Literature review on benchmarking of feedback suppression algorithms

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1 Introduction

This document gives an overview of the literature on the evaluation of feedback suppression algorithms for hearing aid applications. The three major objectives of a feedback suppression algorithm, which have been assessed in the literature, are:

- The feedback suppression algorithm should provide a *high added stable gain* (ASG) compared to the hearing aid system without feedback suppression.
- The feedback suppression algorithm should preserve a *good sound quality* and have a *low susceptibility to tonal signals*.
- The feedback suppression algorithm should be able to *quickly adjust to feedback path changes* so that the possible duration of instability is minimal.

For all three objectives, objective and/or subjective evaluation procedures have been proposed. Section 2 and Section 3 summarize the literature on objective evaluation procedures, while Section 4 briefly discusses subjective evaluation procedures.

2 Objective performance measures

Objective evaluation procedures (such as the assessment of the added stable gain) are based on objective performance measures. This section briefly summarizes the measures that have been proposed in the literature. The measures are classified into

- Broadband energy measures
- Measures for detecting oscillations
- Signal distortion measures

Some measures are based on the hearing aid output only, while other measures compare the hearing aid output with a desired output signal, i.e., the output that would be obtained in the absence of acoustic feedback. In the evaluation of commercial hearing aids, the desired output for example equals an amplified[1] version of the hearing aid output with the feedback canceller switched of and a gain setting far below instability.

In the definition of the measures, the following notation will be used. The output of the feedback suppression algorithm is referred to as $u[k]$. The desired output signal is denoted with $d[k]$. The power spectral density (PSD) of a signal $u[k]$ is denoted with $P_u(f)$. The expectation operator is $\varepsilon\{.\}$. The maximum stable gain with the feedback suppression algorithm disabled is MSG_{off}, the maximum stable gain with the feedback suppression algorithm enabled is MSG_{on}.

2.1 Broadband energy measures

Instability results in a large increase in output signal level compared to the level measured in the absence of feedback. Possible measures to detect instability and assess the amount of feedback are the output to desired signal energy ratio (ODR) and the desired signal to feedback energy ratio.

[1]To compensate for the difference in the studied gain setting and the gain setting below instability.

2.1.1 Output to desired signal energy ratio (ODR)

The output to desired signal energy ratio (ODR) is defined as

$$\mathrm{ODR} = 10\log_{10}\frac{\varepsilon\{u^2[k]\}}{\varepsilon\{d^2[k]\}}. \tag{1}$$

Instability can be assumed, if the ODR exceeds a certain threshold > 0 dB (e.g., 10 dB) [6, 16, 17].

2.1.2 Desired signal to feedback energy ratio (SFR)

To quantify the amount of feedback, the desired signal to feedback energy ratio (SFR) can be used [6]

$$\mathrm{SFR} = 10\log_{10}\frac{\varepsilon\{d^2[k]\}}{\varepsilon\{(d[k]-u[k])^2\}}. \tag{2}$$

2.2 Measures for detecting oscillations

In [4, 15], objective criteria to detect oscillations are defined, referred to as the power concentration ratio (PCR) and the hearing aid transfer function variation criterion (TVC), respectively. The proposed criteria are only defined for a white noise input signal. Possible extensions towards colored input signals are discussed [17].

2.2.1 Transfer function variation criterion (TVC)

Definition

In [15], the hearing aid transfer function with the increased gain is measured by sending white noise through an external loudspeaker. The hearing aid transfer function in the stable stage acts as the reference hearing aid transfer function. The gain difference between the increased gain and the stable gain is applied to the reference hearing aid transfer function. The difference in dB between the hearing aid transfer function and the gain compensated reference hearing aid transfer function is computed and is called transfer function variation function (TVF). The transfer function variation is limited to the frequency range where feedback is assumed to occur (e.g., from 1 kHz to 5 kHz). Using TVF, the transfer function variation criterion (TVC) is defined as

$$\mathrm{TVC} = \max_f(|TVF(f)|). \tag{3}$$

Instability is assumed if TVC $\geq$ 10 dB. When the TVC lies between 5 dB and 10 dB, the hearing aid is assumed to be in the sub-oscillatory stage.

Advantages

- TVC does not depend on the hearing aid frequency response.

