Low-delay hearing aid based on cochlear model with nonuniform subband acoustic feedback cancellation

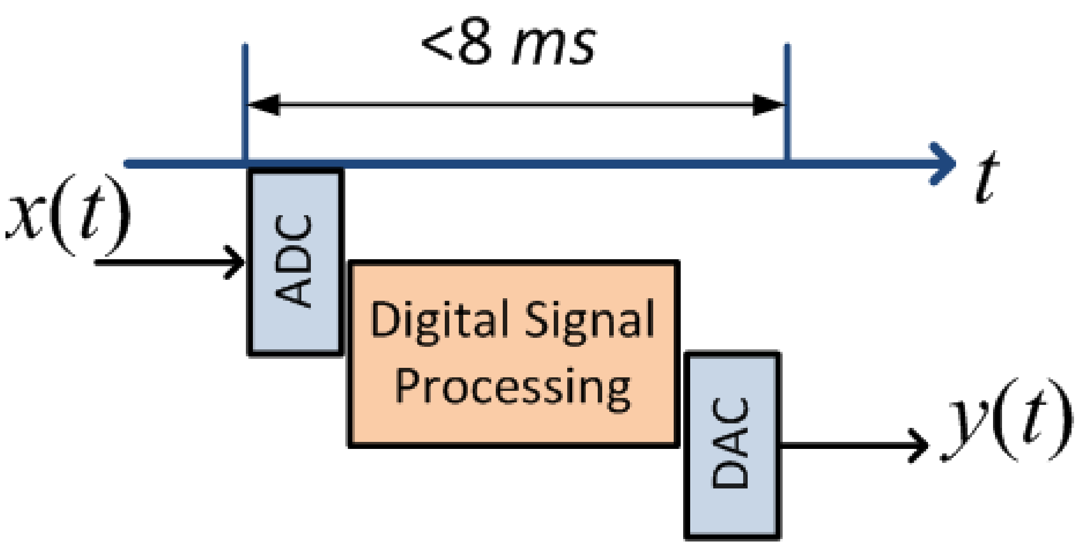
Maxim Vashkevich, Elias Azarov and Alexander Petrovsky

Computer Engineering Department,
Belarusian State University of Informatics and Radioelectronics,

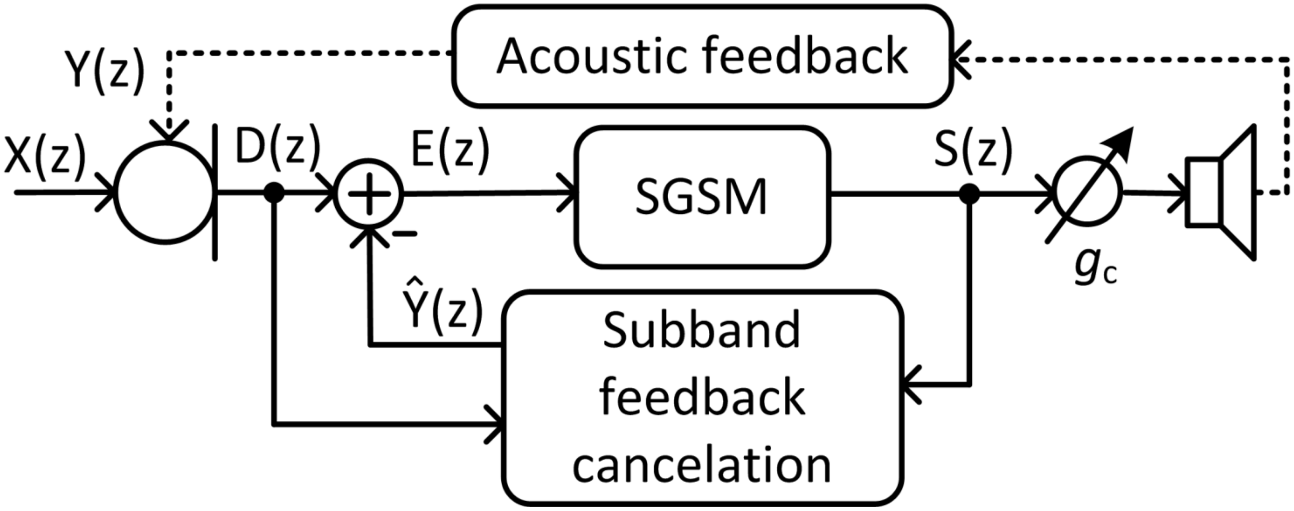
Hearing aid time constraint

Total processing time T_{total} should not exceed 6-8 ms:

$$T_{total} < 8\ ms$$



Proposed low-delay HA system



SGSM – spectral gain shaping method:

- Hearing loss compensation
- Noise reduction
- Dynamic range compression

Low-delay cochlear filter bank (1)

- Cochlear filter bank is obtained from 2D nonlinear cochlear model for discrete space and time:

$$y_k(n) + b_{1,k} y_k(n-1) + b_{2,k} y_k(n-2) = A_k \alpha_{0,k}[u_s(n) - u_s(n-2)],$$

where $b_{1,k}, b_{2,k}, a_{0,k}$ *and* A_k parameters that are determined on the basis of physical characteristics of the basilar membrane [5].

[5] W. Wan, A. Petrovsky, and C. Fan, "A two-dimensional nonlinear cochlear model for speech processing: response to pure tone," in Proc. of 6th International Fase – Congress, Zurich, Switzerland, 1992, pp. 233–236.

Low-delay cochlear filter bank (2)

- Using cochlear model corresponding filter bank can be obtained:

$$H_k(z) = A_k \frac{a_{0k}(1 - z^2)}{1 + b_{1k}z^{-1} + b_{2k}z^{-2}}$$

- Center frequency and bandwidth of *k*-th cochlear filter are expressed as follows

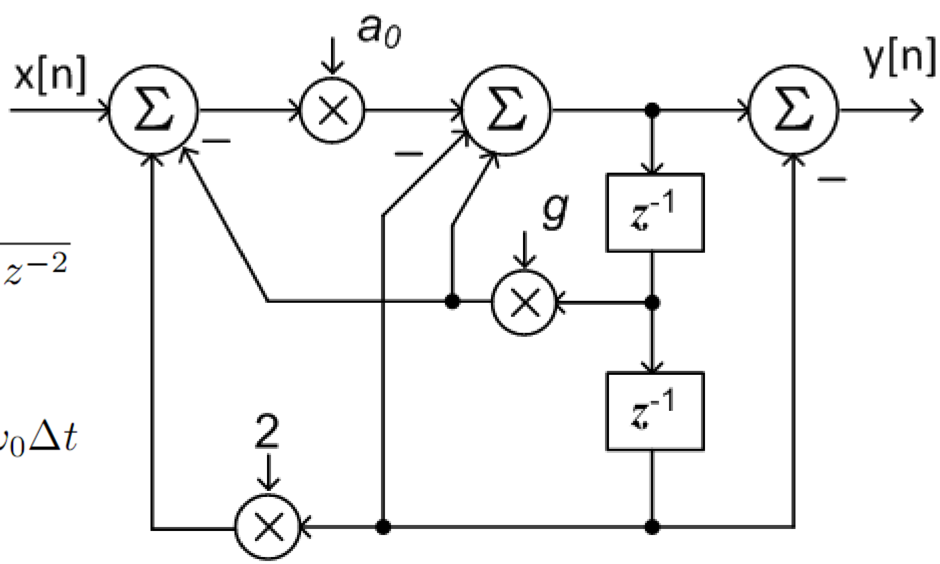
$$\cos\omega_{0k} = -b_{1k}/(1 + b_{2k}), \quad \Delta\omega_{0k} = 2(1 - b_{2k})/(1 + b_{2k})$$

Low-delay cochlear filter bank (3)

