

Hearing aid with adaptive feedback suppression

Field of the invention

5 The invention relates to the field of hearing aids. The invention, more specifically, relates to a hearing aid having an adaptive filter for suppressing acoustic feedback, a method of adaptively reducing acoustic feedback of a hearing aid and to an electronic circuit for a hearing aid.

10 Related prior art

 Acoustic feedback occurs in all hearing instruments when sounds leak from the vent or seal between the earmould and the ear canal. In most cases, acoustic feedback is not audible. But when the in-situ gain of the hearing aid is sufficiently
15 high, or when a larger than optimal size vent is used, the gain of the hearing aid can exceed the attenuation offered by the earmould/shell. The output of the hearing aid then becomes unstable and the once-inaudible acoustic feedback becomes audible, e. g. in the form of a whistling noise. For many users and people around such audible acoustic feedback is an annoyance and even an embarrassment. In addition,
20 hearing instruments that are at the verge of feedback, i. e. sub-oscillatory feedback, may influence the frequency characteristic of the hearing instrument and lead to intermittent whistling.

 Fig. 1 shows a simple block diagram of a hearing aid comprising an input
25 transducer or microphone transforming an acoustic input signal, a signal processor amplifying the input signal and generating an electrical output signal and an output transducer or receiver for transforming the electrical output signal into an acoustic output. The acoustic feedback path of the hearing aid is depicted by broken arrows, whereby the attenuation factor is denoted by β . If, in a certain frequency range, the
30 product of gain G (including transformation efficiency of microphone and receiver) of the processor and the attenuation β is close to 1, audible acoustic feedback occurs.

 To suppress such undesired feedback it is well known in the art to include an adaptive filter in the hearing aid to compensate for the feedback. The adaptive filter

estimates the transfer function from output to input of the hearing aid including the acoustic propagation path from the output transducer to the input transducer. The input of the adaptive filter is connected to the output of the hearing aid and the output signal of the adaptive feedback estimation filter is subtracted from the input transducer signal to compensate for the acoustic feedback. A hearing aid of this kind, disclosed e. g. in WO 02/25996 A1, is schematically illustrated in Fig. 2. The output signal from the signal processor 3 is fed to an adaptive feedback estimation filter 5, which is controlled by a filter control unit 6. The adaptive feedback estimation filter constantly monitors the feedback path providing an estimate of the feedback signal and producing an output signal which is subtracted from the processor input signal in order to reduce, or in the ideal case to eliminate, acoustic feedback in the signal path of the hearing aid.

An overview of adaptive filtering is given in the textbook of Philipp A. Regalia: "Adaptive IIR Filtering in Signal Processing and Control", published in 1995.

One problem associated with adaptive feedback cancelling is a bias introduced by the feedback prediction model itself through narrow band signals included e.g. in speech or music. The correlation analysis of the adaptive feedback estimation algorithm is based on the assumption that a feedback signal (oscillation) is a highly correlated version of the original signal. When signal components of the external hearing aid input, e.g. contained in speech or music, are narrow band signals, a bias is introduced in the feedback prediction model and the external narrow band signal components are removed from the hearing aid signal path by the feedback suppression algorithm.

To solve this problem Siqueira and Alwan propose in "Steady-State Analysis of Continuous Adaptation in Acoustic Feedback Reduction Systems for Hearing Aids", IEEE transactions on speech and audio processing, Vol. XIII, no. 4, pages 443-453, July 2000, the use of a delay in the forward or cancellation path of the hearing aid in order to reduce the bias introduced by narrow band input signals. This delay, however, does still not make a sinusoid signal unpredictable by the feedback cancellation algorithm.

From US 2003/0053647 A1 to Kates a hearing aid is known comprising a cascade of adaptive notch filters applied to the error signal before a signal is supplied to the feedback path estimation algorithm. The series of notch filters removes the narrow band signal components from the feedback estimation algorithm so that the mean square error (MSE) calculation in the adaptive feedback estimation filter does not take into account the external narrow band signal components and interpolates the feedback path model over the absent frequencies.

To ensure a correct mean square error minimization process with respect to the notch filtered error signal the input signal of the adaptive feedback estimation filter must be filtered with copies of the adaptive notch filters before it is fed to the adaptation algorithm.

Furthermore, the notch filters are optimized to cancel the narrow band signal components by minimizing a cost function of the notch filter output.

In order to remove a plurality of narrow band signal components a plurality of notch filters are required. With an increasing number of notch filters for different frequencies, however, the computational costs increase and mutual influence of the different notch filters may occur.

Summary of the invention

It is therefore an object of the present invention to provide a hearing aid with adaptive feedback cancellation and a method of adaptively reducing acoustic feedback of a hearing aid having improved feedback-cancellation properties at optimized calculation costs.

According to a first aspect of the invention the problem is solved by a hearing aid comprising an input transducer for deriving an electrical input signal from an acoustic input, a signal processor for generating an electric output signal, an output transducer for transforming the electrical output signal into an acoustic output, an adaptive estimation filter for generating a feedback estimation signal, at least one first adaptive narrow-band filter for narrow band-filtering an input signal of the signal

processor, at least one second adaptive narrow-band filter for narrow band-filtering a reference signal corresponding to an input signal of the adaptive estimation filter, and an adaptation mechanism for updating the filter coefficients of the adaptive estimation filter based on the output signals of the first and second narrow-band filters, wherein the at least one second adaptive narrow-band filter is configured to derive its output signal from a gradient of the output signal of the at least one first narrow-band filter.