Disadvantages

- Only the largest peak in the TVF is taken into account.
- The identification of the TVF assumes a white input signal

Possible extension

- Spectrally colored input signals

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

$$\begin{aligned} TVF(f) &= 10\log_{10}\left(\frac{P_u(f)}{P_d(f)}\right) \\ TVC &= \max_f(|TVF(f)|). \end{aligned} \tag{4}$$

  where $P_u(f,k)$ is the output power spectral density (PSD) and $P_d(f,k)$ is the PSD of the desired output $d[k]$. To only take into account the frequencies above the noise floor (i.e., level of the environment noise and internal noise), the PSDs $P_u(f,k)$ and $P_d(f,k)$ can be constrained as follows:

$$\begin{aligned} P_u(f) &= \max(P_u(f), \alpha P_n(f)), \\ P_d(f) &= \max(P_d(f), \alpha P_n(f)), \end{aligned} \tag{5}$$

  where $P_n(f)$ is the PSD of the environment and internal noise $n[k]$ and $\alpha > 1$.

2.2.2 Power concentration ratio (PCR)

Definition

In [4], the power concentration ratio, i.e., the degree to which a large amount of power is concentrated at a small number of frequencies in the hearing aid output, is introduced for detecting oscillation. The hearing aid output signal is recorded while presenting a white noise signal to the hearing aid. The output PSD is divided into frequency bins with a bandwidth of 23 Hz. The PCR is defined as the ratio of the power in the five frequency bins f_{c_i} with the highest power to the total power:

$$\text{PCR} = \frac{\sum_{i=1}^{5} P_u(f_{c_i})}{\sum_{i=1}^{N_{FFT}} P(i\frac{f_s}{N_{FFT}})}, \tag{6}$$

where N_{FFT} equals the FFT-size and f_s equals the sampling frequency. The PCR ranges from 0 to 1. The higher the PCR, the more peaky the output frequency response.

Disadvantages

- The PCR depends on the PSD of the input signal. The input signal should be white noise or at least not have any spectral peaks
- The PCR depends on the hearing aid frequency response. Resonances in the hearing aid frequency response can be falsely identified as oscillation frequencies.

Possible extensions

- To *remove the dependency of PCR on the hearing aid frequency response*, the output power spectral density can be normalized with the output power spectral density measured for a gain setting far below the maximum stable gain with the feedback canceller disabled (MSG_{off}) [4].
- To be applicable to *spectrally colored input signals*, the PCR measure can be modified as follows [17]:

  1. First, the oscillation frequencies are detected as the frequencies where the transfer function variation $TVC(f)$ (as defined in 4) equals or exceeds 6 dB. The fraction of the total power $P_u(f)$ of $u[k]$ that is located at the five (or less) strongest oscillation frequencies is computed and referred to as $\text{PCR}_u(k)$.
  2. In the absence of feedback, the detected oscillation frequencies may already have a large PCR due to the non-white PSD of the input signal or due to resonances in the hearing aid response. To reduce the PCR dependency on the input signal PSD and the hearing aid response, the fraction of the total power $P_d(f)$ of the desired signal $d[k]$ that is located at the detected oscillation frequencies in the absence of feedback is computed, referred to as PCR_d.
  3. The difference ΔPCR
  $$\Delta\text{PCR} = \text{PCR}_u - \text{PCR}_d \tag{7}$$
  is used as a measure for the presence of oscillations.

2.3 Spectral signal distortion

To assess the signal distortion of the hearing aid output $u[k]$, a comparison can be made between the PSD of the output signal $u[k]$ and PSD the desired output signal $d[k]$ using a spectral distortion measure [14, 17, 18].

2.3.1 Frequency-weighted log-spectral signal distortion [dB]

An example of a spectral distortion measure is the frequency-weighted log-spectral signal distortion $\text{SD}(k)$, which has been defined within the HearCom project:

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

$P_u(f,k)$ and $P_d(f,k)$ are the short-term PSD of the hearing aid output $u[k]$ and the desired output $d[k]$, respectively, computed using frames of 16 ms. The frequency-weighting factor $w_{ERB}(f)$ gives equal weight for each auditory critical band between $f_l = 300$Hz and $f_u = 6500$Hz, defined by the equivalent rectangular bandwidth (ERB) of auditory filters [13].