Cochlear filters can be efficiently implemented using IIR filter structure with tunable bandwidth and center frequency presented in [6]:

$$H(z) = a_0 \frac{1 - z^{-2}}{1 + (a_0 - 1)gz^{-1} + (1 - 2a_0)z^{-2}}$$

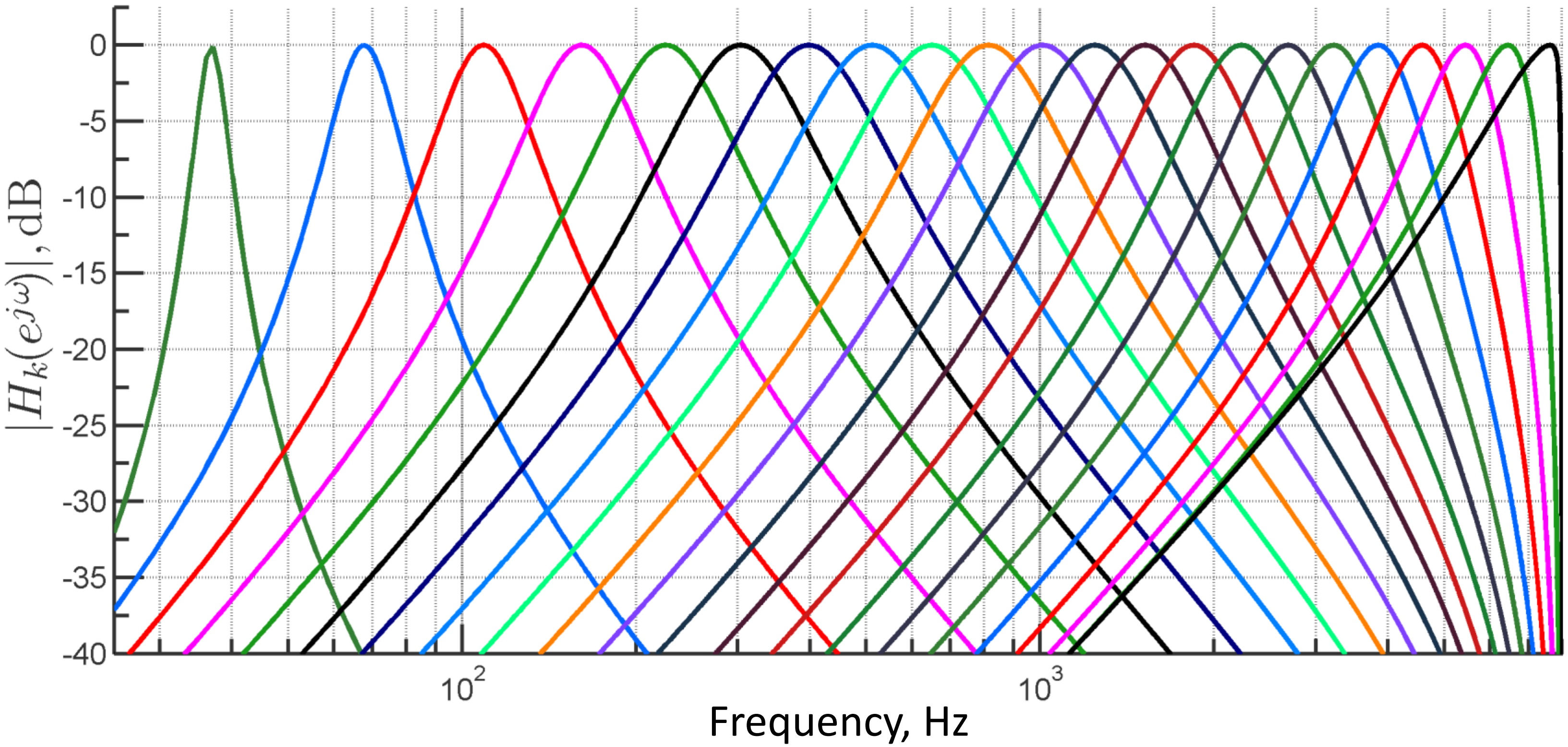
$$a_0 = (\Delta\omega\Delta t)/(2 + \Delta\omega\Delta t), \quad g = 2\cos\omega_0\Delta t$$



[6] A. Petrovsky, "The synthesis of high order digital bandpass filters with tunable center frequency and bandwidth," in Proc. of EUSIPCO'96, Trieste, Italy, 1996, pp. 1527–1530.

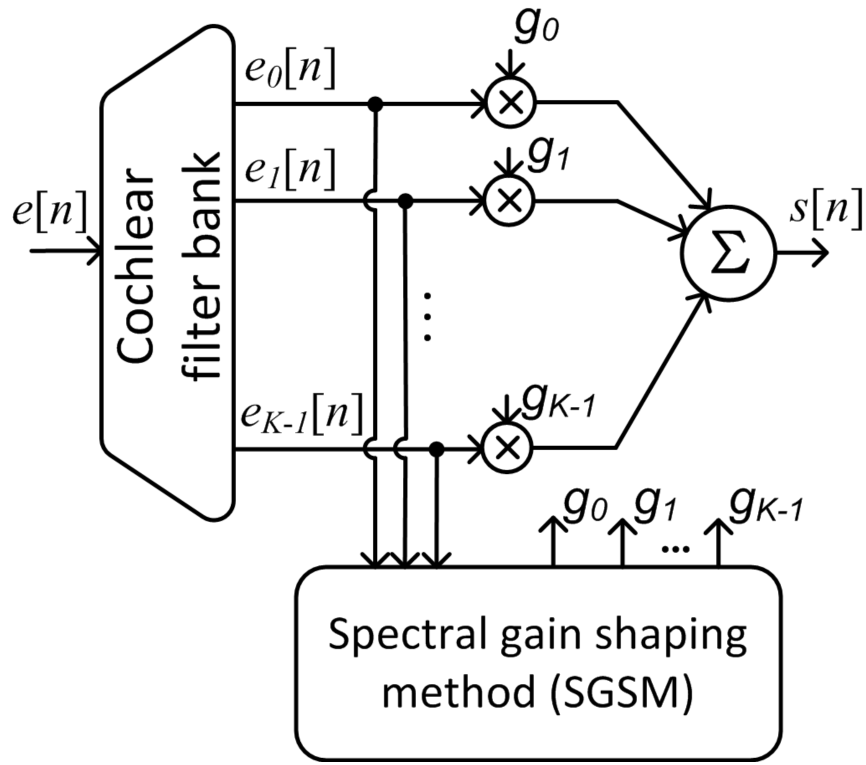
Low-delay cochlear filter bank (4)

- Frequency response of 22 channel cochlear filter bank



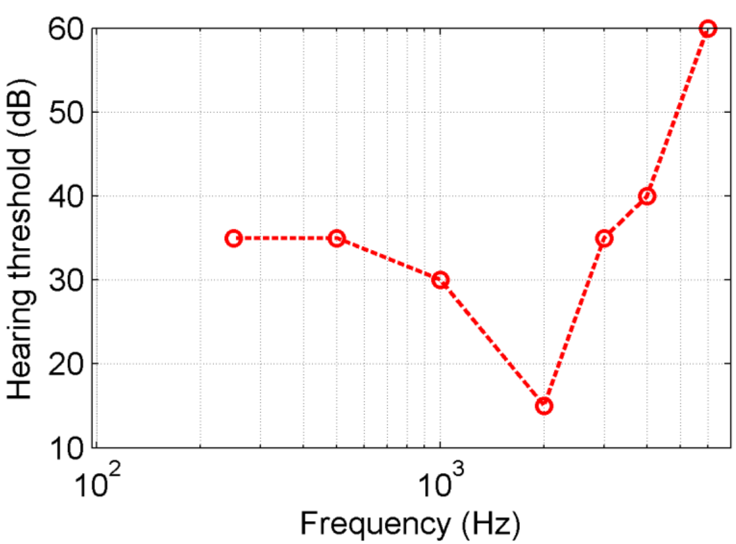
Hearing aid forward path structure

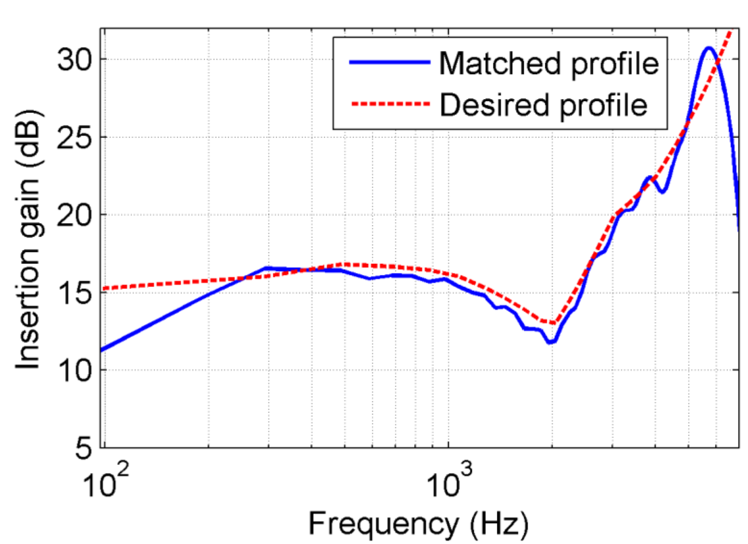
Spectral gain shaping method (SGSM) combines dynamic range compression (DRC), noise reduction (NR) and hearing loss compensation.