To ensure a correct cost function (e.g. mean square error) minimization process of the narrow band-filtered error signal (input signal of hearing aid processor), the input signal of the adaptive estimation filter must also be filtered with copies of the adaptive narrow-band filter(s) before it is fed to the filter control unit. The narrow-band filtered reference signal is according to a first aspect of the present invention derived from a gradient with respect to the filter coefficients of the feedback estimation filter of the narrow-band filtered error signal output by the at least one first narrow-band filter.

Preferably the at least one first adaptive narrow-band filter and the at least one second adaptive narrow-band filter minimize a cost function of its output signal, e.g. the signal energy or a signal norm. The minimization may be performed by a least mean square type or similar algorithm.

Alternatively it is possible instead of minimizing the narrow-band filter output to maximize the output of a hypothetical resonator of a given frequency corresponding to the center frequency of the adaptive narrow-band filter and having a constrained pole radius.

In order to optimize the frequency adaptation of the narrow-band filter a combined gradient may be employed, wherein a narrow band gradient is calculated if the center frequency adaptation rate of the filter is below a predetermined threshold value and a broader band gradient is calculated if the center frequency adaptation rate of the narrow-band filter is above this threshold value.

The adaptive estimation filter preferably employs a least mean square (LMS) algorithm for feedback reduction.

5 The adaptation mechanism advantageously carries out a cross correlation processing of the narrow-band filtered error signal with the narrow-band filtered reference signal.

As adaptive narrow-band filters one or preferably a plurality of adaptive notch filters with predetermined frequency width r may be employed, wherein the plurality
10 of notch filters have different adaptive center frequencies $c(n)$.

The first aspect of the present invention also provides a method of adaptively reducing the acoustic feedback of a hearing aid comprising an input transducer for deriving an electrical input signal from an acoustic input, a signal processor for
15 generating an electrical output signal and an output transducer for transforming the electrical output signal into an acoustic output, the method comprising the steps of generating a feedback estimation signal, deriving an error signal by subtracting the feedback estimation signal from the electrical input signal, narrow band-filtering the error signal and a reference signal corresponding to a feedback estimation input
20 signal, and adapting feedback estimation filter coefficients based on the narrow band-filtered signals, wherein the narrow-band filtered reference signal is derived from a filter gradient of the narrow-band filtered error signal.

According to a second aspect of the present invention a hearing aid is provided
25 comprising an input transducer for deriving an electrical input signal from an acoustic input, a signal processor for generating an electric output signal, an output transducer for transforming the electrical output signal into an acoustic output, an adaptive estimation filter for generating a feedback estimation signal, at least one first adaptive narrow-band filter for narrow band-filtering an input signal of the signal
30 processor, at least one second adaptive narrow-band filter for narrow band-filtering a reference signal corresponding to an input signal of the adaptive estimation filter, and an adaptation mechanism for updating the filter coefficients of the adaptive estimation filter based on the output signals of the first and second narrow-band

filters, wherein the first and second set of adaptive narrow-band filters are configured to minimize a single shared cost function.

For the plurality of narrow-band filters forming the first filter set for filtering the error signal and for the plurality of narrow-band filters forming the second filter set for filtering the reference signal one respective shared cost function is minimized thus improving the overall narrow band signal suppression. The shared cost function makes each narrow-band filter aware of the effectiveness of all the narrow-band filters.

In order to reduce the calculation costs of the gradient calculation a tree structure of the first set of narrow-band filters may be used. In this case the number of narrow-band filters is preferably 2^N ($N=2,3,4,5,\dots$).

Another possibility to reduce the computation costs of the gradient calculation is to perform these independently for every filter but at the same time using a shared error function for all filters of the set of narrow-band filters.

According to the second aspect of the present application there is furthermore provided a method of adaptively reducing an acoustic feedback of a hearing aid comprising an input transducer for deriving an electrical input signal from an acoustic input, a signal processor for generating an electrical output signal and an output transducer for transforming the electrical output signal into an acoustic output, the method comprising the steps of generating a feedback estimation signal, deriving an error signal by subtracting the feedback estimation signal from the electrical input signal, narrow band-filtering the error signal and a reference signal corresponding to a feedback estimation input signal in a plurality of filter stages having different adaptive center frequencies, and adapting estimation filter coefficients based on the narrow band-filtered error and reference signals, wherein the narrow-band filtering using a plurality of different adaptive center frequencies is performed minimizing a single shared cost function.

The invention, in a further aspect, provides a computer program as recited in claim 21 and an electronic circuit for a hearing aid as recited in claim 22.

The invention, in yet another aspect, provides a computer program as recited in claim 41 and an electronic circuit for a hearing aid as recited in claim 42.

