Multirate and Filterbank Approaches in Digital Hearing Aid Design: A Review

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Abstract. Hearing is considered as the most important senses of humans because it connects us to the world always. And the communication between humans are mostly by speaking and listening. So hearing loss can have a significant affect on our life, from our work to our connections and enthusiastic prosperity. A hearing aid is an electro-acoustic gadget, which is intended to increase sound, with the point of making speech more comprehensible. The fundamental task of the hearing aid is to selectively amplify sound signals to such an extent that the handled sound matches ones audiogram. The emerging demand for personalized hearing assistants requires the filter bank to be capable of decomposing the sound waves in accordance with the characteristic of the patients hearing loss. In this review paper, the application of multirate frameworks and filter banks for different procedures in personalized hearing assistants are investigated.

1. Introduction
Hearing aids are gadgets that mostly defeat sound-related shortfalls and are regularly utilized to make up for hearing-loss in hearing-impeded individuals. The fundamental target of a hearing aid is to fit the dynamic range of speech and regular sounds into the limited dynamic range of the hindered ear. With a specific end goal to accomplish a superior comprehension of this gadget and its capacity, we have to clarify how the sound is perceived by the human auditory framework and what are the issues experienced by hearing-disabled individuals [1].

The most well-known sort of hearing loss is sensorineural hearing loss (SNHL) in which the underlying cause lies in the vestibulocochlear nerve, the inner ear and the sound handling centers of the brain. Generally an issue with the cochlea, in the internal ear, in which the cilia don’t distinguish clues of specific frequencies as they should. Most hearing impedances fall into this class, so this is the issue that most hearing aids try to address. Sensorineural disability reduces the capacity of a man to recognize and detect one frequency within the presence different frequencies. Thus, a man with hearing misfortune has diminished capacity to hear a sound that quickly takes after, or is quickly trailed by an alternate sound. This diminished frequency makes it more probable for a hearing-impeded individual that noise will mask speech [2].
These days, multirate frameworks are applied in relatively every field of human interventions. Different applications of multirate frameworks are in speech processing for hearing aids, speaker and speech recognition, spectrum sensing in cognitive radios, signal compression for broadcasting, transmultiplexer design etc [3].

Whatever remains of this paper is sorted out as follows. As a matter of first importance, the basics of multirate frameworks and filter banks are introduced in section 2. In section 3, an overview of hearing aid systems are made. Subtle elements of auditory compensation techniques are talked about in section 4. In section 5, feedback or echo cancelation is talked about. The procedures relating to noise reduction is said in section 6. In section 7, speech enhancement for portable hearing assistants are clarified. At long last, conclusions are made in Section 8 and acknowledgements are specified in section 9.

2. Multirate Systems and Filterbanks

In digital signal processing, Multirate Systems utilizes different sampling rates at various parts of the framework. Multirate Systems primarily comprises of various operations like decimation, interpolation and resampling [4].

Decimation or down-sampling is the way toward lessening the sampling rate of a signal. The decimation factor is the ratio of the input sampling rate to the output sampling rate. It is normally symbolized by a positive integer $M$. It is performed by taking each $M$th sample and disposing of every single other samples. This causes $M$ fold expansion of the spectrum in frequency domain. An anti-aliasing filter with a cutoff of $\pi/M$ is utilized to band-limit the input signal $x(n)$. The decimation procedure in time domain and frequency domain are expressed in (1) and (2) respectively, where $x(n)$ is the input signal and $y(n)$ is the output signal.

$$y(n) = x(Mn)$$

$$Y(z) = \frac{1}{M} \sum_{k=0}^{M-1} X(z^{\frac{1}{M}} W_M^k)$$

Interpolation or up-sampling is the way toward expanding the sampling rate of a signal. The interpolation factor is the ratio of the input sampling rate to the output sampling rate. It is generally symbolized by a positive number $L$. It is performed by including $L-1$ zeros between every sample. This produces $L-1$ images of the spectrum in frequency domain. An anti-imaging filter with a cutoff of $\pi/L$ is utilized to expel these repetitive images. The interpolation procedure in time domain and frequency domain are given in (3) and (4) respectively, where $x(n)$ is the input signal and $x_u(n)$ is the output signal.