Hearing loss compensation

The hearing loss threshold is frequency depended and determined at specified frequencies (250, 500, 1000, 2000, 3000, 4000 and 6000 Hz) using a pure tone audiogram. The shape of the correspondent insertion gain depends on prescription method and does not necessarily follow the shape of the hearing loss thresholds. The figure shows how a given insertion gain matches using proposed SGSM (the insertion gain is obtained with the NAL-RP prescription).





Noise reduction

Noise reduction is implemented using the psychoacoustically motivated spectral weighting rule [10-11].

[10] A. Petrovsky, M. Parfieniuk, and A. Borowicz, “Warped DFT based perceptual noise reduction system,” in AES , Convention Paper 6035, presented at the 116th Convention, Berlin, Germany. – 16p.

[11] A. Borowicz, M.Parfieniuk, A.Petrovsky An application of the warped discrete Fourier transform in the perceptual speech enhancement. – Speech Communication, vol. 48, ELSEVIER, 2006. – pp.1024-1036.

Sample processing iterations (for $n = 1, 2 \ldots$)

$$\mathbf{R}_e(n) = [R_e(0,n), R_e(1,n) \ldots R_e(K-1,n)]^T$$

$$\mathbf{E}(n) = [e_0^2(n), e_1^2(n) \ldots e_{K-1}^2(n)]^T$$

$$\mathbf{R}_e(n) = \gamma \mathbf{R}_e(n-1) + (1-\gamma)\mathbf{E}(n)$$

Block processing iterations ($n = R, 2R \ldots$)

$$\mathbf{R}_n(n) = \mathrm{MCRA}(\mathbf{R}_e(n))$$

$$\mathbf{R}_T(n) = \mathbf{R}_n(n) \odot (\mathbf{H}^{\mathrm{opt}}(n) - \zeta)^2$$

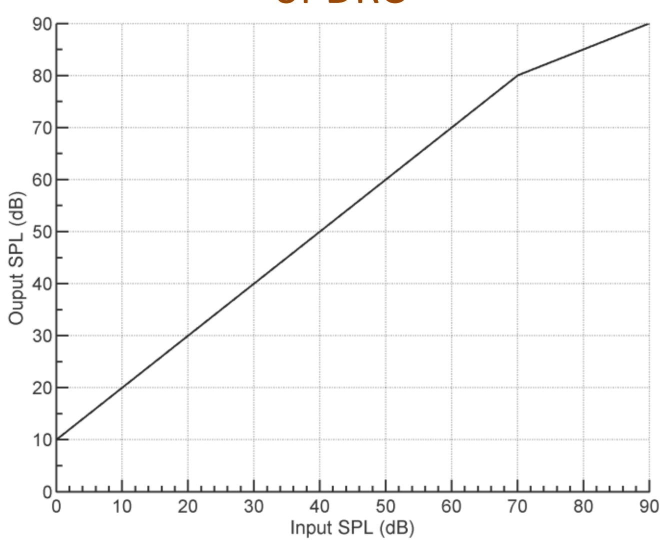
$$\mathbf{H}^{\mathrm{JND}}(n) = \min(\sqrt{\mathbf{R}_T(n)/\mathbf{R}_n(n)} + \zeta, 1)$$

- $\mathbf{H}^{\mathrm{opt}}(n)$ is equal to the Wiener-filter solution
- $\odot$ denotes element-by-element multiplication
- JND stands for Just Notable Distortion
- R is block size in samples

Dynamic range compression (DRC)

The basic idea of DRC is to automatically control the gain in each subband based on the current signal level.

Typical input/output function of DRC



Sample processing iterations (for $n = 1, 2 \dots$)

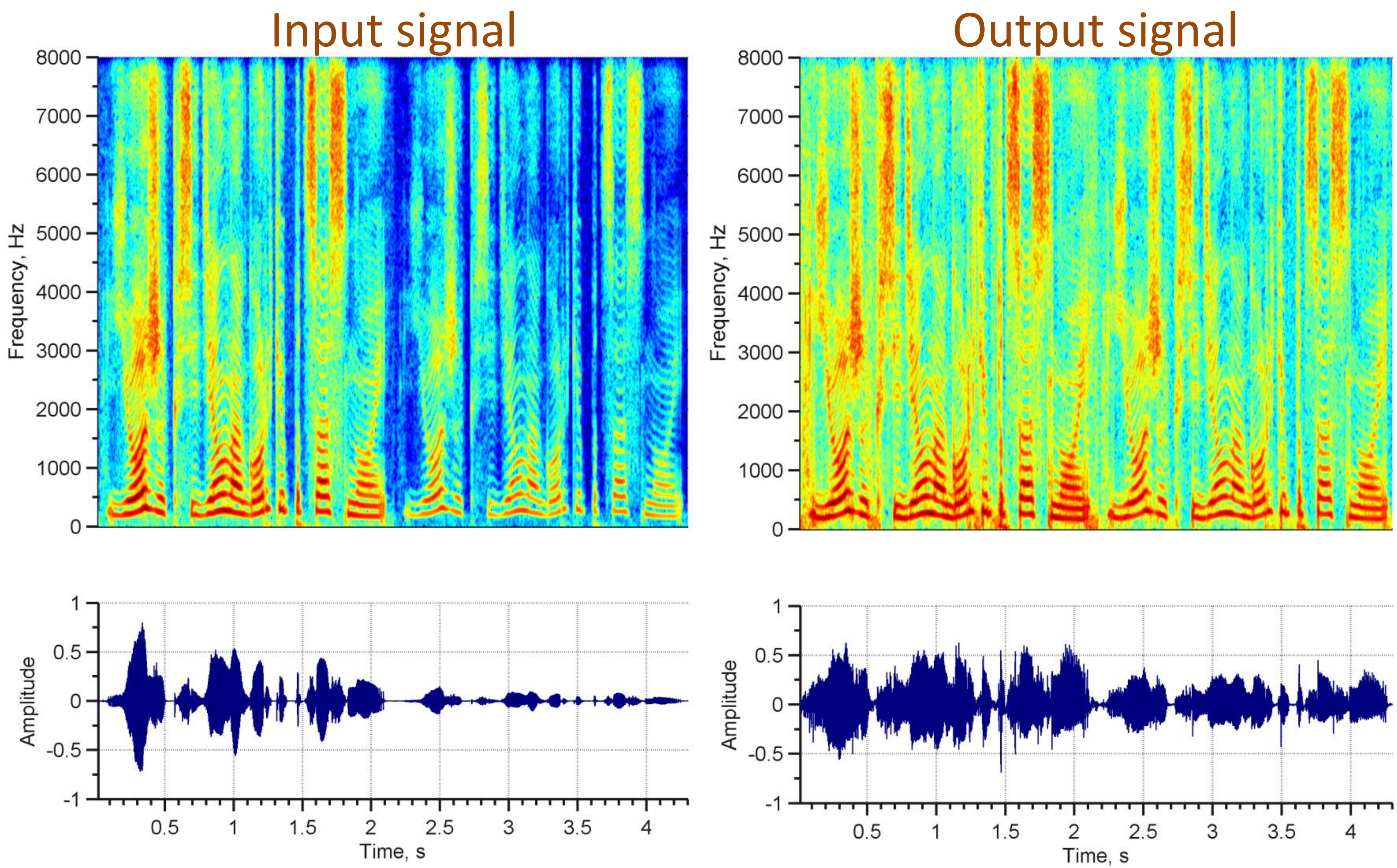
for $k = 0, \dots, K-1$ **do**
    if $P(k, n-1) < e_k^2(n)$ **then**
        $P(k,n) = \alpha P(k, n-1) + (1-\alpha) e_k^2(n);$
    else
        $P(k,n) = \beta P(k, n-1) + (1-\beta) e_k^2(n);$
    end if
end for