5 Further specific variations of the invention are defined by the further dependent claims.

Brief description of the drawings

10 The present invention and further features and advantages thereof will become more readily apparent from the following detailed description of particular embodiments of the invention with reference to the drawings, in which:

Fig. 1 is a schematic block diagram illustrating the acoustic feedback path of a
15 hearing aid;

Fig. 2 is a block diagram showing a prior art hearing aid;

Fig. 3 is a block diagram showing a hearing aid to which the present application
20 may be applied;

Fig. 4 is a diagram illustrating the transfer function of a notch filter;

Fig. 5 is a flow chart illustrating a method of adaptively reducing the acoustic
25 feedback of a hearing aid according to an embodiment of the present invention;

Fig. 6 is a block diagram illustrating a set of adaptive narrow-band filters according to the prior art;

30 Fig. 7 illustrates a set of adaptive narrow-band filters according to an embodiment of the present invention;

Fig. 8 illustrates a set of adaptive narrow-band filters according to a further embodiment of the present invention;

Fig. 9 is a block diagram illustrating the gradient calculation according to an embodiment of the present invention;

5 Fig. 10 is a block diagram illustrating the tree structure for gradient calculation according to a further embodiment of the present invention;

Fig. 11 is a diagram illustrating the sensitivity of two types of gradient filters; and

10 Fig. 12 is a diagram illustrating the sensitivity of three further gradient filters.

Detailed description of preferred embodiments

Fig. 3 is a schematic block diagram of a hearing aid having an adaptive filter for
15 feedback suppression to which the present application may be applied.

The signal path of the hearing aid comprises an input transducer or microphone
2 transforming an acoustic input into an electrical input signal, a signal processor or
amplifier 3 generating an amplified electrical output signal and an output transducer
20 (loudspeaker, receiver) 4 for transforming the electrical output signal into an acoustic
output. The amplification characteristic of the signal processor 3 may be non-linear
providing more gain at low signal levels and may show compression characteristics
as is well known in the art.

25 The electrical output signal or reference signal $u(n)$ is fed to an adaptive filter 5
monitoring the feedback path and comprising an adaptation algorithm 6 adjusting a
digital filter 5 such that it simulates the acoustic feedback path so providing an
estimate of the acoustic feedback. The adaptive estimation filter 5 generates an
output signal $s(n)$ which is subtracted from input signal $d(n)$ at summing node 7. In
30 the ideal case the feedback of feedback path β in Fig. 1 is therefore removed in
processor input signal or error signal $e(n)$.

The adaptive estimation filter 5 is designed to minimize a cost function as for
example the power of the error signal $e(n)$. The adaptive filter may be embodied (but

is not restricted to a K-tap finite impulse response (FIR) filter having adaptive coefficients $b_1(n)$ through $b_k(n)$. A power-normalized adaptive filter update for a sample n of the digital electrical signal can then be expressed as follows:

$$b_k(n+1) = b_k(n) + 2 \frac{\nu}{\sigma_d^2(n)} e(n) u(n-k) \quad (1)$$

wherein ν controls the rate of adaptation and $\sigma_d^2(n)$ is the average power in the feedback path signal $u(n)$. If the input of the adaptive filter is a pure (sine) tone the adaptive feedback cancellation system minimizes the error signal $e(n)$ by adjusting the filter coefficients $b_1(n)$ through $b_k(n)$ so that the output signal $s(n)$ has the same amplitude and phase as the input and will consequently cancel it at summing node 7.

To avoid this undesirable effect of cancelling narrow band components of non-feedback input signals it is known to use narrow-band filters such as notch filters 8, 9 for narrow-band filtering the error signal $e(n)$ as well as the processor output signal or reference signal $u(n)$. The adaptive narrow-band filters 8, 9 operate with mutually identical filter coefficients, i.e. the filter coefficients of narrow-band filter 8 are copied to narrow-band filter 9. In a variant of this embodiment, they are copied from 9 to 8. Both filters may consist of a cascade of filters connected in series to each other and having different adaptive center frequencies. The output signal of the first narrow-band filter, i.e. narrow-band filtered error signal $e_f(n)$ and the output signal of the second narrow-band filter, i.e. narrow-band filtered reference signal $u_f(n)$ are fed to adaptation mechanism 6 controlling the filter coefficients of adaptive error estimation filter 5. Adaptation mechanism 6 performs a cross correlation of its input signals $e_f(n)$ and $u_f(n)$.

Preferably the adaptive narrow-band filters 8, 9 are implemented by digital notch filters, having the transfer function

$$H(z) = \frac{1 - 2 \cos(\omega_0 / f_s) z^{-1} + z^{-2}}{1 - 2r \cos(\omega_0 / f_s) z^{-1} + r^2 z^{-2}} \quad (2)$$

in frequency domain z , wherein r is the pole radius of the notch filter, ω_0 the center frequency in radians, and f_s the sampling frequency. r preferably assumes values between 0,5 and 1 and in particular between 0,95 and 1. A schematic illustration of the transfer function of a notch filter is illustrated in Fig. 4.

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In recursive notation depending on sampling index n the notch filter 8 for error signal $e(n)$ can be expressed as follows

$$\left. \begin{aligned} x(n) &= e(n) - 2 \cdot r \cdot c(n) \cdot x(n-1) - r^2 \cdot x(n-2) \\ e_r(n) &= x(n) + 2 \cdot c(n) \cdot x(n-1) + x(n-2) \end{aligned} \right\} \quad \text{Notch filter} \quad (3)$$

10

wherein $x(n)$ is an output signal from filtering with just the pole pair and $e_r(n)$ is the result of additional filtering with the zero pair, wherein $c(n)$ is the adaptive notch frequency of the notch filter. The frequency adaptation is given by:

$$c(n+1) = c(n) - \frac{\mu}{p(n)} \cdot e_r(n) \cdot \nabla c(n)^2 \quad (4)$$

15

wherein μ determines the update speed of the center frequency of the notch and $p(n)$ is a power normalisation:

$$p(n) = \alpha \cdot p(n-1) + \nabla c(n)^2 \quad (5)$$

20

wherein α is a forgetting factor of the power normalisation and $\nabla c(n)$ is the gradient of the notch filter. This gradient can be calculated in different ways as is explained in the following:

25 (1) True gradient algorithm

The true gradient of a direct form II notch filter is calculated as follows:

$$\begin{aligned} g(n) &= (1-r) \cdot x(n-1) - r \cdot c(n) \cdot g(n-1) - r^2 \cdot g(n-2) \\ \nabla^t c(n) &= g(n) - r \cdot g(n-2) \end{aligned} \quad (6)$$

30

wherein $g(n)$ is the status of the gradient calculation. The true gradient provides a high signal sensitivity in the vicinity of the center frequency $c(n)$ but bears a comparatively high computational cost.