$$x_u(n) = \begin{cases} x(n/L), & n = 0, \pm L, \pm 2L, ... \\ 0, & otherwise \end{cases}$$

$$X_u(z) = X(z^L)$$

Resampling combines interpolation and decimation to change the sampling rate by a rational factor [5]. The resampling factor is typically spoken to by $L/M$. Interpolation is always preceded by decimation. The anti-aliasing and anti-imaging filters are combined to a single filter, which has a cut-off of $\min(\pi/L, \pi/M)$.

If a sequence $x(n)$ is band-limited, then it is possible to decimate it by use of appropriate multirate techniques. The desire to reduce the sampling rate whenever possible is completely understandable since it usually reduces the processing as well as the storage requirements. However if $x(n)$ is not band-limited but has most of its energy in the low frequency region, some
kind of data rate reduction is feasible even though it cannot be decimated without aliasing. This is indeed achieved by a technique called sub-band decomposition in which the average number of bits per sample is reduced while the average number of samples per unit time is unchanged. Sub-band Decomposition is implemented using filterbanks [6].

A digital filter bank is an accumulation of digital filters having a common input or a common output as shown in Fig. 4. The left framework is called an analysis bank and its filters $H_k(z)$
are the analysis filters. An input signal is splitted into M signals $v_k(n)$ typically called subband signals. We can process these individual bands independently and proficiently. The framework on the right part of the figure is a synthesis bank and $F_k(z)$ are the synthesis filters which joins the M-subband signals into a single signal $\hat{x}(n)$. Frequency responses can be marginally overlapping, non overlapping or very much overlapping, depending on the application. The magnitude response of a uniform filterbank with $M = 8$ is shown in Fig. 5.

3. Hearing Aid System Overview

Analog hearing aids give straightforward enhancement of the sound grabbed by the microphone and reproduced louder by the receiver or loud speaker. The amplification can be shaped marginally to consider distinctive hearing losses in spite of the fact that there is not always much flexibility. Many types of analog hearing aids can be programed by the user. They have a microchip which enables the device to have settings customized for various listening situations, for example, in a quiet place like in a library, or in a noisy place like in a hotel, or in an expansive region like a soccer field. The analog programmable hearing aids can store different programs for the different situations [7]. The client can easily select these modes when his listerning condition changes. An overview of analog hearing aid is shown in Fig 6.

Digital hearing aids take the sound waves from the microphone and change them to advanced digital binary code by the utilization of digital signal processing (DSP) strategies. This computerized data would then be able to be controlled in numerous sophisticated ways by the micro chip inside the amplifier, before being transformed back into ordinary analog signals and conveyed to the speakers. An overview of digital hearing aid is shown in Fig 7. Digital hearing aids, usually offer more highlights and adaptability, and are commonly client configurable. They have various memories to store more condition specific settings than analog hearing aids. They can also utilize modern algorithms to recognize and decrease background noise, to expel feedback and whistling, or to specifically distinguish the sound of voices and track them utilizing directional microphones. The distinctive aspects relating to digital hearing aid systems incorporate auditory compensation, feedback cancellation, noise reduction, speech enhancement etc [8].

