.

E	FSR	$FSR_{intellig}$	TVC	ΔPCR	PCR_n
6 dB	0 dB	−6 dB	20 dB	0.5	0.5

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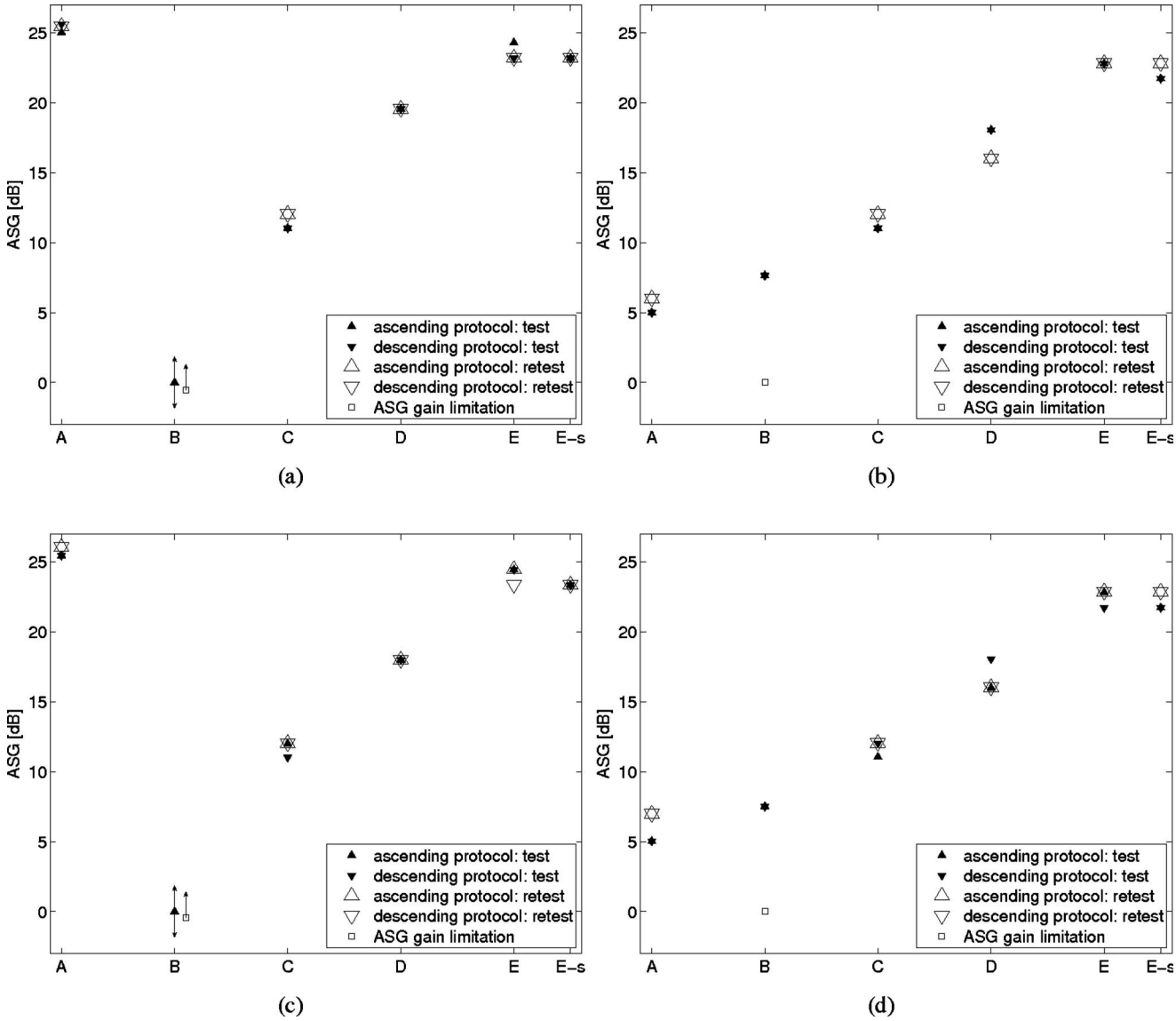


FIG. 6. Added stable gain for [(a) and (b)] the white noise and [(c) and (d)] the HINT noise signal in [(a) and (c)] the normal condition and [(b) and (d)] the handset condition. Up-pointing triangulars show the ASG of the ascending protocol; down-pointing triangulars show the ASG of the descending protocol. 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.

were used for the different measures. If stability was not achieved within 30 s after the change, the duration was set to infinity (∞). To check reproducibility, three test runs were performed for each hearing aid. Between the test runs, the fitting was removed and reconnected to the hearing aid and the hearing aid was repositioned on the artificial head. Within one test run, two recordings of the same condition were made. As a reference signal, the hearing aid output with a closed fitting was recorded. The difference in frequency characteristic of the in-the-ear microphone signal when using an open versus a closed fitting varies with the angle over which the head is rotated. Therefore, the closed fitting reference signals were transformed to open fitting reference signals by means of a time-varying FIR filter. The time-varying filter was determined as the exponentially weighted recursive-least-squares (RLS) filter (with forgetting factor 0.999) (Haykin, 2001) that estimates the hearing aid output with open fitting based on the output with closed fitting at a gain far below instability (i.e., 18 dB below $MSGS_{off}$ in the end position). For the identification of the time-varying FIR filter, a HINT noise input signal with a presentation level of 70 dBA was used.

