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

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(Dated: July 23, 2009)

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

Abstract

This paper presents an objective evaluation of four feedback cancellation techniques in commercial hearing aids and one recently developed adaptive feedback cancellation algorithm. Objective performance measures are discussed 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. Four of the five techniques achieve worse feedback reduction for a tonal opera input signal than for a speech input signal. Constraining the feedback canceller based on 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.

PACS numbers: 43.66.Ts (hearing aids), 43.60.Mn (adaptive processing)

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 oftentimes cannot be amplified sufficiently. To reduce the acoustic feedback, feedback cancellation may be applied (cf. Figure 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. An often used criterion for assessing a feedback reduction technique is the maximum stable gain (MSG), i.e., the maximum gain that can be applied without rendering the system unstable. In Freed and Soli (2006); Greenberg *et al.* (2000); Maxwell and Zurek (1995); Freed (2008); 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. Freed and Soli (2006) and Shin *et al.* (2007) proposed objective criteria for instability based on the power concentration ratio and the hearing aid transfer function variation, respectively. However, these criteria only hold for a white noise input signal to the hearing aid. In 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

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aid gain settings, i.e., a low gain, a high gain and a zero gain, i.e., with the hearing aid switched off. The method uses a white noise input signal and assumes that the hearing aid behaves as a linear amplifier. Objective procedures for determining the MSG are thus generally 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 (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 then 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, existing evaluation procedures typically assume that the hearing aid behaves as a linear amplifier (Shin *et al.*, 2007; Merks *et al.*, 2006). This is rarely the case in practice due to non-linear processing such as dynamic range compression and due to the saturation of the hearing aid receiver at high gains. Finally, the MSG is typically derived based on the hearing aid gain setting (Freed and Soli, 2006; Merks *et al.*, 2006). Some devices, however, reduce their gain in order to avoid feedback. In this case, the gain setting will not correspond to the actually applied gain setting, which then indeed obscures the analysis.

In addition to providing a high MSG, the feedback reduction technique should also preserve a good sound quality. In Freed and Soli (2006); 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 is assessed by measuring the occurrence of entrainment artefacts when presenting the hearing aid with music, machine noise, tonal and human sounds. 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 procedure for computing the entrainment requires that the hearing aid gain be at least 5 dB below instability with the feedback reduction technique disabled.

Finally, the feedback reduction technique should quickly adjust to feedback path changes so that when the system becomes unstable, the duration of instability is minimal. 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.

In this paper, the performance of four adaptive feedback cancellation based feedback reduction techniques as implemented in recent commercial hearing aids is evaluated for spectrally colored input signals. A comparison is made with the prediction error method based adaptive feedback canceller described in Spriet *et al.* (2006) implemented on an experimental hearing aid platform. The evaluation is based on objective performance measures for detecting the presence of feedback, oscillations and signal distortion in the case of a spectrally colored input signal (Spriet *et al.*, 2008, 2009). The measures compare the actual hearing aid output with the hearing aid output obtained in the absence of feedback (i.e., reference signal). 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. Based on the objective 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 actual hearing aid gain. In addition, the reduction of feedback, distortions and oscillations at several gain settings 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 using a Phonak Fit-and-Go open fitting. 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 meter 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 seconds of stationary white noise and 10 seconds of speech weighted noise from the HINT (Hearing in Noise Test) database (Nilsson *et al.*, 1994), 17 seconds of real speech from the HINT database and a 20 seconds opera fragment of 'Der Hölle Rache' from 'Die Zauberflöte' of W.A. Mozart. The signal at the in-the-ear microphone of the Cortex II artificial head was amplified and recorded. The 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.

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 (cf. Figure 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. Section III.C.3).

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 (cf. Figure 1). In addition, 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.

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 fitting 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. In hearing aid C, the adaptation speed of the feedback canceller is selected based on an analysis of the input signal. Hearing aid A uses the measured feedback path to constrain the adaptive feedback canceller (Kates, 1999, 2003). 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 over the feedback limit, 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. 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 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 the 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. To assess the performance of the feedback reduction system only, all 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 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, 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 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 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 hearing aid output PSDs were measured with the feedback canceller disabled at a gain of 18 dB or more below instability[1] such that the measurement was not influenced by the presence of acoustic feedback or the feedback canceller. The hearing aid output signal 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 4 depicts the output PSDs of the different hearing aids. Given the coarse controls for adjusting the frequency response, differences in the resulting output PSD of up to $\pm$ 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 objective measures for quantifying the amount of feedback, oscillations and distortion. 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 $r[k]$ (reference signal) that would be obtained in the same acoustic scenario but then in the absence of acoustic feedback.

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 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 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 that does not assume linear system behavior. 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 (Spriet *et al.*, 2009). For the closed fitting, a temporary foam earmold E-A-RTEMP 13A (EARtone) was used. With the closed fitting, the amount of feedback in the recording is minimal. 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 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 the MSG with the feedback canceller disabled. In the sequel, the MSG with the feedback canceller disabled is referred to as $\mathrm{MSG}_{\mathrm{off}}$. 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.

B. Performance measures

This section defines objective measures for detecting instability, oscillations and signal distortion in the hearing aid output, which are applicable to spectrally colored input signals. 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 will be computed using frames of 0.5 sec 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 will then be used.

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 set to 6 dB below $\mathrm{MSG}_{\mathrm{off}}$[2] and with the feedback canceller disabled. The noise signal was rescaled to compensate for the gain difference between $\mathrm{MSG}_{\mathrm{off}}$- 6 dB and the actual gain.

In the sequel, 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.

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)

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

$L+1$ is the window length used in the energy computation. 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. To reduce the effect of differences in the environmental and internal noise component in $u[k]$ and $r[k]$, $E(k)$ is only computed when the short-term energy of the reference signal $r[k]$ exceeds the environmental noise $n[k]$ by at least 10 dB.

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):

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

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

In addition, the short-term intelligibility weighted feedback-to-reference signal energy ratio

$\text{FSR}_{\text{intellig}}(k)$ in the hearing aid output is computed as:

$$\text{FSR}_{\text{intellig}}(k) = \sum_i w_{ERB}[i]\text{FSR}_i, \tag{3}$$

with FSR_i the feedback-to-reference signal energy ratio in the ith auditory critical band, defined by the equivalent rectangular bandwidth (ERB) of auditory filters (Moore, 2003)

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

with $P_v(f,k)$ the short-term PSD of the feedback signal $v[k] = u[k] - r[k]$, $P_r(f,k)$ the short-term PSD of $r[k]$ and $P_n(f)$ the long-term PSD of $n[k]$. The weight $w_{ERB}[i]$ gives an equal weight to each auditory critical band B_i between 300 Hz and 6500 Hz. 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 $\beta = 2$). In addition, $\text{FSR}_{\text{intellig}}(k)$ is only computed when the intelligibility weighted reference signal to environmental and internal noise ratio was at least 10 dB. Since the bandwidth of the auditory critical bands increases with frequency, high frequency oscillations are weighted less in $\text{FSR}_{\text{intellig}}$ than low frequency oscillations frequencies.

2. Detection of oscillations

Even before instability occurs, oscillations can already be perceived. Freed and Soli (2006) and Shin *et al.* (2007) defined objective criteria to detect oscillations, referred to as the power concentration ratio (PCR) and the hearing aid transfer function variation criterion (TVC), 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 coloured 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

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

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

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

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:

$$\begin{aligned} P_u(f,k) &= \max(P_u(f,k), \alpha P_n(f)), \\ P_r(f,k) &= \max(P_r(f,k), \alpha P_n(f)), \end{aligned} \tag{7}$$

where $P_n(f)$ is the long-term PSD of the internal and environmental noise $n[k]$ and $\alpha > 1$. Hence, only frequencies where the energy exceeds the environmental noise energy by at least $10\log_{10}\alpha$, are taken into account.

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

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

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. Hence, TVF(f,k) is related to the loop gain $G(f,k)\left(F(f,k)-\hat{F}(f,k)\right)$. For example, for a loop gain of 0.9 (which is close to instability), the TVC equals 20 dB.

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

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* $\Delta PCR(k)$ *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 TVF(f,k) (cf. 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 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 PCR$_r(k)$.

3. The difference ΔPCR(k)

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

is then used as a measure for the presence of oscillations. Caution should be used when the power concentration 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, $\Delta\text{PCR}(k)$ 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 $\Delta\text{PCR}(k)$ than oscillation frequencies with a low signal power.