5 (2) Pseudo gradient algorithm

Another way to calculate an update method of $c(n)$ is the simplified pseudo gradient method. This algorithm is derived from the assumption that the first line of (3) can be ignored or regarded as pre-filtering of the
10 second line in (3) and hence the so-called pseudo gradient is calculated as follows:

$$\nabla^p c(n) = x(n-1) \quad (7)$$

15 Besides the lower computational cost compared with the true gradient method, the simplified pseudo gradient is characterized by its larger sensitivity to spectral energies in the periphery of the notch center frequency and hence its relative less sensitivity to the spectral envelope in the vicinity of the notch frequency. This is illustrated by the graph of
20 Fig. 11 showing the sensitivity of the true gradient and the pseudo gradient dependent on a sinusoid input frequency at a given selected notch center frequency of 8000 Hz, notch width of 500 Hz and notch radius $r = 0,995$. The pseudo gradient is advantageous having a narrow band signal component in the periphery of the current notch center
25 frequency, but if the notch has converged to the frequency of the narrow band signal component, it is more advantageous to use the true gradient as it is more accurate in its frequency estimate since it is less disturbed by signals in the periphery.

30 (3) Combined gradient

According to an aspect of the present invention a combined gradient is suggested which monitors some sort of mean pseudo gradient. If this is above a specified threshold the mean pseudo gradient is utilized

instead of the true gradient algorithm, which in turn is utilized below the threshold. A preferred embodiment is given below, which monitors the pseudo gradient with an exponential decaying time window:

$$\begin{aligned}
 5 \quad m(n) &= \lambda \cdot m(n-1) - \frac{\mu}{p_{ps}(n)} \cdot e_f(n) \cdot \nabla^p c(n) \\
 |m(n)| &> \beta ? \quad (8)
 \end{aligned}$$

wherein λ determines the forgetting factor of the exponential decaying time window of the monitored mean pseudo gradient drive $m(n)$ and β specifies the threshold value above which the pseudo gradient is utilized. That is if $|m(n)| > \beta$ then the pseudo gradient of formula (7) is utilized in the frequency update calculation of formula (4) and otherwise the true gradient given in formula (6) is utilized. Also, the respective gradients have to be inserted in the weighting factor calculation defined by (5). This combined filter or "pseudo to true gradient filter" (6) combines the advantages of both gradient algorithms discussed above, i.e. the better sensitivity of the pseudo gradient with respect to narrow band signal components in the periphery of the notch frequency and the higher accuracy of the true gradient close to the current center frequency $c(n)$.

According to the present invention the calculation of the narrow-band filtered reference signal $u_f(n)$ is needed to perform the calculation of the gradient $\nabla_{bk}(n)$ of the notch filtered error signal $e_f(n)$ with respect to the filter coefficients $b_1(n)$ through $b_k(n)$ of the adaptive feedback estimation filter 5 as is defined by the following formula:

$$\nabla_{bk}(n) = Z^{-1} \left(U(z) \left(\prod_{j=1}^M \frac{1 + \omega \cdot z^{-1} + z^{-2}}{1 + r c_j \cdot z^{-1} + r^2 \cdot z^{-2}} \right) z^{-k} \right) \quad (9)$$

Fig. 5 illustrates a particular embodiment of a method of adaptively reducing the acoustic feedback of a hearing aid according to the present invention.

In method step S1 an electrical input signal $d(n)$ is derived from the acoustic input of microphone 2. In subsequent method step S2 error signal $e(n)$ is derived at summing node 7 by subtracting feedback estimation signal $s(n)$ from input signal $d(n)$. Error signal $e(n)$ is then fed to signal processor 3 producing output signal $u(n)$ in step S5 which is then transformed into the acoustic output by receiver 4 in method step S9.

With the at least one narrow-band filter 8 a narrow-band filtered signal $e_f(n)$ of their error signal is calculated in method step S4. In subsequent step S6 the narrow-band filtered signal $u_f(n)$ of reference signal $u(n)$ is calculated in the at least one narrow-band filter 9 utilizing the narrow-band filter coefficients found in S4.

In step S7 the feedback estimation filter parameters of adaptive estimation filter 5 are adapted based on the cross correlation of narrow-band filtered signals $e_f(n)$ and $u_f(n)$. Adaptive estimation filter 5 then derives feedback estimation signal $s(n)$ in method step S8 which is fed to the negative input of summing node 7.

The adaptation algorithm performed by adaptive estimation filter 5 in method step S8 is preferably performed such that a cost function of the narrow-band filtered error signal $e_f(n)$ is minimized. This cost function may be the signal energy or a norm of the signal. Most commonly the mean square error (MSE) function is minimized resulting in the widely known least mean square (LMS) algorithm.