IV. RESULTS

A. Added stable gain

The ASG of the feedback reduction techniques was measured for the four input signals in the two conditions 'Normal' and 'Handset'. Figure 6 depicts the ASG for the white noise and the HINT noise signal. Figure 7 depicts the ASG for the speech signal and the opera signal. The up-pointing triangulars show the ASG defined by the ascending protocol. The down-pointing triangulars show the ASG defined by the descending protocol. For hearing aids A, C, D and E/E-s retest data are also shown.[4] For the opera signal, the ASG for the music mode of hearing aids B, C and D is also depicted (referred to as B-m, C-m, D-m). For hearing aid B, the ASG that was obtained with the gain limitation only (without feed-

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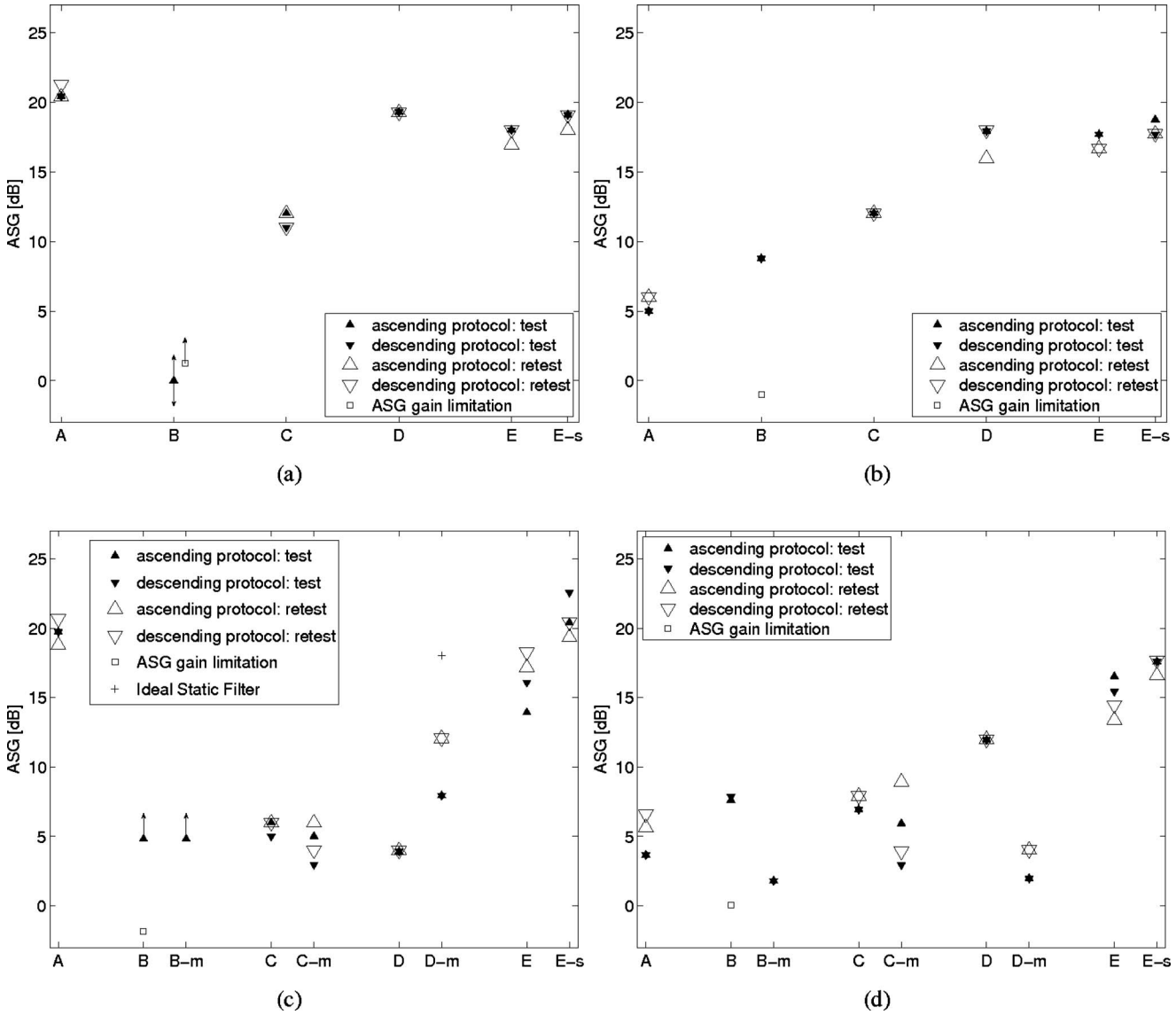


FIG. 7. Added stable gain for [(a) and (b)] the speech signal and [(c) and (d)] the opera signal in [(a) and (c)] the normal condition and [(b) and (d)] the handset condition. Up-pointing triangulars show the ASG of the ascending protocol; down-pointing triangulars show the ASG of the descending protocol. 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.

back cancellation) is depicted with squares. This ASG was determined as the difference between MSG_{off} with feedback path initialization and MSG_{off} without feedback path initialization ('No Init').