4. Auditory Compensation

An audiogram is a diagram of the hearing test, that demonstrates the mildest sound a person can hear at various pitches or frequencies [9]. In the diagram, the Y axis shows the intensity or response of the ear estimated in decibels (dB) relating to the frequency in hertz (Hz) set apart on the X axis. An ‘O’ often is utilized to represent the responses for the right ear and a ‘X’ is utilized to represent the responses for the left ear. Curves showed in decibels generally shows the individual hearing limit of a man compared with the typical normal hearing, which is at 0
dB. Because of individual contrasts, all limits up to 20 dB are considered as would be normal. 20-40 dB is mild hearing loss, 40-70 dB is moderate hearing loss, 70-90 dB is severe hearing loss and above 90 dB is considered as profound hearing loss. Fig 8 shows the audiogram of a person with normal hearing. Fig 9 shows the audiogram of a person with moderate hearing loss in all frequencies. Auditory Compensation or audiogram matching is the process of adjusting the gains of individual bands of the filterbanks based on the patients hearing profile [10].

ANSI S1.11 provides the performance specification for the octave and fractional-octave-band filter bank. The ANSI S1.11 standard characterizes 1/3-octave bands covering the frequency range of 0-20 kHz. Every 1/3-octave band is indicated by its mid-band or central frequency $f_m$ and the bandwidth $\Delta f$. The mid-band frequency of the $n^{th}$ band, denoted by $f_m(n)$, is defined in (5).

$$f_m(n) = 2^{(\frac{n-30}{3})} \times f_r$$

where, $f_r$ the reference frequency, is set to 1 kHz.

In the ANSI S1.11 standard, three classes of filters are portrayed [11], i.e., class-0, class-1, and class-2. For each class of filters, a few parameters with respect to performance necessities are concerned, which incorporate the relative attenuation, the linear operating range, environmental sensitivities (e.g., humidity and temperature), the maximum output signal, the terminating impedance, etc. The specifications in either class-0 or class-1 filters are more stricter than that of class-2 filters. However, the ANSI S1.11 class-2 1/3-octave band has the comparable stopband attenuation requirement compared with other filter banks for hearing aids. Then again, the sampling rate is set to 24 kHz in order to provide the great sound quality [12].
5. Feedback Cancellation
Whenever a sound signal is recorded by a microphone, processed and then instantly played back
by a loudspeaker nearby the microphone, it is unavoidable that the loudspeaker signal is fed
back to the microphone. The depicted phenomenon is called acoustic feedback and it appears
in public address systems and in hearing aids. Acoustic feedback creates a closed signal loop,
which, under specific conditions, causes a phenomenon called howling effect. On account
of listening devices, the howling effect is extremely irritating for the hearing aid client and his
environment [13].

Numerous answers for this issue have been proposed, including phase modulation, gain
reduction, filtering in spatial domain and modeling the acoustic environment. The most common
approach to deal with acoustic feedback reduction is the adaptive feedback cancellation (AFC)
algorithm [14]. The acoustic feedback path is demonstrated by an adaptive filter and subtracted
from the input microphone signal. Thus, the AFC algorithm can cancel the acoustic feedback
totally when the adaptive filter converges to the perfect solution, i.e., the acoustic feedback path.

Fig. 12 displays the block diagram of a typical AFC system.

The idea of the adaptive feedback cancellation (AFC) approach is to assess the feedback path
by an adaptive filter and then to subtract the assessed feedback from the recorded signal before
it is amplified and played back. Because of the adaptive filter, the technique can adjust quick
changes of the feedback path, which is often the case in hearing aid situations. To calculate the
adaptive filter coefficients the normalized least mean squared (NLMS) algorithm is used which
minimizes the power of the error between the microphone signal and the assessed feedback signal,
the error is denoted as $e(n)$.

A idea to conquer this issue is to separate the sound signals into various frequency bands
by utilizing a filterbank and adapt each band independently [15]. Since we need to adjust in
each band independently, we apply the normalized least mean square (NLMS) algorithm based
calculation in each band. The advantage of this method is that we can adapt only in those bands
where the feedback is strong. This prompts less adjustment where the correlation occurs. In
this way, the bias is decreased, the sound signal is distorted to a lesser extent and the possible
gain is increased. The remaining task is to automatically control the adjustment in each band,
so that the echo is reduced to a feasible extent, the howling is prevented and the sound quality
is preserved for all different scenarios and sound signals [16].