A second measure is referred to as the *normalized power concentration ratio* $\text{PCR}_n(k)$ and consists in computing the PCR of Freed and Soli (2006) on the normalized PSD $\frac{P_u(f,k)}{P_r(f,k)}$. First, the oscillation frequencies are determined based on the transfer function variation function (see step 1 above). The fraction of the normalized power $\frac{P_u(f,k)}{P_r(f,k)}$ that is located at the five (or less) strongest oscillation frequencies is referred to as $\text{PCR}_n(k)$. The normalized power concentration ratio does not take into account the power of the oscillation frequencies, i.e., 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 (cf. Figure 4) 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 $\text{SD}(k)$ is defined as

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

The frequency-weighting factor $w_{ERB}(f)$ gives equal weight to each auditory critical band between $f_l = 300\text{Hz}$ and $f_u = 6500\text{Hz}$. 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. In addition, $\text{SD}(k)$ is only computed when the short-term energy of the reference signal $r[k]$ exceeds the environmental and internal noise energy by at least 10 dB.

C. Evaluation procedure

1. Added stable gain

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

$$\mathrm{ASG} = \mathrm{MSG_{on}} - \mathrm{MSG_{off}}. \tag{11}$$

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 in-the-ear microphone signal with the hearing aid switched on was recorded. To avoid adaptation effects, the signal was presented once before the recording was made. The hearing aid output was computed by subtracting the direct signal path contribution from the in-the-ear microphone signal. 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 $\max_k\{\mathrm{E}(k)\}$ over the whole signal segment was computed. The maximum gain setting for which $\max_k\{\mathrm{E}(k)\}$ remained below 6 dB was determined. To compensate for a possible frequency-dependent attenuation 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 (i.e., a gain setting of 6 dB below $\mathrm{MSG_{off}}$).

To be able to compare the output level of the different hearing aids, the sound pressure level of the reference signal with the different hearing aids at $\mathrm{MSG_{off}}$ was computed.

2. Performance measures at gains between MSG_{off} and MSG_{on}

A high ASG does not necessarily guarantee a high feedback suppression and a good sound quality at all gain settings. Therefore, the maximum amount of feedback ($E(k)$, FSR(k) and FSR$_{\text{intellig}}(k)$), oscillations (TVC(k), ΔPCR$(k)$and PCR$_n(k)$) and distortion (SD(k)) for the speech and the opera signal was also determined at MSG$_{\text{off}}$, MSG$_{\text{on}}$ and at the intermediate gain settings that correspond to an actual gain of MSG$_{\text{off}}+6$ dB and MSG$_{\text{off}}+12$ dB. The intermediate gains were only considered when smaller than MSG$_{\text{on}}$. The performance measures of the ascending and the descending protocol were averaged. In addition to the performance measures, the sound pressure level of the reference signals was computed.

3. Tracking performance

In addition to a high added stable gain and a good sound quality, the feedback suppression 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 Figure 3). In the start position, the left ear of the artificial head was turned away from the lamp by 60 degrees (clockwise rotation). 50 seconds of stationary HINT noise was presented with a level of 60 dBA at the center of the head. After 20 seconds of signal presentation, the artificial head was rotated by -60 degrees using the Animatics SmartMotor. The velocity and acceleration of the rotation were equal to 100 degrees/sec 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 gain setting of the hearing aid was set to 6 dB above the maximum stable gain MSG$_{\text{off}}$ of the hearing aid in the end position. Note that this does not necessarily mean that the actual gain of the hearing aid was 6 dB above MSG$_{\text{off}}$. For A, C, D, E and E-s, a 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.

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 seconds with an overlap of 80%. A window length of 0.2 seconds instead of 0.5 seconds 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 were used for the different measures. If stability was not achieved within 30 seconds 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 also depends on the angle over which the head is rotated. Therefore, the closed fitting reference signals were transformed to open fitting reference signals by means of an adaptive recursive-least-squares (RLS) filter. The adaptive filter was determined as the filter 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 instability in the end position with the feedback canceller disabled). For the identification of the adaptive 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 systems was measured for the four input signals in the two conditions 'Normal' and 'Handset'. Figure 5 depicts the ASG for the white noise and the HINT noise signal. Figure 6 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. 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 is depicted with squares. This ASG was determined as the difference between $\mathrm{MSG}_{\mathrm{off}}$ with feedback path initialization and $\mathrm{MSG}_{\mathrm{off}}$ without feedback path initialization. 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 the gain setting $\mathrm{MSG}_{\mathrm{off}}$. In the 'Normal' condition, the largest level difference between hearing aids for a given sound signal is 4 dB. In the 'Handset' condition, the largest level difference is 7 dB. In the 'Handset' condition, $\mathrm{MSG}_{\mathrm{off}}$ 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, the maximum output-to-reference signal ratio $\max_k\{\mathrm{E}(k)\}$ with the feedback canceller disabled slightly exceeds 6 dB at a gain setting of 1dB above $\mathrm{MSG}_{\mathrm{on}}$, but does not further increase for larger gain settings above $\mathrm{MSG}_{\mathrm{on}}$. As a result, the ASG that is offered by the feedback canceller could not be determined. This is indicated Figure 5 and Figure 6 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, 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 MSG_{on} (except for the music mode): the maximum added gain without limitation equals 3 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 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. 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 reduction for the 'Handset' condition. 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 and from 5 dB to 19 dB for the 'Handset' condition across hearing aids. Hence, the performance 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 the ASG), 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 constrained adaptation, 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 gain settings below $\mathrm{MSG_{on}}$ (see Section 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 step size of the gain control. For the opera signal, larger differences sometimes occur, in particular for C-m, D-m, E and E-s. For the adaptive feedback cancellers C-m, E and E-s, the environmental and internal 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 $\mathrm{E}(k)$. The larger differences between ASGs obtained with test and retest data and with the ascending and descending protocol for C-m, E and E-s may be due to a small gain margin at intermediate gain settings. For the static feedback canceller D-m, the differences between ASGs with test and retest data in the 'Normal' condition is 6 dB. A small deviation between the initialized and the actual feedback path may have a large impact on the feedback suppression 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 between MSG_{off} and MSG_{on}

Table IV to Table VII show the maximum performance measures for the speech and the opera signal, at the gains MSG_{off}, MSG_{on} and the intermediate gains MSG_{off}+ 6 dB and MSG_{off} + 12 dB. Table IV and Table V show the results for the speech signal in the 'Normal' condition and the 'Handset' condition. Table VI and Table VII depict the results for the opera signal. The performance of hearing aid B for the speech signal in the 'Normal' condition is not shown because MSG_{off} has never been reached. The performance measures of 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.

- For the speech signal, all feedback cancellers achieve good performance below MSG_{on} in both conditions: the amount of feedback, oscillations and distortion is reduced compared to the hearing aid output when the feedback canceller is disabled at MSG_{off}. In the 'Normal' condition, hearing aid A achieves the best feedback suppression, followed by hearing aids E and E-s. In the 'Handset' condition, the improvement of hearing aid A is smaller due to the constrained adaptation. Hearing aid C and D exhibit slightly more oscillations at MSG_{off}+ 6 dB and MSG_{off} + 12 dB, respectively, compared to the other hearing aids.

- For the tonal opera signal, the performance of the feedback cancellers in hearing aids B, C, D and E/E-s degrades. For D and E/E-s, the amount of feedback, distortion and oscillations is only slightly or even not reduced compared to the hearing aid output at MSG_{off} when the feedback canceller is disabled. At MSG_{off}, the feedback reduction systems of hearing aid B and C succeed to reduce the amount of feedback or oscillations. However, for C, a smaller ASG is achieved than for the speech signal and for B, oscillations appear in the 'Handset' condition at $MSG_{off} + 6$ dB. The performance of hearing aid A is not affected by the tonal signal thanks to the constrained adaptation. In particular for hearing aid D and E, distortion and oscillations are present after feedback cancellation, even at gains below MSG_{on}. 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 improves the performance of hearing aid E for the tonal signal. In the 'Handset' condition, only a small improvement by E-s is observed in the intelligibility weighted feedback-to-reference signal ratio $FSR_{intellig}$, the distortion SD, transfer function variation criterion TVC and PCR_n.

Test and retest data are consistent with each other. Differences between test and retest data for the 'Handset' condition at MSG_{off} are due to small variations in the positioning of the handset. 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 differences are larger at gains where TVC exceeds 20 dB. In the 'Handset' condition, slight changes in the positioning of the handset may also result in differences between test and re-test data (e.g., hearing aid A at MSG_{off}).

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 smaller than PCR_n 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 much smaller than FSR. However, feedback especially occurs at high frequencies (i.e., above 2 kHz). These frequencies are weighted less in $FSR_{intellig}$.

C. Tracking performance

Table VIII depicts the duration of instability of the feedback cancellers based on the different performance measures as well as the sound pressure level (dB SPL) of the feedback-free reference

signal in the start and the end position of the artificial head. 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 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. Due to the constrained adaptation, the feedback canceller of hearing aid A only achieves an ASG of 1 a 2 dB in the end position. As a result, the feedback canceller is not able to track the feedback path change at a gain setting of MSG_{off}+6dB. 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 then hearing aid B. It should however be noted that for hearing aid B, the gain setting MSG_{off}+ 6dB 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.