To compare the output level of the hearing aids, Table III depicts the sound pressure level (in dB SPL) of the feedback-free reference signal for the different hearing aids at MSG_{off}. In the 'Normal' condition, the largest level difference between hearing aids A, B ('No Init'), C, D and E for a given sound signal is 4 dB (i.e., the level difference between

TABLE III. Sound pressure level (dB SPL) of the reference signal at MSG_{off} for the four input signals and the two acoustic conditions. The values separated by slashes for hearing aids A, C, D, and E represent test and retest data. In the handset condition, instability occurs at a 10–16 dB lower gain.

	Normal condition				Handset condition			
HA	White	HINT	Opera	Speech	White	HINT	Opera	Speech
A	87/87	86/85	87/86	86/86	74/78	75/77	74/77	74/79
B	>90	>92	>93	87	80	80	77	77
B ('No Init')	91	90	90	87	80	80	78	77
C	90/89	88/87	88/87	86/86	77/75	76/75	73/72	75/72
D	89/89	87/87	87/87	88/89	77/78	76/78	73/74	76/74
E	91/91	89/89	89/89	90/90	74/74	73/74	72/72	73/74

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A and E for the white noise, the HINT noise and the speech signal). In the 'Handset' condition, the largest level difference is 7 dB (i.e., the level difference between A and C for the speech signal). In the 'Handset' condition, MSG_{off} is 10 dB to 16 dB lower compared to the 'Normal' condition. In the 'Normal' condition, differences in the sound pressure level between test and re-test are within 1 dB. In the 'Handset' condition, larger level differences (i.e., up to 5 dB) between test and re-test occur most probably due to minor changes in the positioning of the handset.

Due to the gain limitation, instability was never reached with hearing aid B in the 'Normal' condition, even when the feedback canceller was disabled, except for the opera signal. For the opera signal only, MSG_{off} could be determined. With feedback canceller enabled, instability was never reached too. As a result, the ASG that is offered by the feedback canceller (depicted with up-pointing triangulars) could not be determined. This is indicated in Figs. 6 and 7 with the double facing arrow in case of the white noise, HINT noise and speech signal and the upward-pointing arrow in case of the opera signal. In addition, for the white noise, HINT noise and the speech signal, MSG_{off} with feedback path initialization could not be determined. Hence, the depicted ASG that is obtained with the gain limitation only (depicted by squares) is a lower bound (which is indicated by the upward-pointing arrow).

In the 'Handset' condition, gain limitation occurred in hearing aid B at frequencies above 4 kHz for gains of 4 dB and more above MSG_{off}. Hence, except for the music mode, the gain above 4kHz was limited at MSG_{on}. 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.

Specific observations from these data include:

- For the white noise and the HINT noise signal, ASGs between 11 dB and 26 dB were obtained for the 'Normal' condition in hearing aids A, C, D and E. 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 26 dB for the 'Normal' condition to 5–7 dB for the 'Handset' condition. The feedback path in the 'Handset' condition largely differs from the feedback path in the 'Normal' condition (Stinson and Daigle, 2004) and hence, from the starting point of the feedback canceller in hearing aid A. The degraded feedback reduction for the 'Handset' condition indicates that the feedback canceller in hearing aid A can only model small deviations from its starting point. The ASG for the white noise and the HINT noise signal are similar.
- For the speech signal, the ASG ranges from 11 dB to 21 dB for the 'Normal' condition across hearing aids A, C, D and E and from 5 dB to 19 dB for the 'Handset' condition across hearing aids A, B, C, D and E. The ASGs are not much lower than the ASGs for the white noise and the HINT noise signal: the largest degradation (about 5 dB) occurs for A in the 'Normal' condition and E/E-s in the 'Normal' and the 'Handset' condition. Hence, the ASG of the feedback cancellers is not seriously affected by the speech signal. For a speech signal, a hearing aid processing delay between 4 and 7 ms (see Table I) significantly reduces the correlation between the input to the feedback canceller and the desired hearing aid input signal, which may explain why the performance is not seriously affected. For hearing aids A and E, the ASG values for the speech signal are about 5 dB lower than for the HINT noise and the white noise. For hearing aid A, compression occurs at high gains (i.e., gains of 11 dB or more above MSG_{off}), in particular at high frequencies (i.e., above 2.5 kHz). As a result, the stationary HINT and white noise signal are amplified less than the low energy speech segments, which may explain the higher ASG for the stationary signals. Unlike in hearing aids A, B, C and D, compression and output limiting is not applied in system E/E-s. Hence, at high gains, non-linearities occur in system E/E-s due to saturation of the hearing aid amplifier and receiver. The speech signal contains segments with a higher energy than the stationary signals. During these segments more non-linearities occur, which result in a lower ASG.
- For the opera signal, the ASG ranges from 3 dB to 22 dB for the 'Normal' condition and from 2 dB to 18 dB for the 'Handset' condition. Hearing aids C, D and E achieve a lower ASG for the opera signal than for the speech signal and the stationary signals. Except for hearing aid D in the 'Normal' condition, the music modes B-m, C-m and D-m do not increase the ASG. Thanks to the exploitation of an initial feedback path measurement in the adaptive feedback canceller, hearing aid A achieves a high ASG for the opera signal in the 'Normal' condition. System E-s still achieves a high ASG for the opera signal. However, oscillations already occur at gains below MSG_{on} (see Sec. IV B).