The distortion measure should not be affected by differences in the environment noise between the recordings of $u[k]$ and $d[k]$. Hence, only the frequencies that are sufficiently above the environment noise level should be taken into account. A possible solution is to constrain the short-term spectra $P_u(f,k)$ and $P_d(f,k)$ as in (5). In addition, signal frames should only included in the computation of the average distortion when their level is sufficiently above the noise floor.

2.4 Other measures

2.4.1 Entrainment

When an adaptive feedback canceller is presented with a spectrally colored input signal, distortion artifacts may be generated (such as addition of tones, modulation-type distortion). This is referred to as entrainment [12]. In [12], a perceptual correlate of overall entrainment level is calculated based on the loudness and the sharpness [19] of the spectral difference between the short-term spectra of $u[k]$ and $d[k]$.

2.4.2 Adaptation time

Some feedback cancellers produce short tone bursts at input signal fluctuations [6, 7]. These tone burst were observed to be annoying. In [6, 7], the occurrence of tone bursts at input signal fluctuations is assessed through the adaptation time of the algorithm when presented with an input signal step. The short-term relative level of the output $u[k]$ to the desired output $d[k]$ is calculated. At the signal step, a decay time constant was fitted to the relative level function. The higher the time constant, the larger the amount of artifacts.

3 Objective evaluation procedures

Based on the objective measures of Section 2, objective procedures for evaluating feedback were proposed.

3.1 Set-up

The performance of feedback suppression algorithms can either be evaluated through simulations using measured acoustic feedback paths and hearing aid microphone recordings or in real-time based on a set-up with an artificial head. Typically, a sound signal is presented to the hearing aid through a loudspeaker located 1 meter from the center of the head.

When testing in real-time, reproducibility of the acoustic scenario should be ensured, e.g., by testing in a soundproof test booth with a low environment noise level. Presented test signals should be well above the environment noise level in the test booth and the hearing aid internal noise.

3.2 Added stable gain

An often used criterion to evaluate feedback suppression algorithms is the added stable gain (ASG) that is provided by the feedback canceller. The ASG equals

$$\mathrm{ASG} = \mathrm{MSG}_{on} - \mathrm{MSG}_{off}. \tag{9}$$

MSG_{off} is the maximum stable gain (in dB) of the hearing aid with the feedback canceller disabled and MSG_{on} is the maximum stable gain with the feedback canceller enabled. In the literature, two procedures for computing the added stable gain have been proposed.

3.2.1 Procedure 1: increasing/decreasing the gain setting of the hearing aid [4, 15, 17, 6, 5]

The most common procedure consists of increasing the hearing aid gain until feedback oscillation or instability occurs. In [4, 6], the ASG is measured under two different protocols: one in which the gain is gradually increased until oscillation starts ('feedback starting'), and one in which the gain is gradually decreased until oscillation stops ('feedback stopping'). In [6], the 'feedback starting' gain and 'feedback stopping' gain are found by an automatic adaptive procedure.

This procedure requires an objective criterion for instability. Examples in the literature are:

- In [4, 15], the ASG is determined for white noise as an input signal. Feedback oscillation is detected when the PCR exceeds a threshold value of 0.5 or the TVC exceeds a threshold value of 10 dB.
- In [6, 17, 16], the ASG is determined for realistic input signals such as real speech. Example criteria for instability are a threshold on the output to desired signal ratio (ODR), the feedback to desired signal ratio (FSR) or the adaptation time.

3.2.2 Procedure 2: impulse response measurements at different gain settings [12, 11]

In [12, 11], the ASG is determined through impulse response measurements at different gain settings. White noise is presented through a loudspeaker at 1 meter in front of an artificial head and recorded through a microphone in the ear-canal of the artificial head. The microphone recording in the ear-canal consists of three components:

- Acoustic leakage from the loudspeaker to the ear-canal

- Hearing aid response
- Acoustic feedback

To determine the feedback component, the impulse response from the loudspeaker to the ear-canal of the artificial head is measured at three different hearing aid gain settings:

- Hearing aid muted or switched off to compute the acoustic leakage. The frequency response $IR_{off}(f)$ from the loudspeaker to the microphone then equals
$$IR_{off}(f) = L(f) \tag{10}$$
with $L(f)$ the acoustic path from the loudspeaker to the ear-canal microphone.