The differences between the two measurements within each run are generally small, which suggests that the rotation of the head by the motor is reproducible. For hearing aid C and in particular for C-m, larger differences are 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 are generally larger, 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, Figure 7 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.

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 can not be perfectly matched and compression can often not be completely disabled. However, from the objective evaluation, the strengths and the weaknesses of the algorithms can be identified. Thanks to the constrained adaptation, the ASG and the sound quality of hearing aid A is not affected by a tonal input signal. However, this goes at the expense of a worse tracking performance and a worse feedback suppression 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 sound quality 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) could be improved. 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 the MSG, oscillations may occur. Previous studies often use 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 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 performance measures give a first indication of the algorithm's performance. However, further research is necessary to measure the perceptual relevance of the objective measures for the hearing aid user. For example, the TVC and PCR_n may detect oscillations that are masked by more powerful frequencies. The perceptual relevance may be improved through the development of psycho-acoustically motivated measures that take into account 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, C-m and E.

VI. CONCLUSIONS

In this paper, the performance of the feedback reduction techniques in four commercial hearing aids and one recently developed feedback cancellation technique were assessed based on an objective procedure. 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 addition, the reduction of feedback, oscillations and distortion at gain values below the maximum stable gain was 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. Four of the five hearing aids achieve worse feedback suppression 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. Constraining the adaptive feedback canceller based on 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.

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

VII. 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 Pro-

gramme 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. Ann Spriet is a postdoctoral researcher funded by FWO-Vlaanderen.

Endnotes

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

2. For B, $\mathrm{MSG}_{\mathrm{off}}$ in the absence of the gain limitation is meant.

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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

Table III. Sound pressure level (dB SPL) of the reference signal at MSG off for the four input signals and the two acoustic conditions. 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

Table I. 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.

	Manufacturer	Model	Algorithm type	Initialization	Music mode	Delay
A	GN Resound	Azure AZ80-DVI	Feedback canceller with constrained adaptation	yes	-	5.6 ms
B	Phonak	Savia Art411 dSZ	Feedback canceller and frequency-dependent gain limitation	yes	reduced adaptation speed	7.1 ms
C	Siemens	Centra HP	Feedback canceller with automatic speed control	no	reduced adaptation speed	4.4 ms
D	Starkey	Destiny 400	Feedback canceller	yes	static feedback canceller	4.7 ms
E	K.U. Leuven	n/a	PEM-based feedback canceler (PemAFC) (E)	no	-	6.5 ms
			PemAFC combining a fast and slowly adaptive filter (E-s)	no	-	6.5 ms

Table IV. Performance measures of the feedback cancellers in the hearing aid systems A, C, D and E at the gains MSG_{off}, MSG_{off}+ 6 dB, MSG_{off}+12 dB and MSG_{on} for the speech signal in the 'Normal' condition for test and retest (test/retest).

	FC	Gain	dB SPL [dB]	E [dB]	FSR [dB]	$FSR_{intellig}$ [dB]	SD [dB]	TVC [dB]	Δ PCR [-]	PCR_n [-]
A	off	MSG_{off}	87/ 86	5.5/ 4.4	1.7/ -0.3	-7.1/ -7.9	2.0/ 1.6	11/ 12	0.29/ 0.26	0.19/ 0.15
	on	MSG_{off}	87/ 86	0.1/ 0.2	-10.2/ -9.8	-10.2/ -11.2	0.9/ 0.6	3/ 2	0.00/ 0.00	0.00/ 0.00
	on	+6	93/ 92	0.2/ 0.3	-9.8/ -10.0	-9.7/ -10.3	0.9/ 0.6	4/ 3	0.00/ 0.00	0.00/ 0.00
	on	+12	99/ 98	0.3/ 0.7	-8.6/ -9.6	-8.7/ -9.3	1.1/ 0.8	4/ 4	0.00/ 0.00	0.00/ 0.00
	on	MSG_{on}	108/ 107	5.2/ 5.3	3.8/ 3.7	-1.6/ -2.2	5.8/ 5.9	30/ 24	0.17/ 0.12	0.82/ 0.69
C	off	MSG_{off}	88/ 87	6.0/ 5.7	2.5/ 1.9	-9.7/ -9.8	1.7/ 1.5	10/ 12	0.17/ 0.15	0.14/ 0.19
	on	MSG_{off}	88/ 87	0.9/ 0.7	-3.6/ -3.4	-10.1/ -9.8	1.5/ 1.3	6/ 6	0.04/ 0.09	0.02/ 0.04
	on	+6	94/ 93	2.0/ 1.6	-1.6/ -1.6	-9.4/ -9.1	1.7/ 1.5	7/ 7	0.18/ 0.14	0.05/ 0.05
	on	MSG_{on}	99/ 98	4.9/ 5.5	2.4/ 3.2	-7.5/ -6.2	2.2/ 2.3	11/ 10	0.47/ 0.42	0.19/ 0.21
D	off	MSG_{off}	87/ 87	3.5/ 3.7	-1.1/ -1.1	-7.7/ -8.0	1.4/ 1.7	6/ 7	0.18/ 0.19	0.04/ 0.08
	on	MSG_{off}	87/ 87	-0.2/ -0.1	-4.7/ -4.3	-7.7/ -8.3	1.3/ 1.7	5/ 5	0.00/ 0.00	0.00/ 0.00
	on	+6	93/ 93	-0.0/ -0.1	-3.1/ -3.6	-7.1/ -7.5	1.4/ 1.8	7/ 6	0.08/ 0.00	0.05/ 0.00
	on	+12	99/ 99	0.2/ 0.3	-2.3/ -1.8	-6.3/ -6.8	1.8/ 2.2	9/ 8	0.13/ 0.20	0.27/ 0.13
	on	MSG_{on}	106/ 107	3.0/ 2.5	1.0/ 0.5	-3.9/ -4.0	3.8/ 3.2	25/ 20	0.22/ 0.31	0.77/ 0.58
E	off	MSG_{off}	89/ 89	4.5/ 5.7	2.8/ 4.3	-6.8/ -8.0	1.5/ 1.6	8/ 9	0.37/ 0.41	0.10/ 0.14
	on	MSG_{off}	89/ 89	0.4/ 0.5	-11.2/ -10.5	-8.3/ -9.5	1.3/ 1.7	5/ 5	0.00/ 0.00	0.00/ 0.00
	on	+6	95/ 95	0.7/ 0.9	-10.5/ -10.0	-8.3/ -9.4	1.4/ 1.8	4/ 5	0.00/ 0.00	0.00/ 0.00
	on	+12	100/ 101	1.5/ 1.9	-7.4/ -6.3	-7.9/ -8.8	1.4/ 1.9	7/ 8	0.00/ 0.00	0.04/ 0.05
	on	MSG_{on}	106/ 106	5.1/ 4.7	2.5/ 1.5	-4.5/ -6.5	3.1/ 3.0	16/ 17	0.40/ 0.32	0.39/ 0.44
	on	MSG_{off}	89/ 89	0.4/ 0.4	-11.1/ -10.9	-8.3/ -9.3	1.3/ 1.7	4/ 4	0.00/ 0.00	0.00/ 0.00
	on	+6	95/ 95	0.7/ 0.8	-11.2/ -10.7	-8.5/ -9.6	1.3/ 1.6	4/ 4	0.00/ 0.00	0.00/ 0.00
	on	+12	100/ 101	1.4/ 1.8	-8.3/ -6.2	-8.3/ -9.3	1.3/ 1.7	7/ 8	0.00/ 0.00	0.04/ 0.05

	on	MSG_{on}	107/ 107	5.5/ 4.7	3.2/ 1.7	-3.8/ -6.0	3.6/ 3.3	18/ 17	0.38/ 0.30	0.45/ 0.45

Table V. Performance measures of the the feedback cancellers in the hearing aid systems A, B, C, D and E at the gains MSG_{off}, MSG_{off}+ 6 dB, MSG_{off}+12 dB and MSG_{on} for the speech signal in the 'Handset' condition for test and retest (test/retest).