Narrow-band filters 8, 9 are preferably optimized to cancel narrow band signal components. This may be obtained by minimizing a cost function of the narrow-band filter output. This cost function may also be the MSE leading to an LMS type algorithm.

Instead of minimizing the output of the narrow-band filter it is alternatively possible to maximize the output of a hypothetical resonator with constrained pole radius. After maximizing the resonator output a notch may be constructed from the very same filter. A notch adaptation algorithm maximizing such resonator energy J can be derived as follows:

$$J = E[x^2(n)] = \text{MSE}$$

$$\frac{\partial J}{\partial c} = E \left[2 \cdot x(n) \cdot \frac{\partial x(n)}{\partial c} \right] \quad (\text{Adjust } c \text{ in the gradients direction as to increase } J) \quad (10)$$

The corresponding gradient is then expressed as follows:

5

$$\begin{aligned} \nabla^m c(n) &= \frac{\partial x(n)}{\partial c} \\ &= Z^{-1} \left(\frac{\partial X(z)}{\partial c} \right) = Z^{-1} \left(\frac{\partial \left(E(z) \cdot \frac{1}{1 + c \cdot r \cdot z^{-1} + r^2 \cdot z^{-2}} \right)}{\partial c} \right) \\ &= Z^{-1} \left(E(z) \cdot \frac{-r \cdot z^{-1}}{(1 + c \cdot r \cdot z^{-1} + r^2 \cdot z^{-2})^2} \right) \end{aligned} \quad (11)$$

10 wherein $E(z)$ is the Z-domain (frequency) representation of the notch input signal and Z^{-1} the inverse-z-transformation back into time-domain signal. In time domain dependent on index n the gradient is represented as follows:

$$g(n) = x(n) - r \cdot c(n) \cdot g(n-1) - r^2 \cdot g(n-2)$$

$$15 \quad \nabla^m c(n) = -r \cdot g(n-1) \quad (12)$$

wherein the notch filter is determined by equation (3) and the weighting function $p(n)$ and the frequency update $c(n+1)$ are given as follows:

$$20 \quad p(n) = \alpha \cdot p(n-1) + \nabla^m c(n)^2$$

$$c(n+1) = c(n) + \frac{\mu}{p(n)} \cdot x(n) \cdot \nabla^m c(n) \quad (13)$$

Similar to the simplified pseudo gradient discussed above a simplified pseudo gradient algorithm can be constructed if one constrains the notch's zeroes to prefilter the input of the adaptive notch. The gradient algorithm is in the following referred to as "pseudo maxres gradient":

$$J = E[e_f(n)^2]$$

$$\begin{aligned}
\frac{\partial J}{\partial c} &= E \left[2 \cdot e_f(n) \cdot \frac{\partial e_f(n)}{\partial c} \right]_{Pseudo \max res} \\
\nabla^{pm} c(n) &= \frac{\partial e_f(n)}{\partial c} \Big|_{Pseudo \max res} \\
&= Z^{-1} \left(\frac{\partial E_f(z)}{\partial c} \Big|_{Pseudo \max res} \right) = Z^{-1} \left(\overbrace{E(z)}^{Input} \cdot \overbrace{(1 + c \cdot z^{-1} + 1 \cdot z^{-2})}^{Zeroprefilter} \cdot \overbrace{\frac{\partial \left(\frac{1}{1 + r \cdot c \cdot z^{-1} + r^2 \cdot z^{-2}} \right)}{\partial c}}^{Maxresgradient} \right) \\
&= Z^{-1} \left(E(z) \cdot (1 + c \cdot z^{-1} + 1 \cdot z^{-2}) \cdot \frac{-r \cdot z^{-1}}{(1 + r \cdot c \cdot z^{-1} + r^2 \cdot z^{-2})^2} \right) \\
5 \quad &= Z^{-1} \left(E(z) \cdot \frac{1 + c \cdot z^{-1} + 1 \cdot z^{-2}}{1 + r \cdot c \cdot z^{-1} + r^2 \cdot z^{-2}} \cdot \frac{-r \cdot z^{-1}}{1 + r \cdot c \cdot z^{-1} + r^2 \cdot z^{-2}} \right) \\
&= Z^{-1} \left(E_f(z) \cdot \overbrace{\frac{-r \cdot z^{-1}}{1 + r \cdot c \cdot z^{-1} + r^2 \cdot z^{-2}}}^{pseudomaxresgradientfilter} \right) \tag{14}
\end{aligned}$$

The main difference between the pseudo maxres algorithm and the normal pseudo gradient algorithm discussed before is that the notch filtered signal can be used as the input to the gradient calculation filter. This can be observed in the frequency sensitivity plot as a dead zone just around the notch frequency (compare Fig. 12). The dead zone is inversely proportional to the radius coefficient r_{dz} . The pseudo maxres gradient filter is expressed as follows:

$$15 \quad \left. \begin{aligned} g(n) &= e_f(n) - r_{dz} \cdot c(n) \cdot g(n-1) - r_{dz}^2 \cdot g(n-2) \\ \nabla^{pm} c(n) &= -r_{dz} \cdot g(n-1) \end{aligned} \right\} \text{pseudo maxres gradient filter} \tag{15}$$

If $r_{dz} \rightarrow 1$ then the pseudo maxres gradient $\nabla^{pm} c(n)$ becomes identical with the pseudo gradient of equation (7). However, setting r_{dz} equal to 1 is not a numerically sound choice.