The differences between ASGs obtained with test and retest data and with the ascending and descending protocol are in general limited to 1 dB to 2 dB, which corresponds to the

TABLE IV. Maximum feedback-to-reference signal ratio (max{FSR(k)}) (in dB) in hearing aid systems A, C, D, and E at the gains MSG_{off}, MSG_{off}+6 dB, and MSG_{on} for the speech signal in the normal condition for test and retest (test/retest). Values above 0 dB are highlighted in bold font.

FC	Gain	A	C	D	E	E-s
Off	MSG_{off}	**1.7**/−0.3	**2.5/1.9**	−1.1/−1.1	**2.8/4.3**	**4.5/5.7**
On	MSG_{off}	−10.2/−9.8	−3.6/−3.4	−4.7/−4.3	−11.2/−10.5	−11.1/−10.9
On	+6 dB	−9.8/−10.3	−1.6/−1.6	−3.1/−3.6	−10.5/−10.0	−11.2/−10.7
On	MSG_{on}	**3.8/3.7**	**2.4/3.2**	**1.0/0.5**	**2.5/1.5**	**3.2/1.7**

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TABLE V. Maximum feedback-to-reference signal ratio (max{FSR(k)}) (in dB) in hearing aid systems A, B, C, D, and E at the gains MSG_{off}, MSG_{off}+6 dB, and MSG_{on} for the speech signal in the handset condition for test and retest (test/retest). Values above 0 dB are highlighted in bold font.

FC	Gain	A	B	C	D	E	E-s
Off	MSG_{off}	−2.3/**0.6**	−0.1	**1.4/1.6**	−2.4/−1.1	**3.6/3.9**	**3.6/3.9**
On	MSG_{off}	−4.2/−8.1	−9.0	−4.5/−3.1	−6.1/−7.0	−4.5/−6.8	−5.0/−6.8
On	+6 dB	—	−7.5	−3.2/−2.1	−7.7/−7.6	−7.7/−8.1	−7.7/−8.0
On	MSG_{on}	−0.4/**0.2**	**1.6**	**2.5/3.0**	**1.6/1.6**	**2.7/2.7**	**0.8/3.4**

step size of the gain control. For the opera signal, larger differences sometimes occur, in particular for C-m, D-m and E. For the adaptive feedback cancellers, small differences in the filter coefficients of the feedback canceller may occur from measurement to measurement (e.g., due to a different realization of the environmental and internal noise or different initial filter coefficients of the adaptive feedback canceller). However, when the hearing aid with the feedback canceller enabled operates close to instability, a small change in filter coefficients may result in a large change in the output signal level and hence $E(k)$. The larger differences between ASGs obtained with test and retest data and with the ascending and descending protocol for C-m and E may be due to hearing aid operation close to instability at intermediate gains between MSG_{off} and MSG_{on}. For the static feedback canceller D-m, the differences between ASGs with test and retest data in the 'Normal' condition is 4 dB. A small deviation between the initialized and the actual feedback path may have a large impact on the feedback reduction performance. After initialization of the feedback path and hence, the static feedback canceller D-m, the earmold was removed and 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.

B. Performance at gains MSG_{off}, MSG_{off}+6 dB and MSG_{on}

Given the large amount of data, only the major results are shown.

The amount of feedback reduction is illustrated by means of the maximum feedback-to-reference signal energy ratio $\max_k\{FSR(k)\}$. In contrast to $\max_k\{E(k)\}$, $\max_k\{FSR(k)\}$ not only indicates instability, but also gives information about the amount of residual feedback below instability. In addition, it was found more appropriate for the detection of feedback than the intelligibility-weighted measure $\max_k\{FSR_{intellig}(k)\}$ (as explained later). Tables IV and V show $\max_k\{FSR(k)\}$ for the speech signal the 'Normal' condition and the 'Handset' condition at the gains MSG_{off}, MSG_{on} and the intermediate gain MSG_{off}+6 dB. The performance of hearing aid B for the speech signal in the 'Normal' condition is not shown because MSG_{off} has never been reached. Tables VI and VII depict $\max_k\{FSR(k)\}$ for the opera signal in the 'Normal' and the 'Handset' condition, respectively. $FSR(k)$ above 0 dB indicates that a substantial amount of feedback is present.

The presence of oscillations is illustrated by means of $\max_k\{\Delta PCR\}$: values above 0.3 may indicate that oscillations are present. Similar results were generally obtained with $\max_k\{TVC(k)\}$ and $\max_k\{PCR_n\}$. Figure 8 graphically represents $\max_k\{\Delta PCR\}$ of the feedback cancellers in A, B, C, D and E at the gain MSG_{off} for the opera signal. The abbreviations A-off, B-off, C-off, D-off and E-off refer to the hearing aids with the feedback canceller disabled. The performance measures with the ascending and the descending protocol were averaged. For hearing aids A, C, D, E and E-s, both test and re-test data are provided.

It should be noted that these measures only give an indication of the physical performance of the feedback reduction techniques. Listening tests are required to measure the perceptual relevance.