	FC	Gain	dB SPL [dB]	E [dB]	FSR [dB]	$FSR_{intellig}$ [dB]	SD [dB]	TVC [dB]	Δ PCR [-]	PCR_n [-]
A	off	MSG_{off}	74/ 77	2.6/ 3.8	-2.3/ 0.6	-6.3/ -6.1	1.7/ 2.1	12/ 16	0.19/ 0.40	0.18/ 0.42
	on	MSG_{off}	74/ 77	1.9/ 0.9	-4.2/ -8.1	-7.7/ -8.7	1.2/ 1.2	5/ 6	0.00/ 0.00	0.00/ 0.00
	on	MSG_{on}	79/ 83	3.0/ 4.1	-0.4/ 0.2	-5.5/ -4.8	2.4/ 2.7	20/ 18	0.39/ 0.37	0.65/ 0.57
B	off	MSG_{off}	77	3.3	-0.1	-6.0	3.0	15	0.25	0.35
	on	MSG_{off}	77	0.7	-9.0	-10.6	1.0	5	0.00	0.00
	on	+6	84	0.9	-7.5	-8.6	1.3	6	0.00	0.04
	on	MSG_{on}	89	4.0	1.6	-4.2	5.8	22	0.29	0.69
C	off	MSG_{off}	73/ 72	4.7/ 4.7	1.4/ 1.6	-8.6/ -8.1	1.5/ 1.7	7/ 9	0.31/ 0.37	0.09/ 0.12
	on	MSG_{off}	73/ 72	0.8/ 0.8	-4.5/ -3.1	-8.7/ -8.3	1.2/ 1.3	7/ 7	0.07/ 0.08	0.05/ 0.06
	on	+6	79/ 78	1.6/ 1.8	-3.2/ -2.1	-8.3/ -7.3	1.5/ 1.6	7/ 8	0.14/ 0.24	0.05/ 0.06
	on	MSG_{on}	85/ 84	5.4/ 5.7	2.5/ 3.0	-4.9/ -4.2	2.7/ 2.9	11/ 12	0.46/ 0.45	0.25/ 0.28
D	off	MSG_{off}	73/ 74	3.3/ 3.7	-2.4/ -1.1	-6.9/ -6.6	1.3/ 1.8	8/ 9	0.17/ 0.33	0.09/ 0.12
	on	MSG_{off}	73/ 74	0.6/ 0.0	-6.1/ -7.0	-7.0/ -7.4	1.2/ 1.6	5/ 5	0.00/ 0.00	0.00/ 0.00
	on	+6	79/ 80	0.3/ -0.0	-7.7/ -7.6	-6.5/ -6.7	1.4/ 1.8	5/ 5	0.00/ 0.00	0.00/ 0.00
	on	+12	85/ 86	0.9/ 0.5	-5.8/ -4.7	-5.9/ -5.7	1.6/ 2.1	7/ 9	0.05/ 0.11	0.05/ 0.12
	on	MSG_{on}	91/ 90	3.9/ 4.2	1.6/ 1.6	-4.0/ -4.6	2.7/ 3.0	15/ 17	0.47/ 0.50	0.40/ 0.53
E	off	MSG_{off}	72/ 72	5.1/ 5.3	3.6/ 3.9	-5.7/ -6.1	2.4/ 2.5	13/ 13	0.52/ 0.54	0.32/ 0.34
	on	MSG_{off}	72/ 72	0.8/ 0.6	-4.5/ -6.8	-7.0/ -7.8	1.6/ 1.5	5/ 5	0.00/ 0.00	0.00/ 0.00
	on	+6	78/ 79	0.7/ 0.9	-7.7/ -8.1	-7.4/ -8.0	1.5/ 1.5	5/ 5	0.00/ 0.00	0.00/ 0.00
	on	+12	84/ 84	1.6/ 1.5	-5.7/ -6.1	-7.4/ -8.1	1.6/ 1.5	5/ 5	0.00/ 0.00	0.00/ 0.00
	on	MSG_{on}	89/ 89	5.0/ 4.9	2.7/ 2.7	-6.4/ -7.0	2.2/ 2.2	10/ 11	0.42/ 0.44	0.24/ 0.25
	on	MSG_{off}	72/ 72	0.6/ 0.6	-5.0/ -6.8	-7.1/ -7.8	1.5/ 1.5	5/ 5	0.00/ 0.00	0.00/ 0.00

	on	+6	78/ 79	0.7/ 0.8	-7.7/ -8.0	-7.4/ -8.1	1.5/ 1.5	5/ 5	0.00/ 0.00	0.00/ 0.00
	on	+12	84/ 84	1.4/ 1.4	-6.4/ -6.9	-7.5/ -8.3	1.5/ 1.4	5/ 4	0.00/ 0.00	0.00/ 0.00
	on	MSG_{on}	89/ 90	3.9/ 5.4	0.8/ 3.4	-6.7/ -6.4	1.9/ 2.4	9/ 11	0.33/ 0.47	0.16/ 0.30

Table VI. Performance measures of the feedback cancellers in the hearing aid systems A, B, C, D and E at the gains MSG_{off}, MSG_{off}+ 6 dB, MSG_{off}+12 dB and MSG_{on} for the opera signal in the 'Normal' condition for test and retest (test/retest).

	FC	Gain	dB SPL [dB]	E [dB]	FSR [dB]	$FSR_{intellig}$ [dB]	SD [dB]	TVC [dB]	Δ PCR [-]	PCR_n [-]
A	off	MSG_{off}	86/ 86	5.1/ 5.5	1.0/ 1.7	-6.4/ -5.8	1.8/ 2.1	11/ 19	0.33/ 0.34	0.15/ 0.57
	on	MSG_{off}	86/ 86	0.5/ 0.8	-13.8/ -14.9	-8.0/ -8.8	0.5/ 0.4	2/ 2	0.00/ 0.00	0.00/ 0.00
	on	+6	93/ 92	1.1/ 1.5	-10.8/ -10.0	-7.3/ -8.2	0.7/ 0.7	3/ 4	0.00/ 0.00	0.00/ 0.00
	on	+12	99/ 98	1.5/ 2.1	-5.8/ -5.2	-5.7/ -6.2	1.5/ 1.5	13/ 14	0.01/ 0.04	0.26/ 0.20
	on	MSG_{on}	106/ 105	2.7/ 4.5	0.0/ 3.4	-1.9/ -1.8	4.5/ 5.0	30/ 32	0.49/ 0.67	0.88/ 0.95
B	off	MSG_{off}	87	5.2	0.5	-7.0	1.4	7	0.27	0.12
	on	MSG_{off}	87	0.7	-8.7	-7.1	0.9	4	0.00	0.00
	on	MSG_{on}	93	0.6	-4.6	-3.7	1.9	9	0.04	0.08
B-m	on	MSG_{off}	87	0.7	-6.2	-7.4	0.9	5	0.00	0.00
	on	MSG_{on}	93	0.6	-3.8	-3.4	1.9	7	0.00	0.05
C	off	MSG_{off}	86/ 86	5.1/ 5.0	0.3/ 0.5	-8.4/ -8.6	1.2/ 1.3	7/ 9	0.05/ 0.06	0.08/ 0.11
	on	MSG_{off}	86/ 86	1.3/ 1.6	-8.9/ -10.3	-7.9/ -7.8	1.5/ 1.5	8/ 9	0.00/ 0.01	0.08/ 0.11
	on	MSG_{on}	91/ 92	2.5/ 3.1	-5.6/ -4.2	-6.4/ -5.9	1.7/ 1.8	12/ 11	0.00/ 0.00	0.25/ 0.24
C-m	on	MSG_{off}	86/ 86	1.6/ 1.6	-9.1/ -9.2	-8.1/ -7.8	1.7/ 1.7	11/ 10	0.01/ 0.01	0.15/ 0.15
	on	MSG_{on}	89/ 90	2.2/ 2.4	-7.6/ -7.4	-7.5/ -6.8	1.6/ 1.6	9/ 9	0.01/ 0.01	0.12/ 0.14
D	off	MSG_{off}	88/ 89	4.4/ 4.3	0.1/ 0.6	-6.0/ -6.8	1.4/ 1.6	11/ 14	0.30/ 0.27	0.27/ 0.39
	on	MSG_{off}	88/ 89	3.4/ 3.4	0.1/ 0.5	-5.6/ -5.5	3.5/ 3.3	19/ 18	0.40/ 0.44	0.48/ 0.47
	on	MSG_{on}	92/ 93	4.4/ 3.7	2.3/ 3.1	-5.0/ -4.7	4.6/ 5.4	25/ 30	0.54/ 0.62	0.79/ 0.93
D-m	on	MSG_{off}	88/ 89	0.5/ 0.6	-16.0/ -16.7	-9.1/ -9.8	0.4/ 0.6	2/ 2	0.00/ 0.00	0.00/ 0.00
	on	+6	94/ 95	0.9/ 1.3	-12.3/ -12.7	-8.5/ -9.3	1.0/ 0.8	7/ 4	0.00/ 0.00	0.10/ 0.00
	on	MSG_{on}	96/ 101	1.1/ 3.2	-10.9/ -4.7	-8.2/ -7.7	1.3/ 1.6	13/ 10	0.01/ 0.00	0.30/ 0.18
E	off	MSG_{off}	90/ 90	5.6/ 5.4	2.4/ 2.5	-5.4/ -7.2	1.7/ 1.7	8/ 9	0.29/ 0.33	0.14/ 0.17