- Hearing aid at low gain $G_{low}(f)$. The frequency response $IR_{low}(f)$ equals
$$IR_{low}(f) = L(f) + S(f)G_{low}(f)E(f) + S(f)G_{low}(f)F(f)G_{low}(f)E(f) + S(f)G_{low}(f)F(f)G_{low}(f)F(f)G_{low}(f)E(f) + \dots \tag{11}$$
where $S(f)$ is the acoustic path from the loudspeaker to the hearing aid microphone, $G_{low}(f)$ is the hearing aid frequency response at low gain, $F(f)$ is the acoustic feedback path and $E(f)$ is the acoustic path from the hearing aid receiver to the ear-canal microphone.

- Hearing aid at high gain $G_{high}(f)$. The frequency response $IR_{high}(f)$ equals
$$IR_{high}(f) = L(f) + S(f)G_{high}(f)E(f) + S(f)G_{high}(f)F(f)G_{high}(f)E(f) + S(f)G_{high}(f)F(f)G_{high}(f)F(f)G_{high}(f)E(f) + \dots \tag{12}$$

Based on these three impulse responses, an estimate of the hearing aid response $H_{HA}(f) = S(f)G(f)E(f)$ and the first feedback $H_{1FB}(f) = S(f)G(f)F(f)G(f)E(f)$ is made. This assumes that the first feedback can easily be separated from the higher-order feedback components (i.e., sufficient delay in $G(f)$, low reverberation). Based on $H_{HA}(f)$ and $H_{1FB}(f)$ the gain margin $G(f)F(f)$ as a function of frequency is estimated.

Advantages

- Fast computation of ASG
- No need for instability or oscillation criteria

Disadvantages

- Limited to white noise input
- Accuracy of feedback estimate if hearing aid delay is small or in the presence of reverberation?
- The feedback suppression performance is only assessed at hearing aid gain setting. However, the accuracy of the feedback path estimate of a feedback canceller may depend on the gain setting.

3.2.3 General remarks

1. Choice of input signal

   The ASG is often computed for white noise as an input signal. In practice, the input signals to the hearing aid are spectrally colored and modulated. Spectral colouration and signal fluctuations may both affect the performance of the feedback canceller. As a result, the ASG for white noise will generally be an overestimation of the actual added stable gain that can be used in real-life. A better estimate could be achieved when using a realistic sound signal (i.e., with spectral coloration and modulation).

2. Hearing aid gain setting versus actual gain

   When computing the ASG, the hearing aid gain setting is often used. However, some feedback suppression algorithms (e.g., notch filter) may achieve stability at the expense of reducing the gain in specific frequency regions. When computing ASG, the actual gain that is provided by the hearing aid should thus be measured!

3. A high ASG does not necessarily guarantee a good sound quality at all gain settings. A comparison of algorithms based on ASG only is therefore insufficient.

3.3 Sound quality and susceptibility to tonal signals

In addition to ASG, the sound quality of the hearing aid output should be assessed for realistic sound signals (e.g., speech, music) and at different gain settings.

3.3.1 Sound quality as a function of gain setting

The hearing aid should provide a good sound quality at all gain settings. A possible procedure to assess sound quality is to record the hearing aid output at different gain settings for several realistic input signals (such as speech, music) and to compute objective measures for distortion or the presence of oscillation (cf. Section 2) as a function of the hearing aid gain setting [18, 17].

3.3.2 Susceptibility to tonal signals

Adaptive feedback cancellation algorithms may especially encounter problems for tonal input signals. In [4, 12, 11], procedures for quantifying the susceptibility to tonal signals are proposed.

Extraneous frequency ratio (EFR)

In [4], the extraneous frequency ratio (EFR) is defined to measure artifacts resulting from tonal input signals (such as the suppression of the input signal and the generation of extraneous tones due to oscillation). EFR measures the output power at extraneous frequencies as a percentage of the output power at the input frequency. EFR is measured for tones of 0.5 kHz, 1 kHz, 2 kHz and 4 kHz at MSG_{off} and MSG_{off}+4 dB [4].

Method:

- Enable feedback suppression algorithm; Record output in quiet for 10 sec.
- Present tone and record output for 10 sec.
- Subtract power spectrum in quiet from power spectrum with tonal input. Compute EFR.

Entrainment

In [12, 11], the robustness to periodic input signals is assessed by computing the entrainment when presenting the hearing aid with music, machine noise, tonal and human sounds. The entrainment was computed for a single hearing aid gain setting of 5dB below MSG_{off}.