- For the speech signal, all feedback cancellers achieve good performance below MSG_{on} in both conditions: $\max_k\{FSR_{intellig}(k)\}$ and hence, the amount of feedback is reduced compared to the hearing aid output when the feedback canceller is disabled at MSG_{off}. In addition, no oscillations were detected below MSG_{on} when the feedback canceller was enabled ($\Delta PCR < 0.3$ for all hearing aids). In the 'Normal' condition, hearing aids A, E and E-s achieve the best feedback reduction. In the 'Handset' condition, the improvement of hearing aid A is smaller than in the 'Normal' condition.
- For the tonal opera signal, the performance of the feedback cancellers degrades, except for hearing aid A. The performance of hearing aid A is less affected by the tonal signal

TABLE VI. Maximum feedback-to-reference signal ratio (max{FSR(k)}) (in dB) in hearing aid systems A, B, C, D, and E at the gains MSG_{off}, MSG_{off} +6 dB, and MSG_{on} for the opera signal in the normal condition for test and retest (test/retest). Values above 0 dB are highlighted in bold font.

FC	Gain	A	B	B-m	C	C-m	D	D-m	E	E-s
Off	MSG_{off}	**1.0/1.7**	**0.5**	**0.5**	**0.3/0.5**	**0.3/0.5**	**0.1/0.6**	**0.1/0.6**	**2.4/2.5**	**2.4/2.5**
On	MSG_{off}	−13.8/−14.9	−8.7	−6.2	−8.9/−10.3	−9.1/−9.2	**0.1/0.5**	−16.0/−16.7	−4.2/−4.6	−6.1/−5.9
On	+6 dB	−10.8/−10.0	—	—	—	—	—	−12.3/−12.7	−1.8/−3.2	−6.1/−6.0
On	MSG_{on}	**0.0/3.4**	−4.6	−3.8	−5.6/−4.2	−7.6/−7.4	**2.3/3.1**	−10.9/−4.7	**0.7/1.5**	**2.1**/−1.6

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TABLE VII. Maximum feedback-to-reference signal ratio ($\max\{FSR(k)\}$) (in dB) in hearing aid systems A, B, C, D, and E at the gains MSG_{off}, MSG_{off} +6 dB, and MSG_{on} for the opera signal in the handset condition for test and retest (test/retest). Values above 0 dB are highlighted in bold font.

FC	Gain	A	B	B-m	C	C-m	D	D-m	E	E-s
Off	MSG_{off}	−0.8/**0.9**	**1.9**	**1.9**	**1.8/1.6**	**1.8/1.6**	**1.5**/−2.0	**1.5**/−2.0	**1.4/1.0**	**1.4/1.0**
On	MSG_{off}	−5.2/−7.1	−6.6	−1.5	−3.8/−3.5	−3.3/−3.5	−0.7/−1.6	−3.6/−5.8	**0.2**/−0.4	−2.1/**0.1**
On	+6 dB	—	−2.7	—	—		−0.2/−0.0	—	−0.1/−2.1	**1.3/1.9**
On	MSG_{on}	−1.3/**1.0**	**2.5**	**2.9**	**1.6/0.2**	−2.0/−1.8	**1.2/2.4**	**2.2/2.5**	**2.2/2.9**	**0.9/1.0**

thanks to the exploitation of an initial feedback path measurement. For D, the amount of feedback (see $\max_k\{FSR(k)\}$ in Tables VI and VII) and oscillations (see $\max_k\{\Delta PCR\}$ in Fig. 8) are reduced only slightly or even not at all compared to the hearing aid output at MSG_{off} with the feedback canceller disabled in both the 'Normal' and the 'Handset' condition. The same holds for E /E-s in the 'Handset' condition. At MSG_{off}, the feedback reduction systems of hearing aid B and C succeed to reduce the amount of feedback and oscillations in both conditions. However, for C, a smaller ASG is achieved than for the speech signal and for B, less feedback is reduced in the 'Handset' condition at MSG_{off}+6 dB. For hearing aid D, the performance is improved by the music mode D-m. However, the ASG of D-m in the 'Handset' condition is worse due to the static feedback canceller that is not optimal for this condition. The music modes B-m and C-m do not result in an improvement for the opera signal in the 'Normal' and the 'Handset' condition. In the 'Normal' condition, E-s slightly improves $\max_k\{FSR(k)\}$ of hearing aid E for the tonal signal.

Test and retest data are consistent with each other. In the 'Handset' condition, small variations in the positioning of the handset may also result in differences between test and retest data (e.g., hearing aid A at MSG_{off} for the speech signal). In addition, it should be noted that when operating close to instability, a small difference in the estimated feedback path may result in a large difference in the hearing aid output and hence, in a larger difference in the performance measures. This explains why the difference between test and retest data is larger at gains close to instability. In the 'Normal' condition, for example, the largest differences in $\max_k\{FSR(k)\}$ between test and retest generally occur at gains where $\max_k\{FSR(k)\}$ exceeds 0 dB (see for example, A, D, E and E-s at MSG_{on} in Table VI).

In general, the performance measures have a similar behavior: they all degrade with increasing gain and hence, increasing feedback. However, the degradation in one measure is sometimes smaller than the degradation in another measure. This is due to the fact that some measures (i.e, FSR, E, ΔPCR) take into account the signal power at the oscillation frequencies, while others (i.e., TVC and PCR_n) do not and/or use a frequency weighting factor (i.e., SD and $FSR_{intellig}$). For example, for hearing aid E, ΔPCR is relatively small (i.e., below 0.15 at all gains) in contrast to PCR_n (between 0.14 and 0.96) and TVC (between 10 and 30 dB) for the opera signal in the 'Normal' condition because of the low power at the oscillation frequencies with respect to the total signal power. Close to instability, $FSR_{intellig}$ is often much smaller than FSR. Feedback especially occurs at high frequencies (i.e., above 2 kHz). These frequencies are weighted less or not at all (if above 6400 Hz) in $FSR_{intellig}$, making it less appropriate for the detection of feedback.