	on	MSG_{off}	90/ 90	2.3/ 1.9	-4.2/ -4.6	-3.8/ -3.0	2.5/ 3.2	12/ 15	0.01/ 0.01	0.14/ 0.34
	on	+6	95/ 96	2.9/ 2.2	-1.8/ -3.2	-3.3/ -3.2	4.4/ 3.9	19/ 17	0.01/ 0.01	0.54/ 0.42
	on	+12	101/ 102	3.7/ 3.0	0.2/ -1.9	-2.1/ -2.5	6.2/ 5.0	26/ 21	0.02/ 0.03	0.80/ 0.60
	on	MSG_{on}	103/ 107	4.0/ 3.9	0.7/ 1.5	-0.3/ 1.1	7.4/ 9.3	30/ 35	0.07/ 0.10	0.78/ 0.96
	on	MSG_{off}	90/ 90	1.3/ 0.9	-6.1/ -5.9	-4.6/ -4.7	2.5/ 2.6	10/ 13	0.16/ 0.17	0.16/ 0.17
	on	+6	95/ 96	1.5/ 1.2	-6.1/ -6.0	-5.0/ -4.2	2.6/ 2.7	10/ 11	0.09/ 0.03	0.16/ 0.17
	on	+12	101/ 102	2.1/ 2.0	-5.0/ -5.7	-3.7/ -3.0	2.5/ 2.7	10/ 12	0.07/ 0.02	0.19/ 0.27
	on	MSG_{on}	110/ 109	4.8/ 4.1	2.1/ -1.6	0.0/ -0.3	3.7/ 3.7	18/ 19	0.36/ 0.09	0.52/ 0.53

Table VII. Performance measures of the feedback cancellers in the hearing aid systems A, B, C, D and E at the gains MSG_{off}, MSG_{off}+ 6 dB, MSG_{off}+12 dB and MSG_{on} for the opera signal in the 'Handset' condition for test and retest (test/retest).

	FC	Gain	dB SPL [dB]	E [dB]	FSR [dB]	$FSR_{intellig}$ [dB]	SD [dB]	TVC [dB]	Δ PCR [-]	PCR_n [-]
A	off	MSG_{off}	74/ 79	4.1/ 5.1	-0.8/ 0.9	-4.6/ -4.9	1.7/ 2.0	10/ 12	0.23/ 0.32	0.17/ 0.21
	on	MSG_{off}	74/ 79	1.6/ 1.8	-5.2/ -7.1	-5.9/ -6.2	1.3/ 0.8	7/ 6	0.00/ 0.00	0.09/ 0.04
	on	MSG_{on}	78/ 85	3.1/ 4.2	-1.3/ 1.0	-4.1/ -3.3	2.6/ 3.8	20/ 30	0.26/ 0.35	0.68/ 0.94
B	off	MSG_{off}	77	4.3	1.9	-3.9	2.7	17	0.49	0.40
	on	MSG_{off}	77	1.8	-6.6	-5.8	1.2	6	0.00	0.04
	on	+6	86	3.1	-2.7	-3.5	2.3	12	0.21	0.21
	on	MSG_{on}	89	5.4	2.5	-1.5	4.0	18	0.34	0.52
B-m	on	MSG_{off}	77	3.0	-1.5	-4.6	1.9	12	0.37	0.18
	on	MSG_{on}	79	5.0	2.9	-3.0	3.1	19	0.53	0.52
C	off	MSG_{off}	75/ 72	5.5/ 5.4	1.8/ 1.6	-6.9/ -6.8	1.6/ 1.6	9/ 9	0.36/ 0.37	0.15/ 0.13
	on	MSG_{off}	75/ 72	3.3/ 3.0	-3.8/ -3.5	-6.9/ -6.7	1.4/ 1.4	7/ 7	0.03/ 0.05	0.09/ 0.09
	on	MSG_{on}	82/ 78	5.8/ 4.9	1.6/ 0.2	-5.4/ -5.5	2.0/ 1.7	10/ 9	0.30/ 0.16	0.16/ 0.12
C-m	on	MSG_{off}	75/ 72	3.6/ 3.3	-3.3/ -3.5	-6.7/ -6.7	1.6/ 1.6	9/ 9	0.06/ 0.09	0.15/ 0.12
	on	MSG_{on}	77/ 76	4.1/ 4.1	-2.0/ -1.8	-6.5/ -6.1	1.7/ 1.6	10/ 8	0.14/ 0.08	0.18/ 0.12
D	off	MSG_{off}	76/ 74	4.8/ 3.1	1.5/ -2.0	-5.4/ -6.0	1.7/ 1.4	10/ 7	0.29/ 0.09	0.17/ 0.09
	on	MSG_{off}	76/ 74	0.8/ 0.8	-0.7/ -1.6	-5.0/ -5.6	2.7/ 2.9	15/ 16	0.16/ 0.16	0.27/ 0.38
	on	+6	82/ 80	1.5/ 1.7	-0.2/ -0.0	-4.5/ -4.7	3.9/ 3.8	20/ 21	0.47/ 0.44	0.65/ 0.64
	on	MSG_{on}	88/ 86	2.0/ 3.0	1.2/ 2.4	-2.4/ -2.5	5.9/ 5.9	32/ 30	0.59/ 0.60	0.93/ 0.93
D-m	on	MSG_{off}	76/ 74	2.9/ 2.1	-3.6/ -5.8	-5.8/ -6.7	1.3/ 1.0	6/ 5	0.01/ 0.00	0.07/ 0.00
	on	MSG_{on}	78/ 78	4.9/ 5.3	2.2/ 2.5	-5.0/ -5.1	2.2/ 2.4	11/ 12	0.32/ 0.34	0.21/ 0.30
E	off	MSG_{off}	73/ 74	4.2/ 4.6	1.4/ 1.0	-5.2/ -5.2	2.1/ 2.0	11/ 11	0.49/ 0.50	0.26/ 0.23
	on	MSG_{off}	73/ 74	2.8/ 2.5	0.2/ -0.4	-4.0/ -3.5	2.0/ 2.2	12/ 10	0.35/ 0.35	0.13/ 0.10

	on	+6	79/ 80	3.8/ 2.7	-0.1/ -2.1	-3.5/ -2.3	3.2/ 2.9	14/ 15	0.28/ 0.18	0.29/ 0.34
	on	+12	85/ 85	4.6/ 4.2	0.8/ 0.7	-2.4/ -2.4	3.1/ 2.9	17/ 15	0.47/ 0.40	0.44/ 0.34
	on	MSG_{on}	89/ 87	5.4/ 5.6	2.2/ 2.9	-2.3/ -2.3	3.2/ 4.8	17/ 21	0.52/ 0.55	0.41/ 0.66
	on	MSG_{off}	73/ 74	1.8/ 2.6	-2.1/ 0.1	-3.9/ -3.5	1.9/ 2.1	8/ 8	0.01/ 0.35	0.06/ 0.09
	on	+6	79/ 80	3.4/ 3.8	1.3/ 1.9	-4.0/ -3.9	2.1/ 2.3	12/ 13	0.39/ 0.43	0.14/ 0.17
	on	+12	85/ 85	3.2/ 3.0	0.9/ 0.4	-4.3/ -3.7	2.1/ 2.3	12/ 10	0.37/ 0.37	0.12/ 0.10
	on	MSG_{on}	91/ 91	3.9/ 3.5	0.9/ 1.0	-3.5/ -3.7	2.4/ 2.4	11/ 10	0.37/ 0.39	0.15/ 0.15

Table VIII. Duration of instability (in seconds) based on the performance measures E, FSR, $FSR_{intellig}$, TVC, ΔPCR and PCR_n and the sound pressure level of the feedback free reference signal in the start and the end position.