Block processing iterations ($n = R, 2R \dots$)

for $k = 0, \dots K-1$ **do**
    $P_{out}(k) = \mathrm{IOFunc}(10 \log_{10}(P(k,n) H^{\mathrm{JND}}(k,n) D(k)));$
    $G(k) = P_{out}(k) - P(k,n);$
    $g_k = 10^{G(k)/10};$
end for

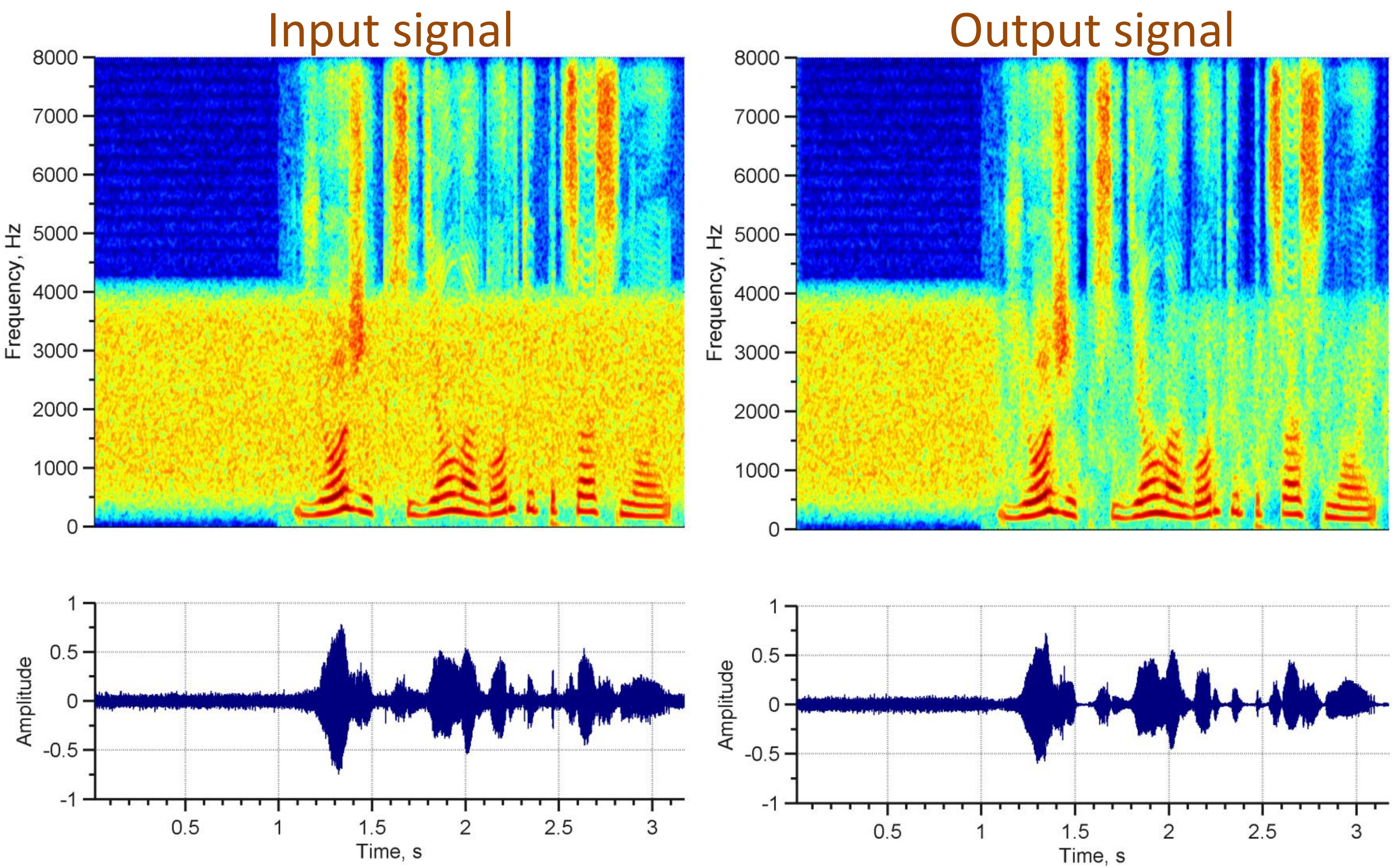
- IOFunc – input/output function of DRC
- $D(k)$ – insertion gains of k-th channel
- α, β – smoothing constants of attack and release times

DRC example



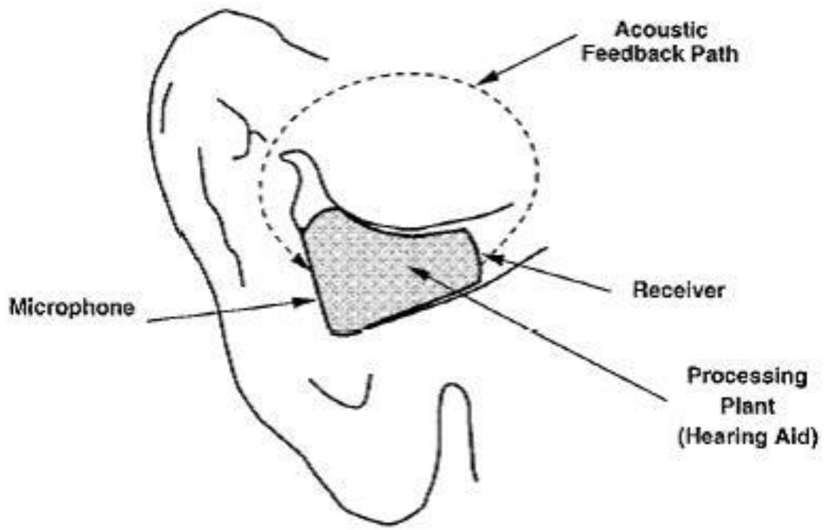
Input signal
Output signal
Frequency, Hz
8000
7000
6000
5000
4000
3000
2000
1000
0
Amplitude
1
0.5
0
-0.5
-1
0.5
1
1.5
2
2.5
3
3.5
4
Time, s

Noise reduction example



Acoustic feedback

Acoustic feedback is an important factor that degrades the overall performance of hearing aids