Similar as in the above described cases a true maxres gradient algorithm may be employed. When this algorithm is derived, a pseudo to true gradient filter is observed expressed by the following formulae:

$$\begin{aligned}
 g(n) &= e_f(n) - r_{dz} \cdot c(n) \cdot g(n-1) - r_{dz}^2 \cdot g(n-2) \\
 \nabla^{pm} c(n) &= -r_{dz} \cdot g(n-1) \\
 g(n) &= (1-r) \cdot \nabla^{pm} c(n) - r \cdot c(n) \cdot g(n-1) - r^2 \cdot g(n-2) \\
 \nabla^{tm} c(n) &= g(n) - r \cdot g(n-2) \\
 p(n) &= \alpha \cdot p(n-1) + \nabla^{tm} c(n) \\
 c(n+1) &= c(n) + \frac{\mu}{p(n)} \cdot x(n) \cdot \nabla^{pm} c(n)
 \end{aligned} \tag{16}$$

The sensitivities of the maxres gradient, the pseudo maxres gradient and the true maxres gradient are depicted in Fig. 12. The dead zone of the latter two gradient filters can be readily recognized in the plot.

As explained in detail before the adaptive narrow-band filter or in particular adaptive notch filter is configured such as to minimize a given cost function as for example the signal energy of the output signal. As mentioned, alternatively, a signal energy of a hypothetical resonator can be maximized.

It is known to use a cascade of adaptive narrow-band filters connected in series as shown in Fig. 6. Error signal $e(n)$ is fed to adaptive notch filter 1 having a center frequency f_1 . The narrow-band filter output signal $e_{f1}(n)$ is then fed into adaptive notch filter 2 having center frequency f_2 and so forth. As much as eight or ten or more notch filters may be employed for achieving a satisfactory feedback cancellation. Every filter of the cascade of adaptive narrow-band filters minimizes its own immediate output. This is a perfectly sufficient algorithm in the case of a static signal composition. After each notch stage one further sinusoid is removed from the signal. When the signal spectrum is fluctuating, however, this method proves to be inadequate. Now the first notch may jump from one sinusoid to another not taking into account that one of the later notch stages may already have adapted to this other sinusoid frequency. This leads to the generation of audible artefacts of the feedback cancellation system.

To avoid this problem the present invention provides according to one aspect a set of adaptive narrow-band filters connected in series configured such that a single shared cost function is minimized. An optimization (minimization or maximization) according to this cost function makes each narrow-band filter of the set of narrow-band filters aware of the effectiveness of all other notch filters. The cost function derived from the output signal of the last filter of the set of adaptive narrow-band filters is fed back to all filters for the optimization process as is shown schematically in Fig. 7.

With this method the effectiveness of the narrow-band filtering can be greatly improved, in particular for rapidly fluctuating signals.

One problem appearing with the filter arrangement shown in Fig. 7 is the increase of the amount of mathematical operations required for the gradient calculation with the increase of the number of notch filters. The calculation cost is roughly proportional to the square of the number of filters thus increasing heavily if a large number of narrow-band filters (and center frequencies) is utilized.

In order to solve this problem an arrangement as shown in Fig. 8 is proposed wherein a single shared cost function derived from the output of the last stage narrow-band filter is used as in the arrangement shown in Fig. 7, but the gradient calculations are performed independently for each filter stage. This shared error methodology works well as long as the center frequencies of the respective notch filters are sufficiently spaced from each other. For this reason it is preferable to use the filter arrangement of Fig. 8 in connection with more narrow band gradient algorithms as e.g. the true gradient algorithm, maxres gradient algorithm or true maxres algorithm explained before.

Another possibility to reduce the computational costs of the gradient calculation of a set of narrow-band filters using a shared cost function is illustrated in Fig. 9. The calculations performed by the second and further notch filter can to some extent be re-used for the gradient calculations of the other filters since the gradient calculation result is order invariant, i.e. the computation result of a cascade of linear filters is independent of the order of these filters. Furthermore, if the notch filters are

implemented in a direct form II realization a part of the gradient calculation can be extracted from the notch filters themselves. In the example of Fig. 8 the number of calculations for $N = 3$ adaptive notch filters is reduced from $1+2+3=6$ gradient calculations to three gradient calculations.

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If a larger number of narrow-band filters is required, however, a further reduction of computational costs may be necessary. For this purpose, according to one aspect of the present invention, a tree structure for the narrow-band filter arrangement is provided as schematically shown in Fig. 10. In this figure, notch filters are illustrated as squares, pseudo to true gradient conversion filters as circles and the octogons symbolize pseudo gradient calculation filters, which - again - are equivalent to the calculation of the notch filter's internal state $x(n)$ given in formula (3).

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In the embodiment shown in Fig. 9, however, the tree structure is after two stages replaced by the end structure of Fig. 9 proving somewhat more effective than the complete tree structure. In this realization the relationship between the number of calculations and the number of effective notch filters is given by:

$$M = k_1 N \log_2(N) + k_2 N \quad (17)$$

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wherein N is the number of filters and k_1 and k_2 are implementation dependent constants. For implementing a tree structure, naturally, the number of filters N should be an integer power of 2, that is $2^2, 2^3, 2^4, \dots$.

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A similar result can be obtained by implementing the tree structure to the maxres gradient algorithm (see above) which requires that each and every filter stage is realized as the very last of all filters.