HA	Run	Duration of instability [s] based on the following measures					
		E	FSR	$FSR_{intellig}$	TVC	ΔPCR	PCR_n
A	1	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞
A	2	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞
A	3	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞
B	1	0.5/ 0.4	0.6/ 0.6	0.4/ 0.3	0.4/ 0.3	0.4/ 0.3	0.4/ 0.4
B	2	0.2/ 0.2	0.4/ 0.4	0.1/ 0.2	0.2/ 0.2	0.2/ 0.2	0.3/ 0.2
B	3	0.3/ 0.3	0.4/ 0.4	0.2/ 0.2	0.2/ 0.2	0.2/ 0.2	0.3/ 0.3
B-m	1	1.2/ 1.3	1.6/ 1.6	1.4/ 1.3	1.3/ 1.3	1.2/ 1.2	1.3/ 1.3
B-m	2	0.2/ 0.4	0.8/ 0.8	0.9/ 0.9	0.0/ 0.3	0.0/ 0.3	0.3/ 0.4
B-m	3	1.2/ 1.2	1.4/ 1.4	1.2/ 1.1	1.2/ 1.2	1.2/ 1.2	1.2/ 1.2
C	1	0.9/ 0.9	1.0/ 1.0	0.8/ 0.9	0.9/ 0.9	0.9/ 0.9	0.9/ 0.9
C	2	0.8/ 0.8	0.9/ 0.9	0.8/ 0.8	0.8/ 0.8	0.8/ 0.8	0.8/ 0.8
C	3	0.7/ 0.8	1.0/ 0.8	0.5/ 0.7	0.8/ 0.8	0.6/ 0.8	0.8/ 0.8
C-m	1	6.5/ 6.6	11.0/ 11.1	3.6/ 3.7	3.8/ 4.6	6.5/ 6.6	4.4/ 4.7
C-m	2	6.6/ 6.6	11.4/ 11.4	4.2/ 4.6	5.0/ 4.7	4.1/ 6.6	5.8/ 5.8
C-m	3	6.6/ 6.6	11.2/ 11.3	3.8/ 4.0	5.8/ 4.2	6.6/ 6.6	5.8/ 5.8
D	1	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	2	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	3	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	1	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞
D-m	2	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞
D-m	3	∞/ ∞	∞/ ∞	∞/ ∞	∞/ ∞	∞/ 29.0	∞/ ∞
E	1	1.4/ 1.3	1.4/ 1.4	1.4/ 1.3	1.4/ 1.3	1.4/ 1.2	1.4/ 1.3

E	2	1.9/ 1.8	2.0/ 1.8	1.8/ 1.7	1.9/ 1.7	1.9/ 1.7	1.9/ 1.7
E	3	1.0/ 1.0	1.0/ 1.0	1.0/ 1.0	1.0/ 0.9	0.6/ 0.5	1.0/ 0.9
E-s	1	0.6/ 0.5	0.6/ 0.6	0.6/ 0.5	0.6/ 0.5	0.6/ 0.3	0.6/ 0.5
E-s	2	0.6/ 0.5	0.6/ 0.6	0.6/ 0.5	0.6/ 0.5	0.6/ 0.5	0.6/ 0.5
E-s	3	0.4/ 0.4	0.4/ 0.5	0.4/ 0.4	0.4/ 0.5	0.4/ 0.4	0.4/ 0.5

List of Figures

Figure 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). 31

Figure 2 'Headset' condition. 31

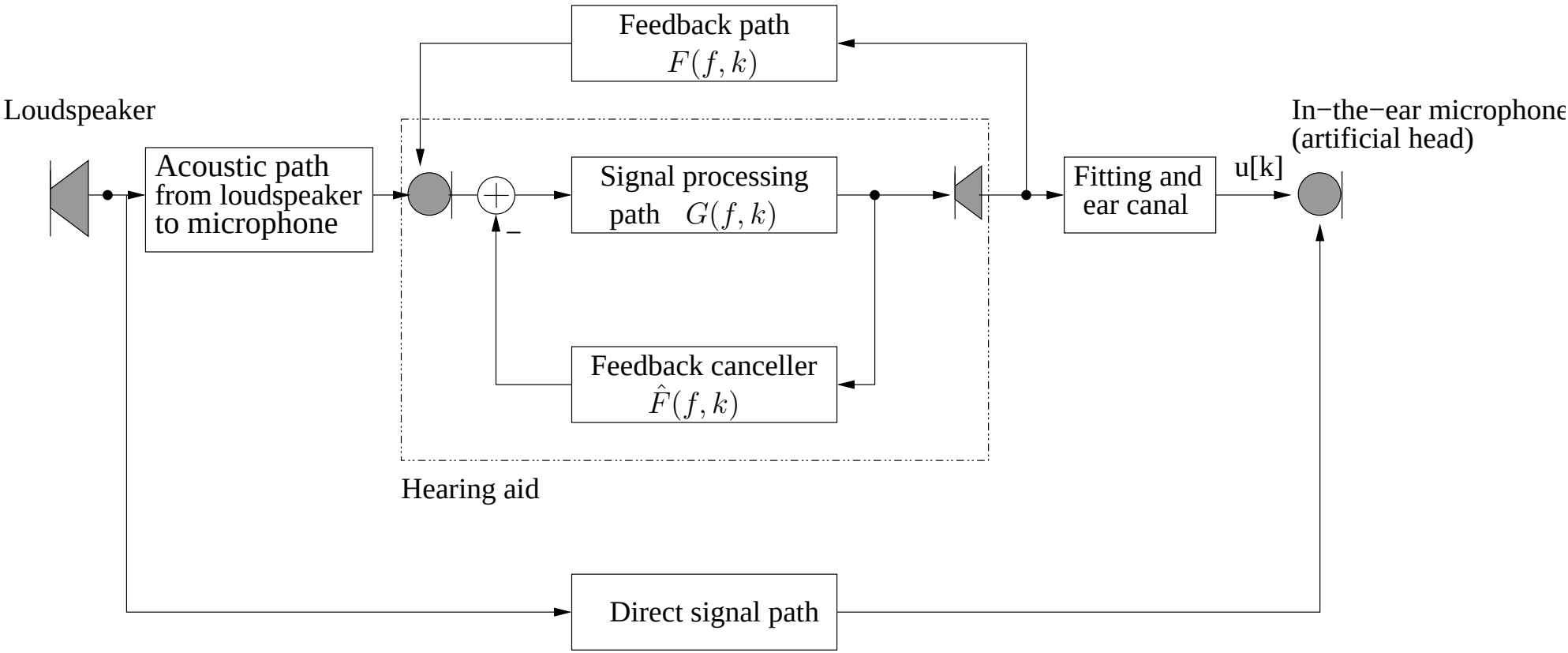
Figure 3 Set-up of the tracking experiment. An office lamp is positioned close to the left ear of the artificial head. 33

Figure 4 Hearing aid output power in dB SPL per frequency for a flat-spectrum input signal of 60 dBA. 33

Figure 5 Added stable gain for the white noise (a),(b) and the HINT noise signal (c),(d) 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. 34

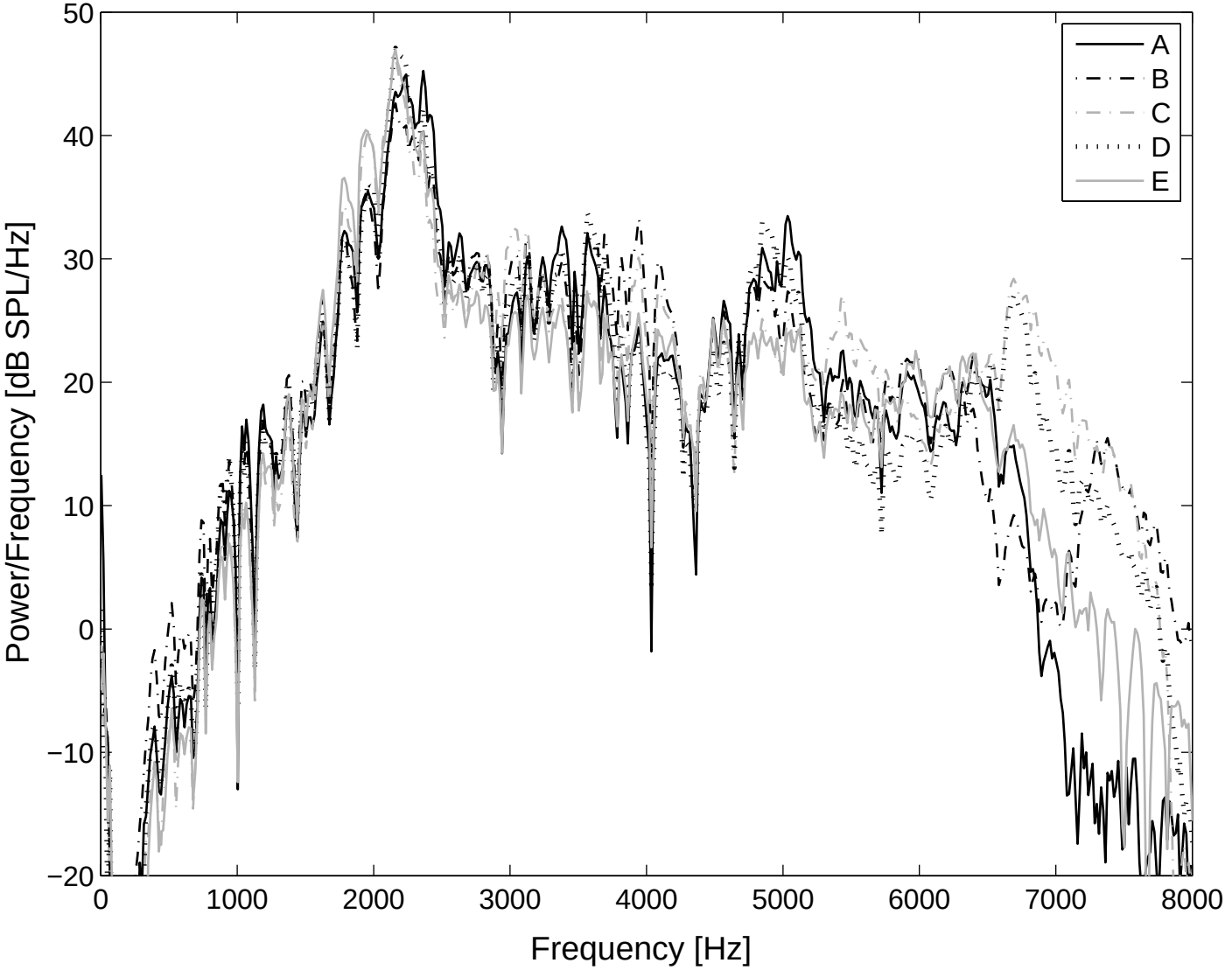
Figure 6 Added stable gain for the speech signal (a), (b) and the opera signal (c), (d) in the 'Normal' condition (a),(c) and the 'Handset' condition (b),(d). 35