6. Noise Reduction
The most clear impact of noise for individuals with hearing loss, is its effect on their ability to
converse. It impacts their capacity to convey each time they endeavor to have a discussion with
someone in a noisy place. Every day we are gone up against with wanted and unwanted noise.
Wanted noise like music or unwanted noise like a noisy group or background traffic noise. The
cause of the noise doesn’t make a difference, regardless of whether it is viewed as attractive or
bothersome, it can still tamper with a man’s capacity to get it [17]. Individuals with hearing
Figure 13. Overview of two-stage filterbank system for noise reduction.

Aids need to have the capacity to convey in all circumstances, much the same as individuals without hearing loss. There are various hearing aids includes that work in various different approaches to enhance speech perception in the presence of background noise for a hearing aid client. Directional microphones are one and noise reduction is the other.

A two-stage filterbank system is proposed by A. Schasse et al. [18] for reducing the single channel noise in hearing aids. Fig. 13 demonstrates the above two-stage filterbank system. The DFT coefficients processed in the Filterbank System (FBS) can be deciphered as subband signals with time index $m$ and subband index $k$. In this way, we can easily realize a second stage FBS to perform a further frequency analysis and synthesis of each subband signal. This prompts a short-time frequency-domain representation of the $k^{th}$ subband signal $Y(k, m)$ characterized as $Y_k(\kappa, \mu)$. The indexes $\mu$ and $\kappa$ demonstrate the time allotment and frequency bin within the second filter-bank stage. While applying no decimation in the second stage, the time allotment indexes of the first and second FBS are the same.

The time-domain signal $y(n)$ is decayed into $k$ frequency bands by the primary analysis filter-bank system (AFBS). The secondary stage of filterbank is only applied in lower frequency channels up to the subband index $k'$, while the rest of the bands are postponed w.r.t. the implementation of the second stage. At last, the time domain signal $\hat{x}(n)$ is the output of the last synthesis filterbank system (SFBS). A noise reduction algorithm, for example, the Wiener filter can be actualized in the second stage. All smoothing parameters, must be adjusted to the new channel bank settings to accomplish a similar measure of reduction in variance. The challenge of this two-stage design, is to accomplish a low overall signal delay [18].

7. Speech Enhancement

In ordinary circumstances, an individual can experience a variety of acoustic situations. For a person with a hearing aid following speech in various kinds of background noise can regularly face a challenge. Thus, assessing the signal-to-noise ratio (SNR) is a key factor to look at the hearing-aid design. Speech enhancement strategies can extract the speech signal in noisy environment. It is accounted that each improvement of 1 dB in noisy speech can bring about a 10% promotion in clarity of speech [19]. So, when the speech enhancement technique is utilized for a pre-treatment of noisy speech signal preceding to speech recognition, it will enhance the comfort of hearing aid in noisy environment and improve the comprehensibility of speech. Consequently, speech enhancement strategy for enhancing performance of hearing aid systems is desirable.

Speech enhancement technology can separate the speech signal in noisy condition and improve
the recognition rate. The hearing aid system using Kalman filter based speech enhancement can increase the rate of speech recognition and the hearing comfort of hearing disabled people in a noisy situation, compared with the hearing aid system using Wiener filter based speech enhancement [20].

8. Conclusion

Various applications of Multirate signal processing techniques and filterbanks in the field of hearing aids are reviewed in this paper. Nowadays, Multirate filterbanks are very commonly used on various parts of the hearing aids. These filterbanks are made reconfigurable to adjust the hearing aid suitable for different patients with individual hearing profiles. The different aspects pertaining to hearing aid systems include auditory compensation, feedback cancellation, noise reduction, speech enhancement etc are mentioned in this review paper.