C. Tracking performance

Table VIII depicts the average duration of instability of the feedback cancellers of the three test runs based on the different performance measures, together with the difference between the maximum and minimum duration (shown in brackets). The duration of instability depends on the used performance measure and the chosen threshold for instability. For all hearing aids, except for C-m, the duration of

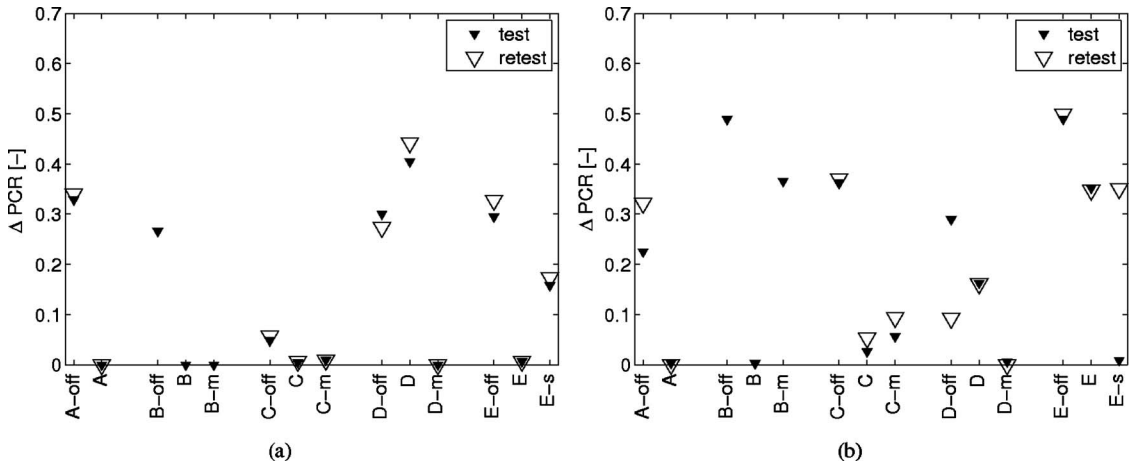


FIG. 8. $\max_k\{\Delta PCR(k)\}$ of the feedback cancellers in the hearing aid systems A, B, C, D, and E at the gain MSG_{off} for the opera signal. (a) Normal condition; (b) handset condition.

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TABLE VIII. Average duration of instability (in seconds) of three test runs based on the performance measures E, FSR, $FSR_{intellig}$, TVC, ΔPCR, and PCR_n. Between brackets the difference between the minimum and maximum duration of the three test runs is shown.

	Duration of instability (s) based on the following measures					
HA	E	FSR	$FSR_{intellig}$	TVC	ΔPCR	PCR_n
A	∞(–)	∞(–)	∞(–)	∞(–)	∞(–)	∞(–)
B	0.3 (0.2)	0.5 (0.2)	0.2 (0.3)	0.3 (0.2)	0.3 (0.3)	0.3 (0.2)
B-m	0.9 (1.0)	1.3 (0.8)	1.1 (0.5)	0.9 (1.3)	0.8 (1.2)	1.0 (1.0)
C	0.8 (0.2)	0.9 (0.2)	0.7 (0.4)	0.8 (0.2)	0.8 (0.3)	0.8 (0.2)
C-m	6.6 (0.1)	11.2 (0.4)	4.0 (1.0)	4.7 (2.0)	6.1 (2.5)	5.4 (1.4)
D	0.0 (0.0)	0.0 (0.0)	0.0 (0.0)	0.0 (0.0)	0.0 (0.0)	0.0 (0.0)
D-m	∞(–)	∞(–)	∞(–)	∞(–)	∞(–)	∞(–)
E	1.4 (1.0)	1.4 (1.0)	1.4 (0.9)	1.4 (1.0)	1.2 (1.4)	1.4 (1.0)
E-s	0.5 (0.2)	0.5 (0.2)	0.5 (0.2)	0.5 (0.2)	0.4 (0.2)	0.5 (0.2)

instability is similar for the different performance measures. Feedback canceller C-m has a slow adaptation speed. As a result, the system operates close to instability for a longer period of time such that the duration of instability is then highly dependent of the criterion used.

Hearing aid D (normal mode) exhibits the best tracking performance: during the feedback path change, no instability is observed. The duration of instability is larger for the music modes B-m, C-m and D-m than for the normal modes B, C and D. The feedback canceller of hearing aid A only achieves an ASG of 1 or 2 dB (depending on the test run) in the end position. As a result, the feedback canceller is not able to track the feedback path change at a gain setting of $MSGS_{off}$+6 dB. The same holds for D-m. The combination of a slowly and a fast adapting filter in hearing aid E-s reduces the duration of instability. Hearing aid E-s outperforms hearing aid A, B-m, C, C-m and D-m and obtains only slightly worse results than hearing aid B. It should however be noted that for hearing aid B, the gain setting $MSGS_{off}$ +6 dB only corresponds to an increase in actual gain of 6 dB at frequencies below 2 kHz. At high frequencies, the increase in gain is smaller due to the gain limitation. As a result, comparison with hearing aid B is difficult as the gain limitation reduces the duration of instability.