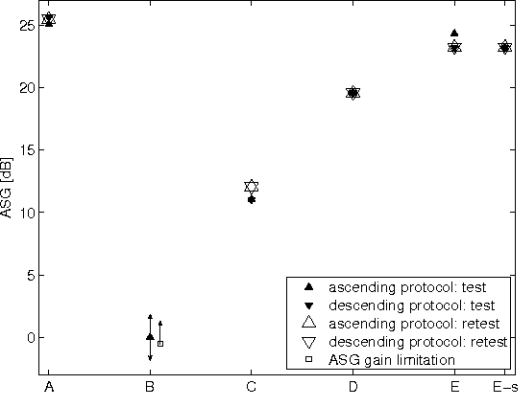
Figure 7 Amplitude frequency response of the feedback path of hearing aid E in the start (a) and end (b) position for the three trials. 36


Feedback path
$F(f,k)$
Loudspeaker
In−the−ear microphone
(artificial head)
Acoustic path
from loudspeaker
to microphone
Signal processing
path $G(f,k)$
Fitting and
ear canal
u[k]
−
Feedback canceller
$\hat{F}(f,k)$
Hearing aid
Direct signal path

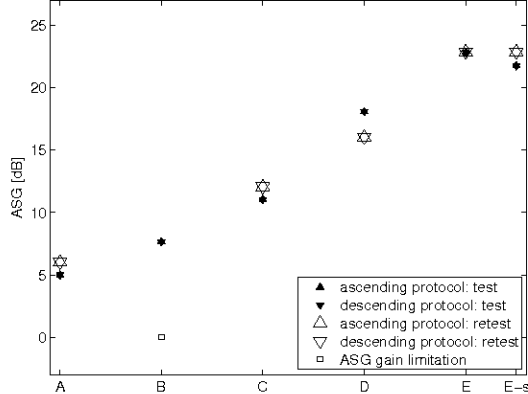



SMART

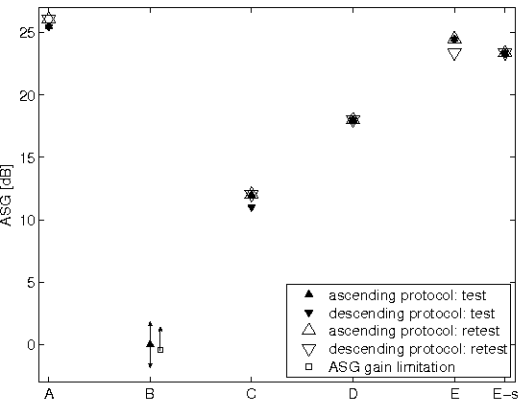

A
B
C
D
E
Power/Frequency [dB SPL/Hz]
Frequency [Hz]
50
40
30
20
10
0
−10
−20
0
1000
2000
3000
4000
5000
6000
7000
8000


ASG [dB]
25
20
15
10
5
0
A
B
C
D
E
E–s
ascending protocol: test
descending protocol: test
ascending protocol: retest
descending protocol: retest
ASG gain limitation

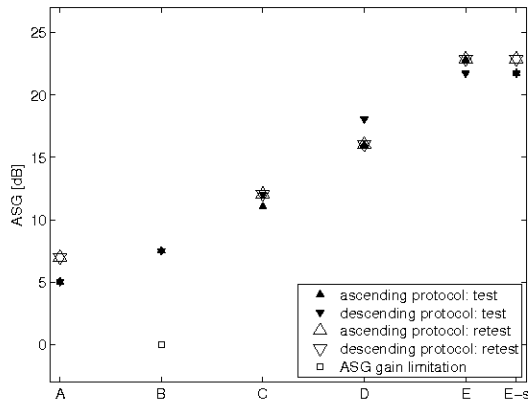
(a)


ASG [dB]
25
20
15
10
5
0
A
B
C
D
E
E–s
ascending protocol: test
descending protocol: test
ascending protocol: retest
descending protocol: retest
ASG gain limitation

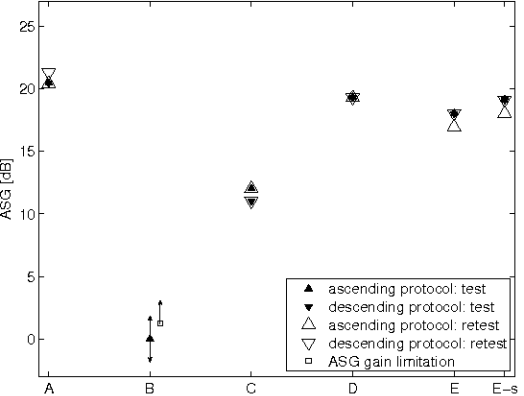
(b)


ASG [dB]
25
20
15
10
5
0
A
B
C
D
E
E–s
ascending protocol: test
descending protocol: test
ascending protocol: retest
descending protocol: retest
ASG gain limitation

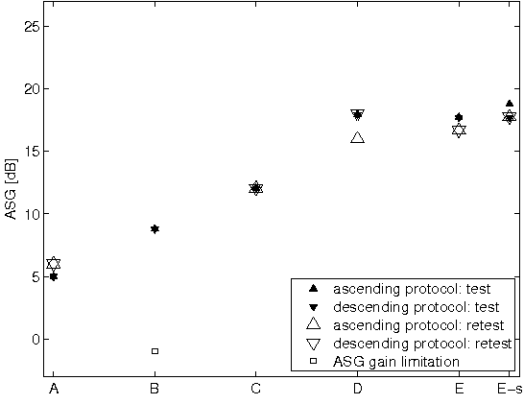
(c)


ASG [dB]
25
20
15
10
5
0
A
B
C
D
E
E–s
ascending protocol: test
descending protocol: test
ascending protocol: retest
descending protocol: retest
ASG gain limitation

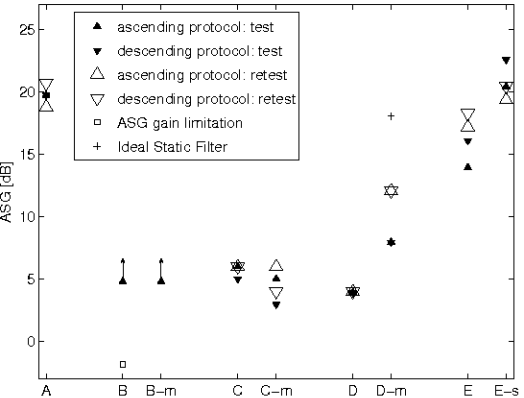
(d)


ASG [dB]
25
20
15
10
5
0
A
B
C
D
E
E–s
ascending protocol: test
descending protocol: test
ascending protocol: retest
descending protocol: retest
ASG gain limitation

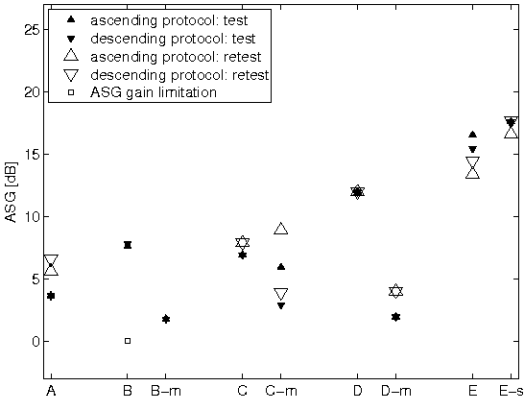
(a)


ASG [dB]
25
20
15
10
5
0
A
B
C
D
E
E–s
ascending protocol: test
descending protocol: test
ascending protocol: retest
descending protocol: retest
ASG gain limitation