3.4 Tracking performance

A third objective of a feedback suppression algorithm is robustness to changes in the feedback path, in particular when these changes result in an unstable hearing aid system. To ensure repeatability of the feedback path change, the change should be introduced through a mechanically controlled apparatus [12]. When evaluating the algorithms through simulations, a feedback path change can be simulated using a time-varying feedback path model [18].

3.4.1 Oscillation time / time required to recover from instability

In [4], a sudden change in feedback path was introduced by rapidly placing a hat on the manikin's head. The hat was placed manually, so it was not possible to reproduce the feedback path change (no precise control over timing and positioning). The hearing aid gain was set so as to require the algorithm to provide 4 dB of ASG with the hat in place, i.e., at $\text{MSG}_{off,hat}$+ 4 dB. The duration of oscillation based on a time-varying PCR. Instead of PCR, also other measures could be used, e.g., the output to desired signal energy ratio.

An alternative method to assess the time required to recover from instability is to initialize the feedback suppression system in an unstable condition (e.g., by setting the gain above MSG_{off} and initializing the feedback canceller coefficients with zero) and to measure the duration it takes for the feedback suppression algorithm to recover from instability [16, 17].

3.4.2 Annoyance

In [12, 11], a reflective object is moved towards the mannikin's ear through a PC controlled motor while noise was played. The hearing aid gain was set 1 dB below the algorithm's MSG for the end position to ensure stability of the device under static conditions. Based on the loudness and sharpness of the hearing aid output, an overall annoyance level was derived.

3.5 Other issues

3.5.1 Repeatability

Repeatability of the results should be checked through repeated measurements.

3.5.2 Measurement of signal delay

A high processing delay in the hearing aid may improve the robustness to tonal input signals at the expense of introducing other unwanted side-effects (e.g., degraded intelligibility and sound quality by comb filtering effects between the direct sound to the eardrum and the delayed and amplified sound [9, 8]) [18]. Hence, when comparing algorithms, the signal delay should also be considered.

4 Subjective evaluation procedures

In [5, 7, 10, 3, 2, 1], feedback suppression methods were evaluated subjectively.

4.1 Sound quality at fixed gain settings

In [5, 10, 3], sound quality was assessed at fixed gain settings using speech and music as an input signal.

- In [10], the hearing aid output was compared to the desired output signal for three gain settings, i.e., MSG_{on}, MSG_{on}-3dB and MSG_{on}-6dB. Sound quality was assessed on a scale out of 10, where 10 corresponds to listening to the desired output signal, i.e., 1-4: unacceptable quality, 5-8: tolerable degradations, 9-10: very good to excellent quality.
- In [5], sound quality of the hearing aid output was assessed on a scale out of 10, i.e., 1: very annoying, 3: rather unpleasant, 5: neutral, 7: good and 9: very pleasant. The subjects were asked to rate the pleasantness of a 20 s speech stimuli for four gain values (MSG_{on}, MSG_{on}-3 dB and MSG_{on}-6 dB).
- In [3], a band-limited and broadband adaptive feedback canceller were compared using a "signed preference score". Comparison was done at the MSG_{off} and MSG_{off}+10 dB at signals levels of 65 and 80 dB A, both for speech and music. The order of the algorithms was randomized for each condition. The subjects were asked to pick the sound file with the better sound quality and to indicate how better this one is compared to the other on a 0-5 scale (0 = 'no difference', 1 = 'almost negligibly better', 5 = 'much better').

In [1], entrainment of two feedback canceller was evaluated using six stimuli known to cause artifacts, i.e., ambulance siren, car horn, choir music, pure tone at 3kHz, telephone ring, and whistling. For each stimulus, the hearing aid output with the feedback canceller enabled and disabled were presented in random order to the subjects. The subjects were asked to rate the strength of their preference for the sound quality in the first or the second presentation. The hearing aid gain setting was set below MSG_{off}.

4.2 Maximum tolerable gain

In [5, 7], the maximum tolerable gain of feedback suppression techniques were determined through listening tests for a speech signal. The subjects were asked to adjust the gain as high as possible while the sound quality is still acceptable. In [5], the subject was wearing the hearing aid and the amplified signal was presented to the subject, while in [7] the hearing aid was mounted on a dummy head and the non-amplified input signal was presented. The latter has the benefit of not measuring uncomfortable level thresholds, and allows evaluation and comparison with subjects of different amount of hearing loss.

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