Within each test run, two measurements were made. Except for C-m, the differences between these two measurements were within the window length of 0.2 s, which suggests that the rotation of the head by the motor is reproducible. For hearing aid C-m, larger differences (up to 2.5 s) were sometimes found between the two measurements. Due to the slow adaptation, the hearing aid operates close to instability during a larger time period. Close to instability, the variability in the performance measures due to differences in the environmental noise is larger than at gains below instability. The differences between the three different test runs (shown in Table VIII) are generally larger than the differences within test run, in particular for the hearing aids with slow adaptation (i.e., B-m, C-m and E). Close to instability, small changes in the dynamic feedback path may result in large differences in the output signal and hence, the input to the adaptive feedback canceller. Between each run, the hearing aid and the fitting were removed and replaced on the artificial head. This may cause small differences in the feedback path, in particular when the ear is in the near vicinity of an object. As an illustration, Fig. 9 depicts the amplitude frequency response of the feedback path of hearing aid E in the start and end position of the artificial head for the three runs. In the start position, the three feedback paths are very similar. However, in the end position, the differences between the acoustic feedback paths are larger (more than 5 dB) at frequencies above 3kHz (in particular, the feedback path of the third run). This is due to a different positioning of the fitting and the hearing aid and/or a slight change in the positioning of the lamp with respect to the artificial head.

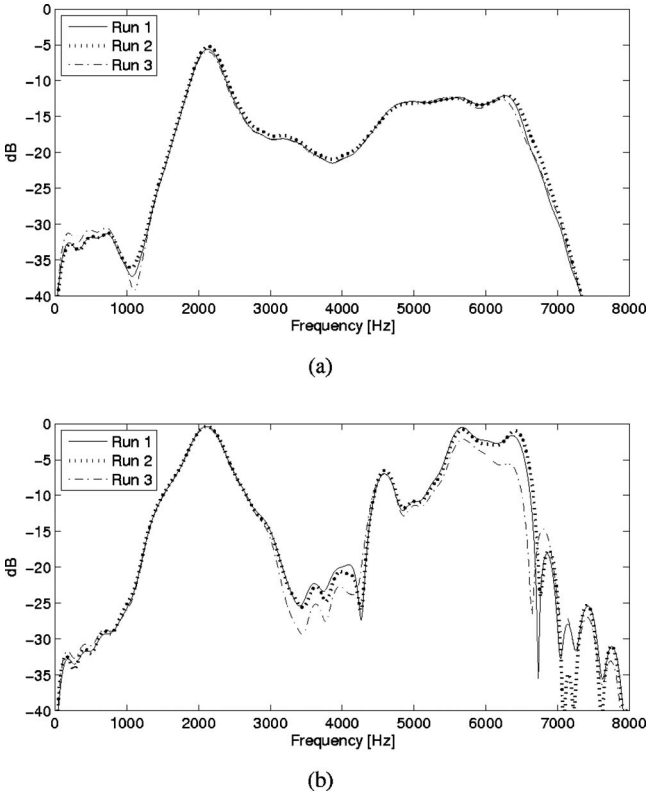


FIG. 9. Amplitude frequency response of the feedback path of hearing aid E in the (a) start and (b) end position for the three test runs.

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V. DISCUSSION

A. Algorithm performance

The performance comparison of feedback reduction systems in hearing aids is complicated by the coarse controls in the fitting software. The frequency characteristics of the hearing aids cannot be perfectly matched and compression can often not be completely disabled. However, from the physical evaluation, the strengths and the weaknesses of the algorithms can be identified. Thanks to the exploitation of an initial feedback path measurement in the feedback canceller, the feedback reduction of hearing aid A is not seriously affected by a tonal input signal. However, this comes at the expense of a worse tracking performance and a worse feedback reduction in acoustic conditions that differ from the condition for which the initialization was performed. Hearing aids D and E offer a high ASG in the 'Normal' condition as well as in the 'Handset' condition. For D and E-s, a good tracking performance is also obtained. However, the feedback reduction for tonal input signals could be improved. Hearing aid B and C offer a moderate ASG (around 10 dB), but are less sensitive to tonal input signals than D and E. However, the high frequency gain in B is strongly limited. This may degrade audibility and complicates the comparison with other hearing aids. The tracking performance of C (in particular the music mode C-m) is poor. No improvement was observed for music mode B-m and C-m of the feedback reduction systems in hearing aids B and C, which use a reduced adaptation speed. In the 'Normal' condition, music mode D-m improves the performance of hearing aid D thanks to the static feedback canceller.

B. Performance measures

In the literature, the ASG is generally computed for a white noise input signal. Measurements of the ASG for the four different input signals however show that the ASG depends on the input signal. Especially for tonal input signals, the ASG may be seriously affected and even below MSG_{on}, oscillations may occur. Previous studies often used the hearing aid gain setting to derive the ASG (Merks *et al.*, 2006; Freed and Soli, 2006; Shin *et al.*, 2007). However, some hearing aids sacrifice gain in the high frequencies, e.g, hearing aid B. If the ASG of hearing aid B were derived based on the hearing aid gain setting, the ASG would be overestimated.