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If the pseudo maxres or a true maxres gradient calculation algorithms are utilized, the implementation is very effective as these two gradient algorithms can be calculated from the output of the entire series of notch filters, that is the notch filtered signal can be used as the input of the gradient calculation filter. The consequence of this effective implementation is the central "dead zones" reflected in the sensitivity

plots of Fig. 12. This is also true for multiple notch filters, where the pseudo maxres gradient filters belonging to each adaptive notch filter are applied to the final output of the set of notch filters. If the pseudo to true gradient filter is extended to this filter result the true maxres gradient algorithm is obtained for multiple notches. The
5 computational cost of both these algorithms increases only linearly with the number of notch filters applied.

Claims

1. A hearing aid comprising:

5 an input transducer (2) for deriving an electrical input signal from an acoustic input,

a signal processor (3) for generating an electric output signal,

an output transducer (4) for transforming the electrical output signal into an acoustic output,

10 an adaptive estimation filter (5) for generating a feedback estimation signal,

at least one first adaptive narrow-band filter (8) for narrow-band-filtering an input signal of the signal processor (3),

15 at least one second adaptive narrow-band filter (9) for narrow-band-filtering a reference signal corresponding to an input signal of the adaptive estimation filter (5), and

an adaptation mechanism (6) for updating the filter coefficients of the adaptive estimation filter (5) based on the output signals of the first and second narrow-band filters.

20 2. The hearing aid of claim 1, wherein the at least one first adaptive narrow-band filter (8) and the at least one second adaptive narrow-band filter (9) are configured to minimize a cost function of its output signal.

25 3. The hearing aid of claim 2, wherein the cost function minimization is performed by a least mean square (LMS) type algorithm.

4. The hearing aid of claim 2 or 3, wherein the cost function is a signal energy.

30 5. The hearing aid of claim 1, wherein the at least one first adaptive narrow-band filter and the at least one second adaptive narrow-band filter are

configured to maximize the output of a hypothetical resonator of a given frequency with constrained pole radius.

6. The hearing aid of one of claims 1 to 5, wherein the at least one
5 second adaptive narrow-band filter (9) performs a calculation of a combined gradient, wherein a narrow-band gradient is calculated if the center frequency adaptation rate of the filter is below a predetermined threshold value and a broader band gradient is calculated if the center frequency adaptation rate of the filter is above the predetermined threshold value.

10 7. The hearing aid of one of claims 1 to 6, wherein the adaptive estimation filter (5) employs a least mean square (LMS) algorithm for feedback reduction.

15 8. The hearing aid of one of claims 1 to 7, wherein the adaptation mechanism (6) performs a cross correlation processing of the output $e_f(n)$ of the at least one first adaptive narrow-band filter (8) with the output $u_f(n)$ of the at least one second adaptive narrow-band filter (9).

20 9. The hearing aid of one of claims 1 to 8, wherein the at least one first and second adaptive narrow-band filters are notch filters having an adaptive center frequency $c(n)$ with frequency width r .

25 10. The hearing aid of one of claims 1 to 9, comprising a plurality of first and second adaptive narrow-band filters having different adaptive center frequencies $c(n)$.

30 11. A method of adaptively reducing the acoustic feedback of a hearing aid comprising an input transducer for deriving an electrical input signal from an acoustic input, a signal processor for generating an electrical output signal and an output transducer (4) for transforming the electrical output signal into an acoustic output, the method comprising the steps of:

generating a feedback estimation signal,
deriving an error signal by subtracting the feedback estimation signal
from the electrical input signal,
narrow-band-filtering the error signal and a reference signal
5 corresponding to a feedback estimation input signal, and
adapting feedback estimation filter coefficients based on the narrow-
band-filtered signals.

12. The method of claim 11, wherein a cost function of the narrow-band
10 filtered error signal or of the narrow-band filtered reference signal are minimized.

13. The method of claim 12, wherein the cost function minimization is
performed by a least mean square (LMS) type algorithm.

15 14. The method of claim 12 or 13, wherein the cost function is the signal
energy.

15. The method of claim 11, wherein the narrow-band filtering is
performed maximizing the output of a hypothetical resonator of a given frequency
with constrained pole radius.

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16. The method of one of claims 11 to 15, wherein the at least one of the
first or the second adaptive narrow-band filters (8, 9) performs a calculation of a
combined gradient, wherein a narrow-band gradient is calculated if the center
frequency adaptation rate of the filter is below a predetermined threshold value and
25 a broader band gradient is calculated if the center frequency adaptation rate of the
filter is above the predetermined threshold value.

17. The method of one of claims 11 to 16, wherein the feedback
estimation signal is generated using a least mean square (LMS) algorithm.

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18. The method of one of claims 11 to 17, wherein a cross correlation processing of the narrow-band-filtered error signal with the narrow-band-filtered reference signal is performed.

5 19. The method of one of claims 11 to 18, wherein the narrow-band filtering is performed by notch filters having an adaptive center frequency $c(n)$ with frequency width r .

20. The method of one of claims 11 to 19, wherein the narrow-band
10 filtering is performed in a plurality of subsequent stages connected in series and having different adaptive center frequencies.

21. A computer program comprising program code for performing the method according to one of claims 11 to 20.

15

22. An electronic circuit for a hearing aid comprising:

 a signal processor (3) for processing an electrical input signal derived from an acoustic input and generating an electric output signal,

20 an adaptive estimation filter (5) for generating a feedback estimation signal,

 at least one first adaptive narrow-band filter (8) for narrow-band-filtering an input signal of the signal processor (3),

25 at least one second adaptive narrow-band filter (9) for narrow-band-filtering a reference signal corresponding to an input signal of the adaptive estimation filter (5), and

 an adaptation mechanism (6) for updating the filter coefficients of the adaptive estimation filter (5) based on the output signals of the first and second narrow-band filters.