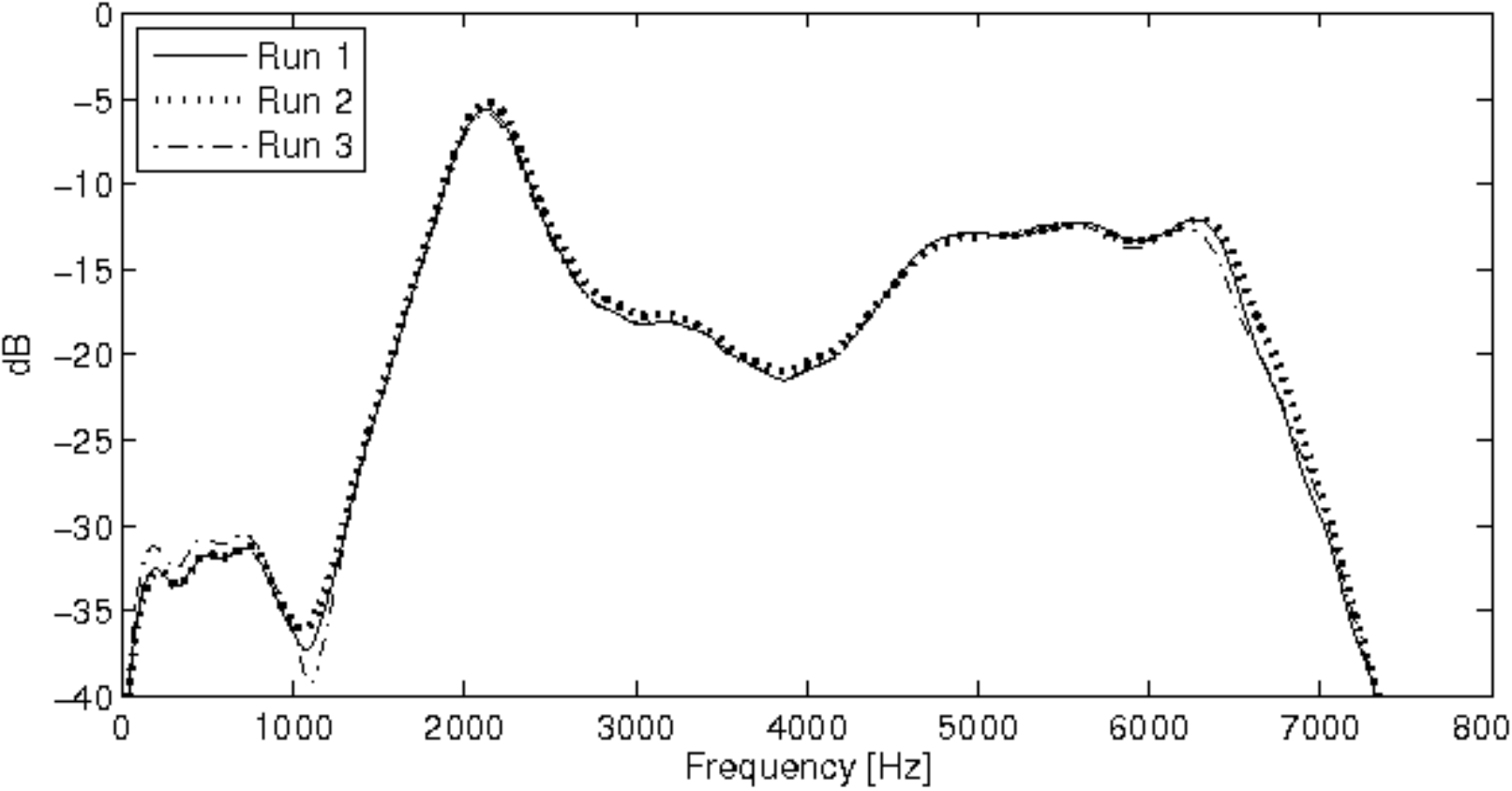
(b)


ASG [dB]
25
20
15
10
5
0
A
B
B–m
C
C–m
D
D–m
E
E–s
ascending protocol: test
descending protocol: test
ascending protocol: retest
descending protocol: retest
ASG gain limitation
Ideal Static Filter

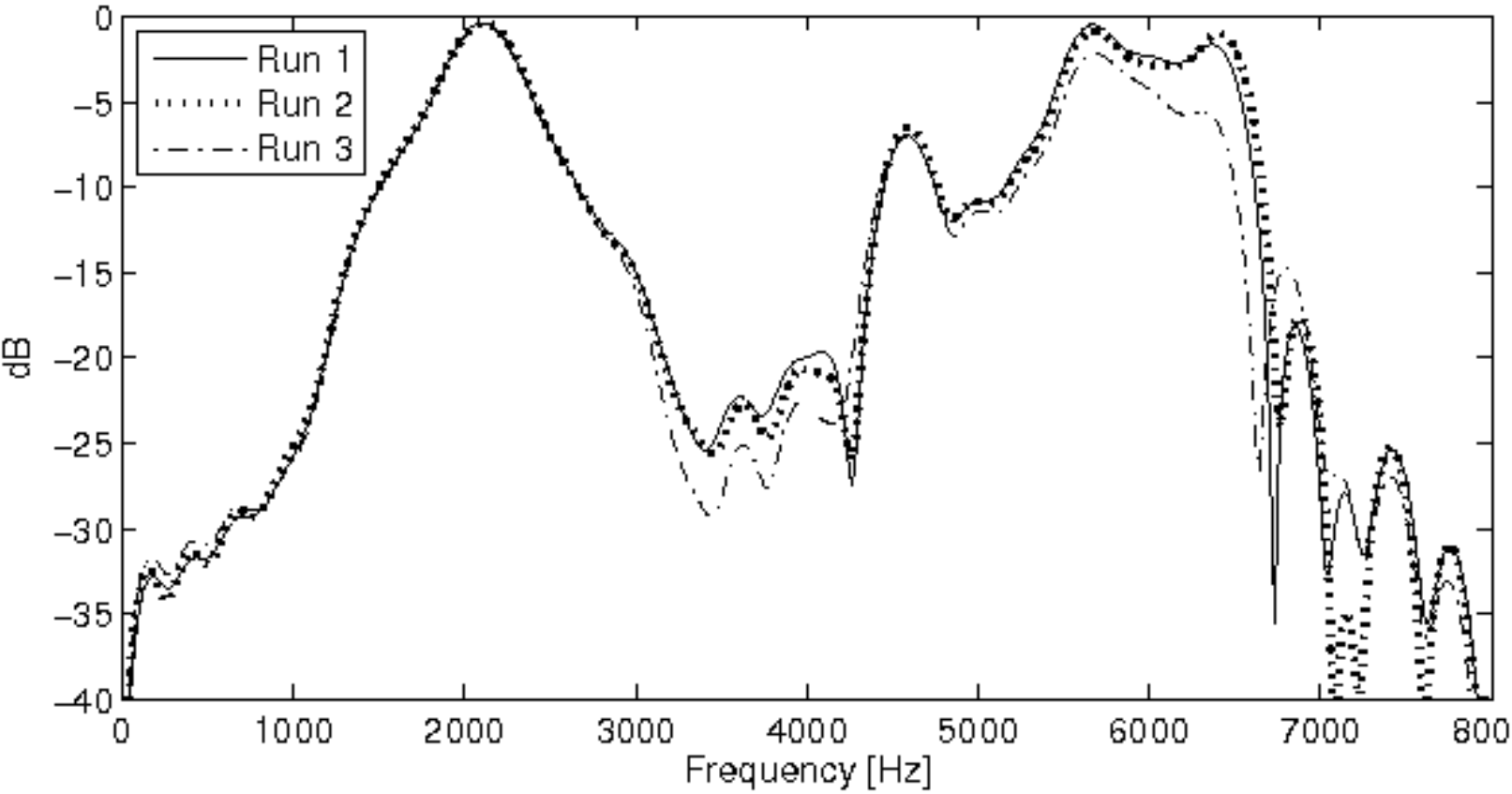
(c)


ASG [dB]
25
20
15
10
5
0
A
B
B–m
C
C–m
D
D–m
E
E–s
ascending protocol: test
descending protocol: test
ascending protocol: retest
descending protocol: retest
ASG gain limitation

(d)


Run 1
Run 2
Run 3
dB
0
−5
−10
−15
−20
−25
−30
−35
−40
0
1000
2000
3000
4000
5000
6000
7000
8000
Frequency [Hz]

(a)


Run 1
Run 2
Run 3
dB
0
−5
−10
−15
−20
−25
−30
−35
−40
0
1000
2000
3000
4000
5000
6000
7000
8000
Frequency [Hz]

(b)