The performance measures are based on a comparison with a feedback-free reference signal. The reference signal for the black-box hearing aids is created by replacing an open fitting by a closed fitting, with appropriate compensation. However, some types of hearing aids are only used in combination with an open fitting, e.g., receiver-in-the-canal or receiver-in-the-ear hearing aids. For the generation of the reference signal, appropriate sealing of the ear canal is then required. The feedback reduction systems that were evaluated in this paper did not modify the hearing aid signal processing path that is applied when the feedback canceller is disabled. Some feedback reduction systems however reduce the gain at the oscillation frequencies (cf. notch filtering) or apply non-linearities such as frequency shifting and phase modifications. Since these modifications are not present in the reference signal, they may have an impact on the physical measures.

The performance measures give a first indication of the algorithm's performance. The energy ratio $E(k)$ is a useful measure to detect a strong amount of feedback or instability. The measure is however dominated by the frequencies with the highest output power. As a result, feedback and oscillations at frequencies with little energy may not always be detected. Below instability, $FSR(k)$ gives more information on the residual amount of feedback and hence, the feedback reduction by the feedback canceller. In $E(k)$ and $\Delta PCR(k)$, oscillations at frequencies with high energy will result in larger measures than oscillations at frequencies with little energy. $TVC(k)$ and $PCR_n(k)$ do not take into account the signal power of the oscillation frequencies. Compared to $\Delta PCR(k)$, oscillations at frequencies with little energy will be better detected. However, these oscillations may be masked by more powerful frequencies. In addition, $TVC(k)$ and $PCR_n(k)$ are more susceptible to environment noise and should only be applied in the frequency range where the hearing aid output to environment and internal noise ratio is high. In $FSR_{intellig}(k)$ and $SD(k)$, a frequency weighting is applied according to intelligibility. Since the critical bandwidth increases with increasing frequency, oscillations at high frequencies (above 2 kHz) where feedback most often occur, are weighted less compared to oscillations at lower frequencies. In addition, oscillations above 6400 Hz are not detected. As a result, these measures are generally less degraded by the presence of high-frequency feedback than $FSR(k)$. Further research is necessary to measure the perceptual relevance of the physical measures for the hearing aid user. The perceptual relevance may be improved through the development of psycho-acoustically motivated measures that take into account models for loudness and masking as well as the users' hearing loss.

C. Repeatability

Repeated measures show that the reproducibility of the test set-up is crucial. If the hearing aid is located in the near vicinity of an object, minor changes in the set-up may cause significant feedback path differences at higher frequencies. In particular, when operating close to instability, a small change in the set-up (e.g., position of the earmold) or in the feedback canceller may result in a large difference in the hearing aid output. This was, for example, observed in the ASG measurement of C-m and E for the opera signal and in the assessment of the tracking performance of B-m, C-m and E.

VI. CONCLUSIONS

In this paper, the performance of the feedback reduction techniques in four commercial hearing aids and two implementations of a recently developed feedback cancellation technique were assessed based on physical performance measures. The ASG was determined for white noise, HINT noise, a speech signal and an opera signal in two static acoustic conditions with and without a handset positioned. In

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addition, the reduction of feedback, oscillations and distortion at gain values below the maximum stable gain were assessed and the ability of the feedback cancellers to track feedback path changes was measured. The ASG ranges from 3 dB to 26 dB and varies widely across different hearing aids. The ASG is observed to depend on the input signal: the ASG for a tonal input signal may be considerably smaller than the ASG for a white noise input signal. Five of the six feedback reduction techniques achieve worse feedback reduction for the tonal opera input signal than for the speech input signal. Reducing the adaptation speed did not result in an improved performance for the opera signal and degrades the tracking performance. Exploiting an initial feedback path measurement in the feedback canceller results in improved performance for tonal signals at the expense of a worse feedback reduction in the acoustic conditions that differ from the condition for which the initialization was performed as well as a worse tracking of feedback path changes. Repeated measures indicated that the reproducibility of the test set-up is crucial, in particular when the hearing aid operates close to instability.

This study is a first step toward the development of a benchmark approach for feedback reduction in commercial (black-box) devices and experimental algorithms, based on relevant physical performance measures and appropriate real-life test signals.

ACKNOWLEDGMENT

The authors would like to thank Mark Laureyns of GN Resound and Michael Cock of Phonak for providing the commercial hearing aids and Sara Wambacq for help in the measurements.

This research was carried out at the ESAT laboratory and the ExpORL laboratory of K.U. Leuven, in the frame of the European Union FP6 Project 004171 HEARCOM, the Belgian Programme on Interuniversity Attraction Poles initiated by the Belgian Federal Science Policy Office IUAP P6/04 (DYSCO, 2007-2011) and the K.U. Leuven Research Council Concerted Research Action GOA-AMBioRICS.

[1]For B, instability in the absence of the gain limitation is meant.
[2]MSGS refers to the maximum stable hearing aid gain setting, whereas MSG refers to the measured and hence, true maximum stable gain.
[3]For B, $MSGS_{off}$ in the absence of the gain limitation is meant.
[4]Since the MSG of hearing aid B in the 'Normal' condition could not be determined, a retest of hearing aid B was not done to reduce measurement time.

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