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23. A hearing aid comprising:

 an input transducer (2) for deriving an electrical input signal from an acoustic input,

a signal processor (3) for generating an electric output signal,
an output transducer (4) for transforming the electrical output signal into
an acoustic output,
an adaptive estimation filter (5) for generating a feedback estimation
5 signal,
at least one first adaptive narrow-band filter (8) for narrow-band-filtering
an input signal of the signal processor (3),
at least one second adaptive narrow-band filter (9) for narrow-band-
filtering a reference signal corresponding to an input signal of the adaptive
10 estimation filter (5),
an adaptation mechanism (6) for updating the filter coefficients of the
adaptive estimation filter (5) based on the output signals of the first and second
narrow-band filters,
wherein the filters of the first and second set of adaptive narrow-band
15 filters are each configured to minimize a single shared cost function.

24. The hearing aid of claim 23, wherein the second set of adaptive
narrow-band filters (9) is configured to derive its output signals from gradients of the
output signals of the respective filters of the first set of adaptive narrow-band filters
20 (8).

25. The hearing aid of claim 24, wherein the first set of adaptive narrow-
band filters is at least partially arranged in a tree structure.

26. The hearing aid of claim 24, wherein the gradient calculations of the
plurality of filters of the first or the second set of adaptive narrow-band filters are
performed independently of each other.

27. The hearing aid of one of claims 23 to 26, wherein the filters of the
30 first or the second set of adaptive narrow-band filters are configured to maximize an
output of a hypothetical resonator with constraint pole radius.

28. The hearing aid of one of claims 24 to 27, wherein the first or the second set of adaptive narrow-band filters (9) performs a calculation of a combined gradient, wherein a narrow-band gradient is calculated if the center frequency adaptation rate is below a predetermined threshold value and a broader band gradient is calculated if the center frequency adaptation rate of the adaptive narrow-band filter is above the predetermined threshold value.

29. The hearing aid of one of claims 23 to 28, wherein the adaptive estimation filter (5) employs a least mean square (LMS) algorithm for feedback reduction.

30. The hearing aid of one of claims 23 to 29, wherein the adaptation mechanism (6) performs a cross correlation processing of the outputs $e_f(n)$ of a first set of adaptive narrow-band filters (8) and the outputs $u_f(n)$ of the second set of adaptive narrow-band filters (9).

31. The hearing aid of one of claims 23 to 30, wherein the first and second sets of adaptive narrow-band filters comprise notch filters having an adaptive center frequency $c(n)$ with frequency width r .

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32. A method of adaptively reducing an acoustic feedback of a hearing aid comprising an input transducer for deriving an electrical input signal from an acoustic input, a signal processor for generating an electrical output signal and an output transducer for transforming the electrical output signal into an acoustic output, the method comprising the steps of:

generating a feedback estimation signal,
deriving an error signal by subtracting the feedback estimation signal from the electrical input signal,
narrow-band-filtering the error signal and a reference signal
corresponding to a feedback estimation input signal in a plurality of filter stages having different adaptive center frequencies,

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adapting feedback estimation filter coefficients based on the narrow-band-filtered error and reference signals,

wherein the narrow-band filtering using a plurality of different adaptive center frequencies is performed minimizing a single shared cost function.

5

33. The method of claim 32, wherein the narrow-band filtered reference signal is derived from a gradient of the narrow-band filtered error signal.

34. The method of claim 33, wherein the gradient calculation is performed employing a least a partial tree structure of narrow-band filters.

10

35. The method of claim 33 or 34, wherein the gradient calculations of different adaptive narrow-band filter stages are performed independently of each other.

15

36. The method of one of claims 33 to 35, wherein the narrow-band filtering is performed maximizing an output of a hypothetical resonator with constrained pole radius.

37. The method of claim 32, wherein a combined gradient calculation is performed, wherein a narrow-band gradient is calculated if the center frequency adaptation rate is below a predetermined threshold value and a broader band gradient is calculated if the center frequency adaptation rate of the adaptive narrow-band filter is above the predetermined threshold value.

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38. The method of one of claims 32 to 37, wherein the feedback estimation signal is generated using a least mean square (LMS) algorithm.

39. The method of one of claims 32 to 38, wherein the feedback estimation filter coefficients are adapted utilizing a cross correlation processing of the narrow-band filtered error signal with the narrow-band filtered reference signal.

25

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40. The method of one of claims 32 to 39, wherein the narrow-band filtering is performed by notch filters having an adaptive center frequency $c(n)$ with frequency width r .

5 41. A computer program comprising program code for performing the method of one of claims 32 to 40.

42. An electronic circuit for a hearing aid comprising:

10 a signal processor (3) for processing an electrical input signal derived from an acoustic input and generating an electrical output signal,
 an adaptive estimation filter (5) for generating a feedback estimation signal,

 at least one first adaptive narrow-band filter (8) for narrow-band-filtering an input signal of the signal processor (3),

15 at least one second adaptive narrow-band filter (9) for narrow-band-filtering a reference signal corresponding to an input signal of the adaptive estimation filter (5),

 an adaptation mechanism (6) for updating the filter coefficients of the adaptive estimation filter (5) based on the output signals of the first and second
20 narrow-band filters,

 wherein the first and second set of adaptive band filters are each configured to minimize a single shared cost function.