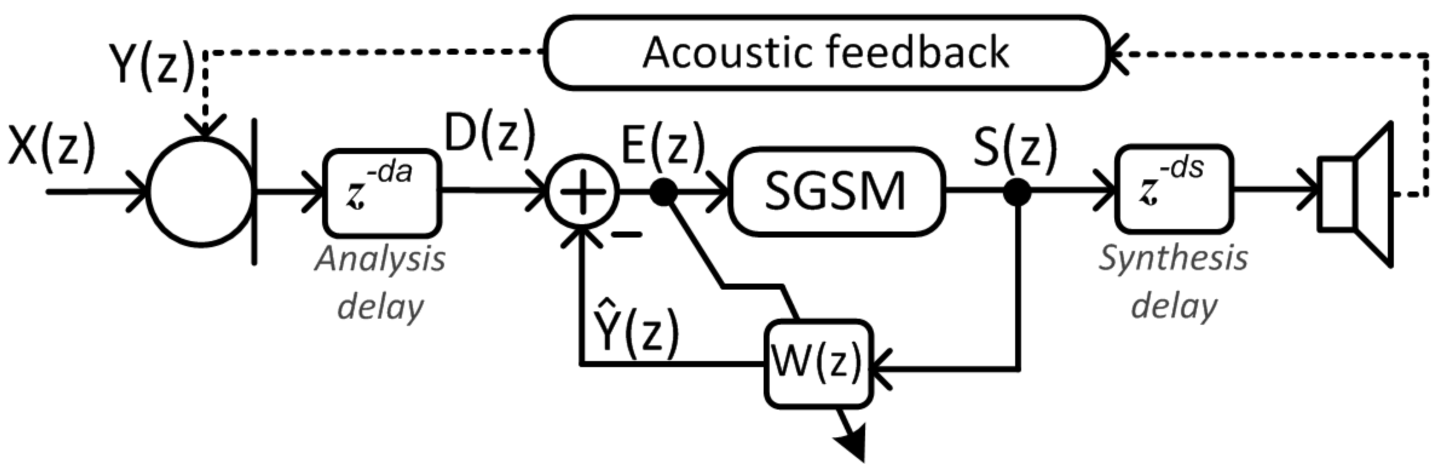


Acoustic feedback

- ✓Limits maximum hearing aid gain
- ✓Reduces speech intelligibility

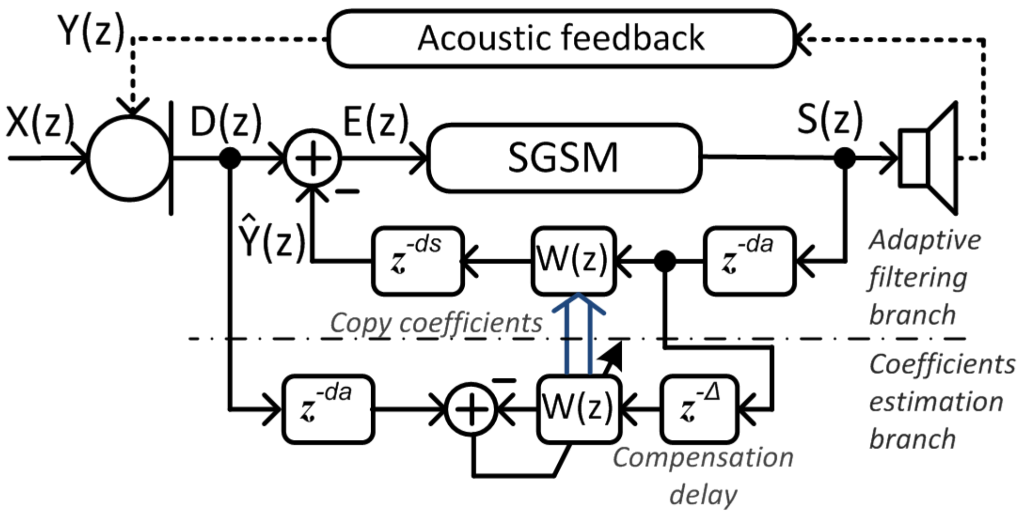
Acoustic feedback: problem formulation

It is known that subband decomposition can considerably improve acoustic feedback cancelation. However straightforward incorporation of a filter bank in the feedback cancellation branch adds an ***additional analysis/synthesis delay*** to the forward path.



Acoustic feedback: solution

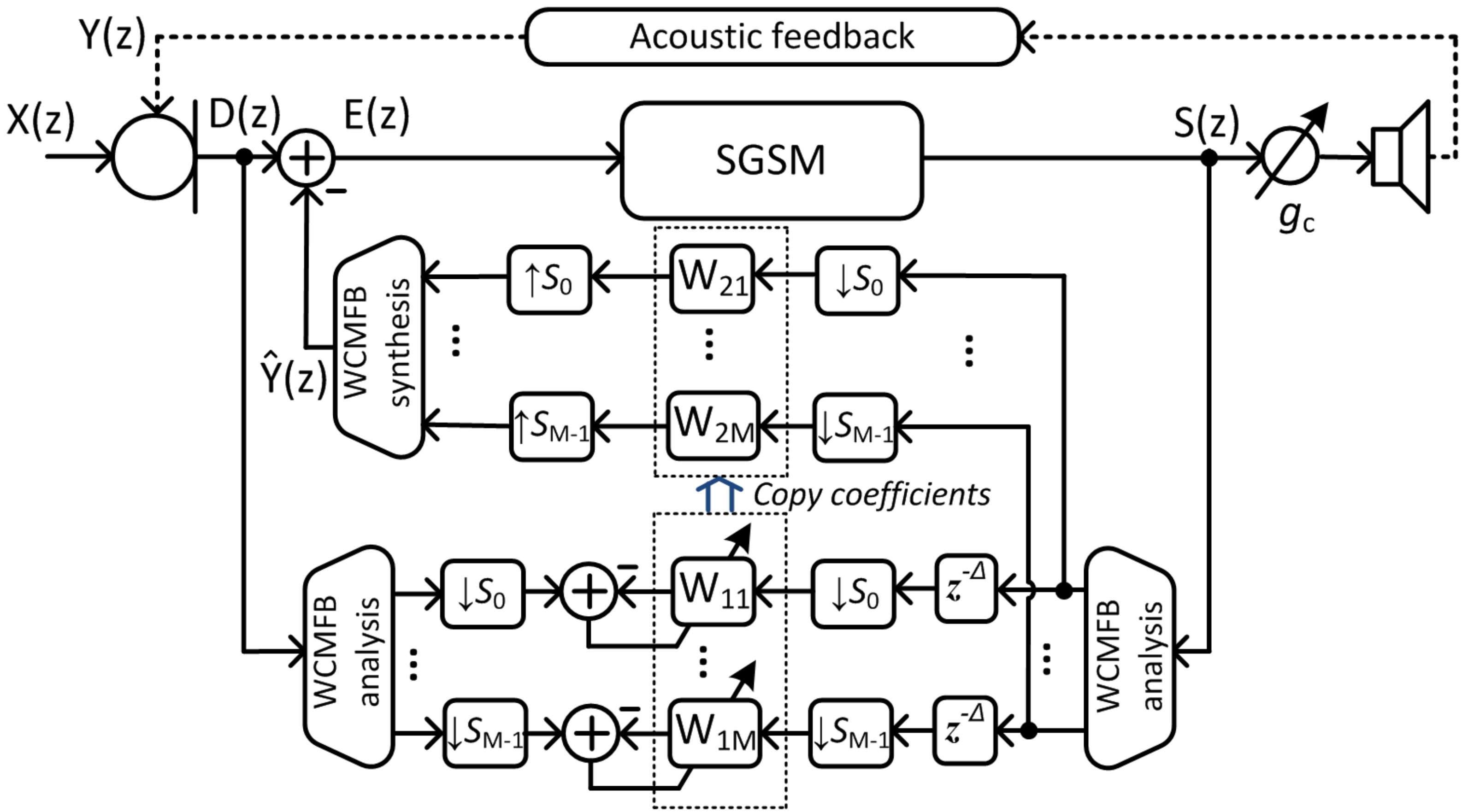
- In order to avoid forward path extension we propose an alternative scheme



- The AFC system is divided into two parts: the adaptive filtering branch and the coefficients estimation branch. The synthesis delay ***ds*** is compensated by prediction of the feedback signal.

Subband acoustic feedback cancellation

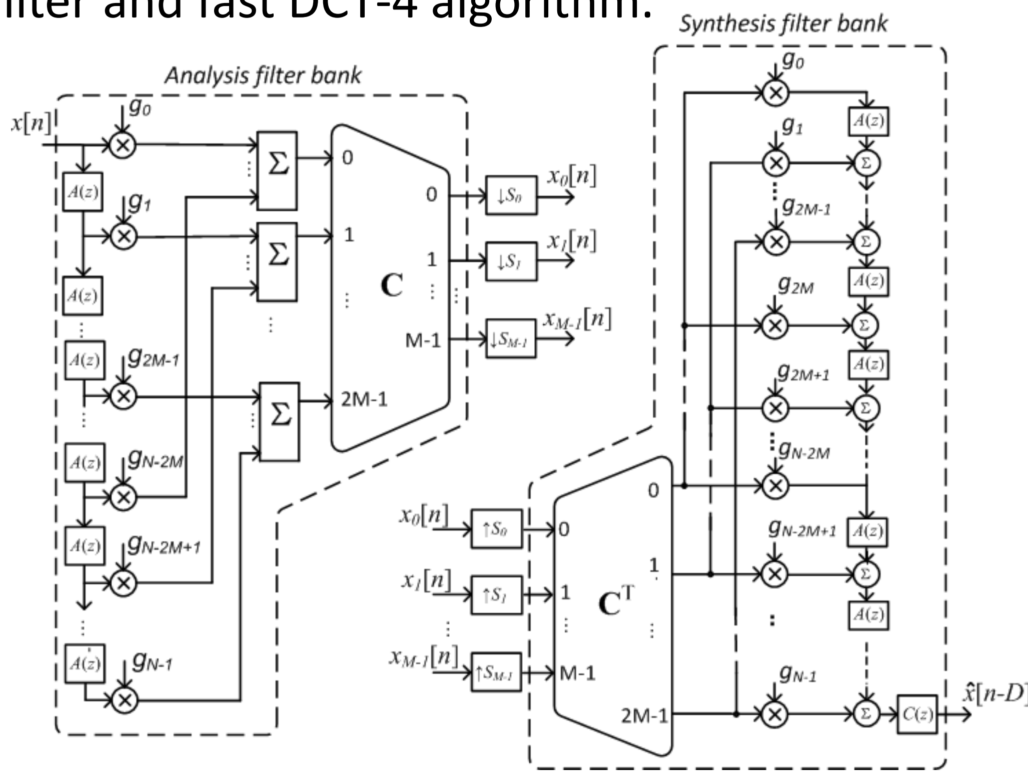
Proposed subband acoustic feedback cancellation system



WCMFB - warped cosine-modulated filter bank

Subband decomposition (1)

Subband decomposition in AFC system is carried out using oversampled warped cosine-modulated filter bank (WCMFB). The filter bank is obtained by applying all-pass transform to uniform CMFB [12]. WCMFB has an efficient implementation based on polyphase representation of the prototype filter and fast DCT-4 algorithm.

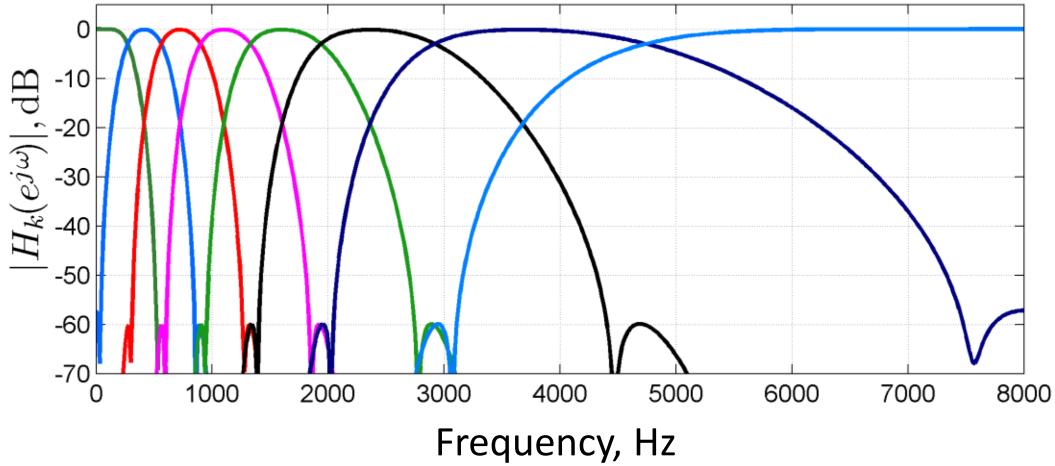


[12] M. Parfieniuk and A. Petrovsky "Near-perfect reconstruction oversampled nonuniform cosine-modulated filter banks based on frequency warping and subband merging" Inter. Journal of Electronics and Telecommunications, vol.58, no.2, pp. 177-192, 2012.

Subband decomposition (2)

For subband AFC system 8-channel oversampled WCMFB was designed using method [7].

- Subsampling factors $S_k = \{19, 9, 6, 4, 2, 1, 1, 1\}$.
- Oversampling ratio $\mathcal{O} = \sum_{k=0}^{7} \frac{1}{S_k} \approx 4.08$

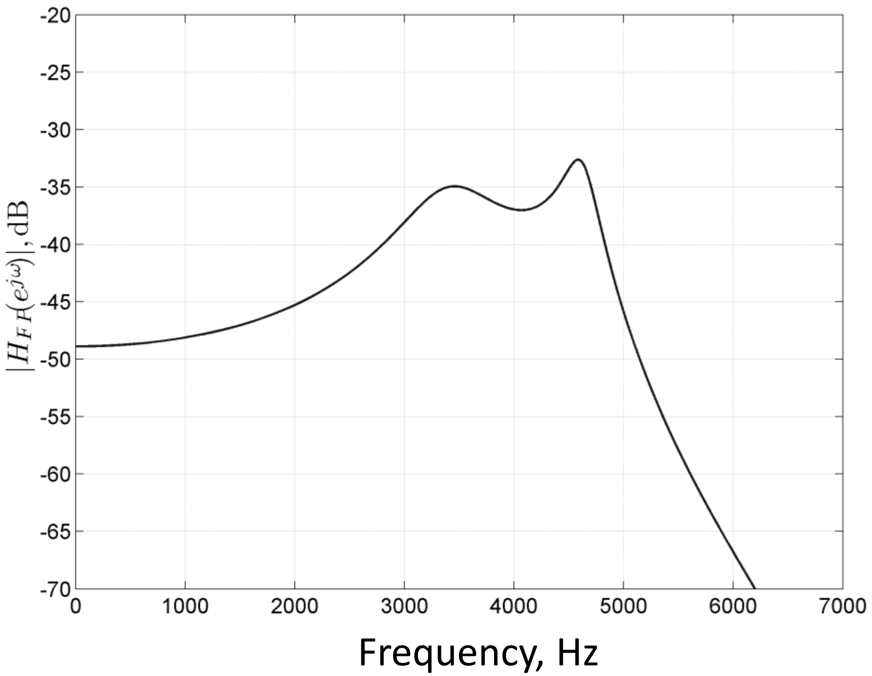


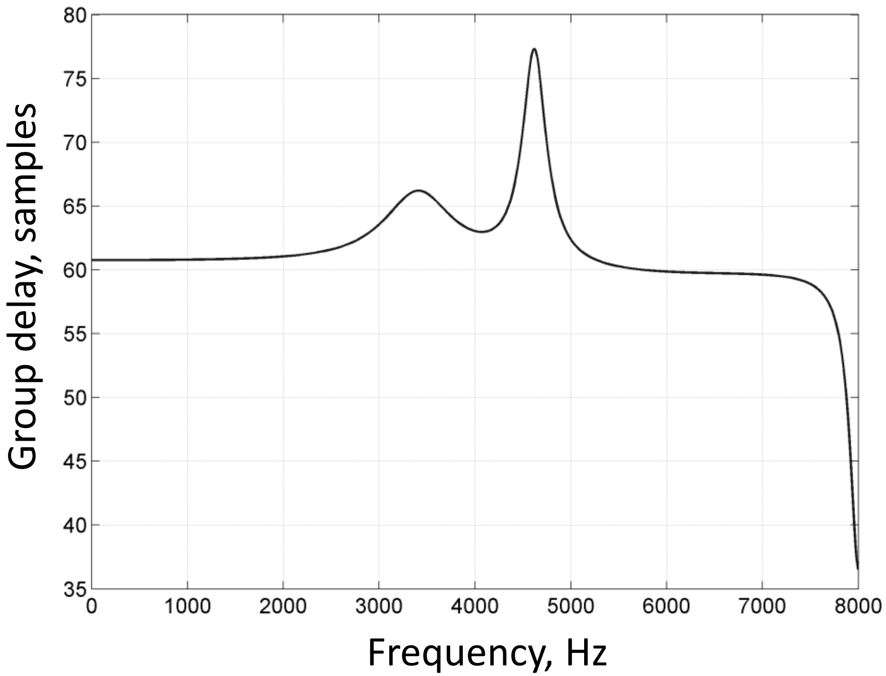
[7] M. Vashkevich, A. Petrovsky, and W. Wan, “Practical design of multi-channel oversampled warped cosine-modulated filter banks,” in Proc. of CCWMC’11, Shanghai, China, Nov. 2011, pp. 44–49.

Acoustic feedback path model

Acoustic feedback path is modeled as a transfer function in the Laplace domain using following expression:

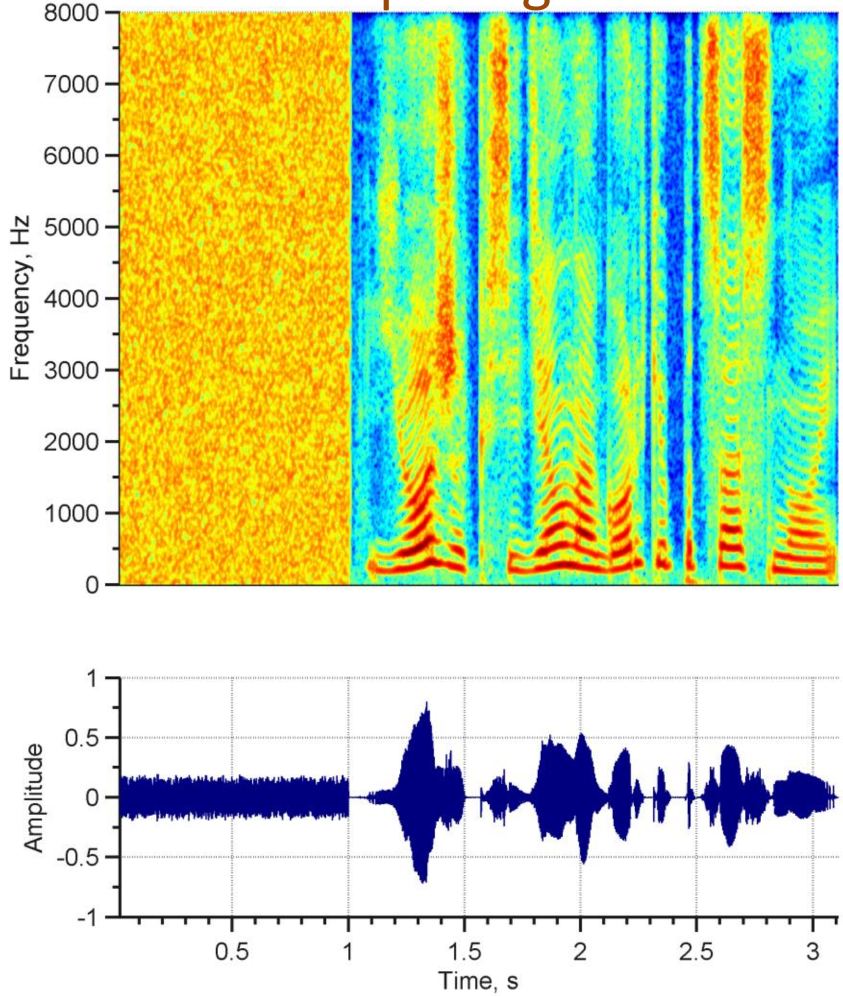
$$H_{FP}(s) = Ks^{-sT}\frac{(s-z_1)(s-z_2)(s-z_2^*)}{(s-p_1)(s-p_1^*)(s-p_2)(s-p_2^*)}$$



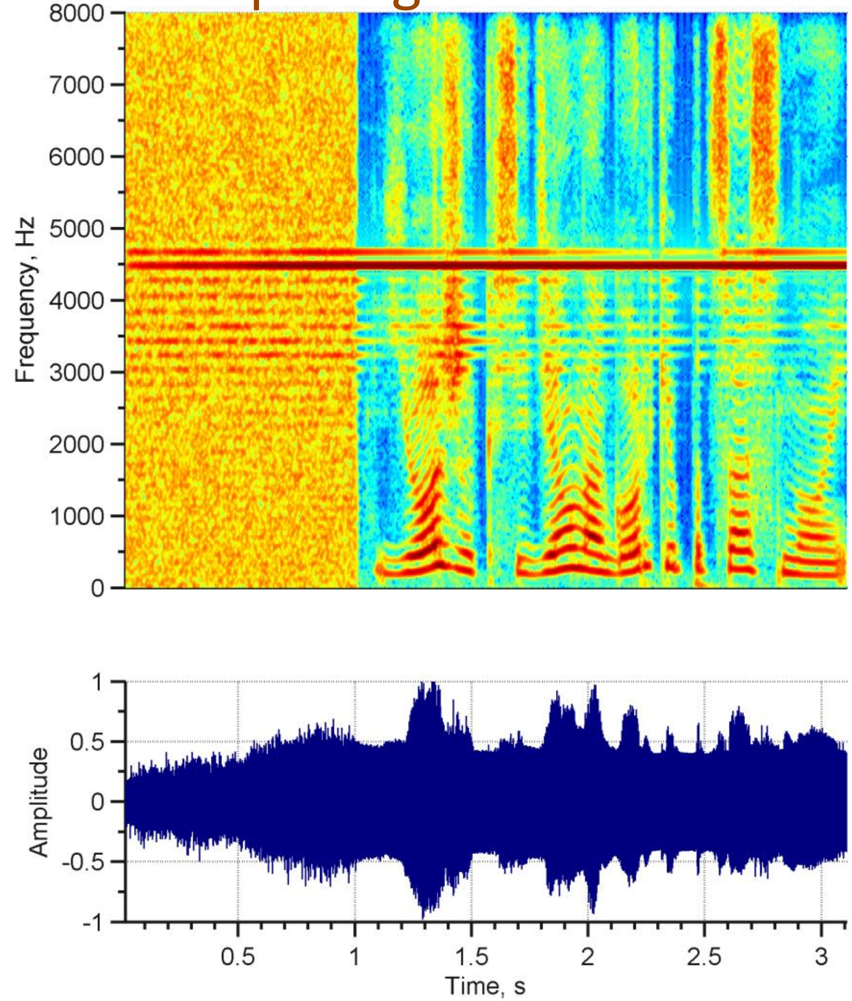


Acoustic feedback cancellation example (1)

Input signal

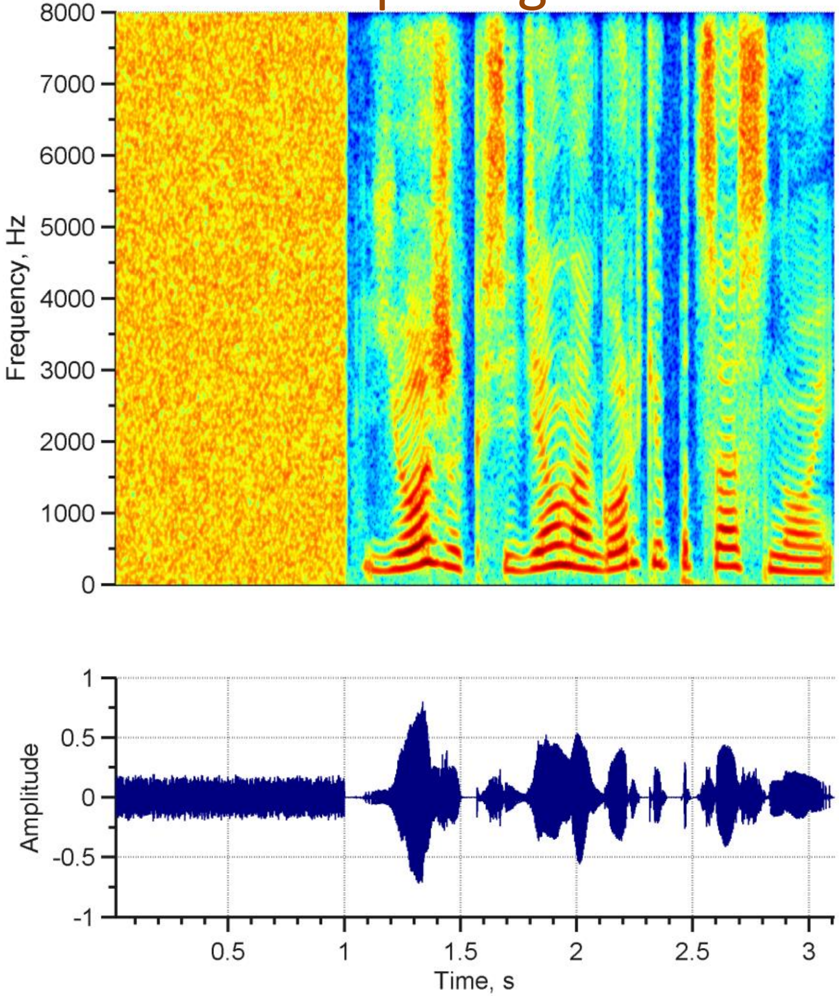


Output signal without AFC

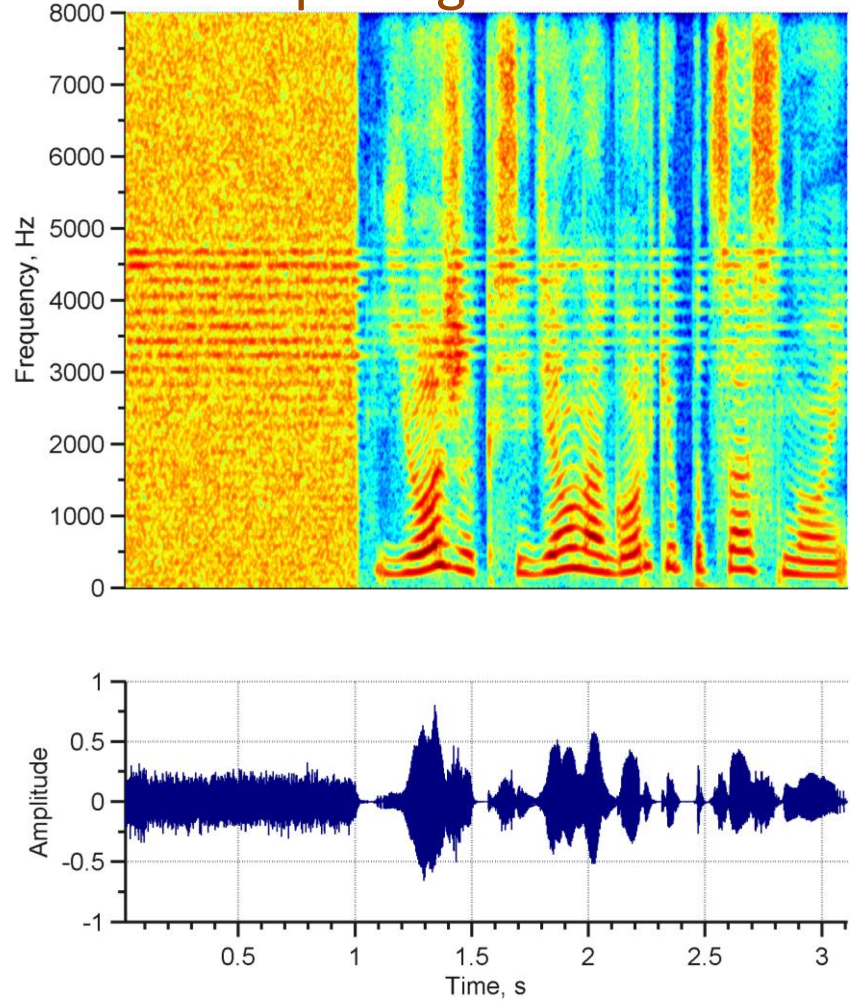


Acoustic feedback cancellation example (2)

Input signal

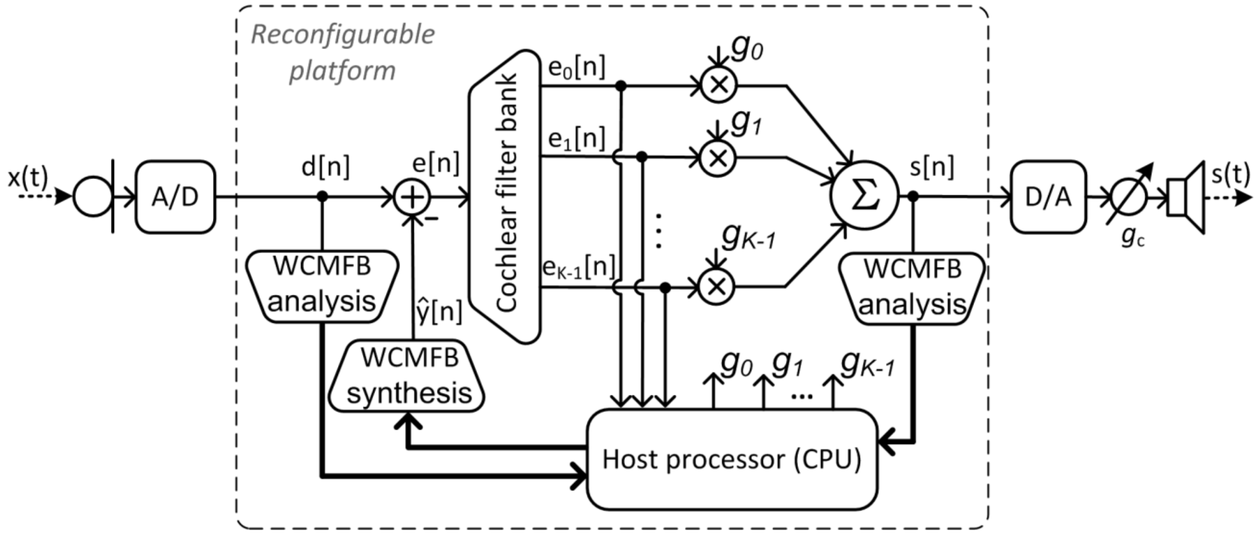


Output signal with AFC



Hearing aid prototype (1)

The prototype of hardware part of the hearing aid



Computational complexity of HA algorithms

Algorithm	Complexity
WCMFB	$O(\frac{M}{2}\log_2(M) + M + 2N)$
Cochlear FB	$O(2rK)$
Subband AFC	$O(\mathcal{O} \cdot (L + 2L))$
NR + DRC	$O(3K + \frac{11}{R}K)$

M – number of WCMFB channels
N – order of WCMFB prototype filter
K – number of cochlear filters
L – adaptive filter order

Hearing aid prototype (2)

- Hardware prototype of hearing aid is tested using Xilinx ML401 Evaluation Platform (XC4VLX2).
- MIPS 32 IP-core used as a host CPU.
- Cochlear filter bank is implemented using internal MAC units.
- To increase efficiency of polyphase structure of WCMFB distributed arithmetic is employed.
- Fast DCT-4 algorithm is carried out with multiplierless algebraic integer technique.

Conclusion

- A low-delay hearing aid system based on cochlear filter bank has been presented.
- Subband acoustic feedback cancellation system with prediction of feedback signal is proposed. The system does not increase the forward path delay.
- The spectral gain shaping method that includes noise reduction, hearing loss compensation and dynamic range compression is developed.

[2] J. M. Kates and K. H. Arehart, "Multichannel dynamic range compression using digital frequency warping," EURASIP J. Adv. Sig. Proc, vol. 2005, no. 18, pp. 3003–3014, 2005.

[3] A. Pandey and V. J. Mathews, "Low-delay signal processing for digital hearing aids," IEEE Transactions on Audio, Speech, and Language processing, vol. 19, no. 4, pp. 699–710, 2011.

Thank you for listening !