References

[1] A. R. Moller, Hearing: Anatomy, Physiology and Disorders of the Auditory System, 2nd ed. New York, USA: Academic, Sep. 11, 2006.
[2] H. Dillon, Hearing Aids, New York: Thieme Medical Publisher, 2001.
[3] J. M. Kates, Digital Hearing Aids, Plural Pub., 2008.
[4] S. K. Mitra, Digital Signal Processing, A Computer-Based Approach, 2nd ed. Reading, PA: McGraw-Hill Irwin, 2001.
[5] P. P. Vaidyanathan, Multirate Systems and Filter Banks, New Jersey: Prentice Hall, 1993.
[6] R. Crochiere and L. Rabiner, Multirate digital signal processing, Englewood Cliffs, NJ, USA: Prentice-Hall, 1983.
[7] Christopher Schweitzer, "Development of Digital Hearing Aids", Trends in Amplification, Vol. 2, No. 2, 1997.
[8] A. M. Engebretson, "Benefits of digital hearing aids", IEEE Eng. Med. Biol. Mag., Vol. 13, No. 2, pp. 238-248, Apr. 1994.
[9] Yu-Ting Kuo, Tay-Jyi Lin, Yueh-Tai Li, Wei-Han Chang, Chih-Wei Liu and Shuen-Tsong Young "Design of ANSI S1.11 Filter Bank for Digital Hearing Aids", 14th IEEE International Conference on Electronics, Circuits and Systems, pp. 242-245, 2007.
[10] Yu-Ting Kuo, Tay-Jyi Lin, Yueh-Tai Li, Chou-Kun Lin and Chih-Wei Liu, "Low-power ANSI S1.11 filter bank for digital hearing aids", International Conference on Signals and Electronic Systems, pp. 347-350, 2008.
[11] Yu-Ting Kuo, Tay-Jyi Lin, Yueh-Tai Li, Chou-Kun Lin and Chih-Wei Liu, "Ultra low-power ANSI S1.11 filter bank for digital hearing aids" Asia and South Pacific Design Automation Conference, pp. 115-116, 2009.
[12] Yu-Ting Kuo, Tay-Jyi Lin, Yueh-Tai Li, and Chih-Wei Liu "Design and Implementation of Low-Power ANSI S1.11 Filter Bank for Digital Hearing Aids", IEEE Transactions on Circuits and Systems, Vol. 57, No. 7, July 2010.
[13] S. Haykin, Adaptive Filter Theory, Upper Saddle River, NJ, USA: Prentice-Hall, 2002. Vol. 2.
[14] T. van Waterschoot and M. Moonen, Adaptive feedback cancellation for audio applications, Signal Process., Vol. 89, No. 11, pp. 2185-2201, 2009.
[15] Falco Strasser and Henning Puder, Subband feedback cancellation with variable step sizes for music signals in hearing aids, in Proc. IEEE Int. Conf. Acoust., Speech, Signal Process. (ICASSP), May 2014, pp. 8207-8211.
[16] Falco Strasser and Henning Puder "Adaptive Feedback Cancellation for Realistic Hearing Aid Applications", IEEE/ACM Transactions on Audio, Speech and Language Processing, Vol. 23, No. 12, Dec 2015.
[17] A. Schasse, R. Martin, W. Sorgel, T. Pilgrim, and H. Puder, Efficient implementation of single-channel noise reduction for hearing aids using a cascaded filterbank, in Proc. ITG Symp. Speech Commun., Braunschweig, Germany, Sep. 2012, pp. 287-290.
[18] Alexander Schasse, Timo Gerkmann, Rainer Martin, Wolfgang Sorgel, Thomas Pilgrim, and Henning Puder "Two-Stage Filter-Bank System for Improved Single-Channel Noise Reduction in Hearing Aids", IEEE/ACM Transactions on Audio, Speech and Language Processing, Vol. 23, No. 2, Feb 2015.
[19] P. Vary, An adaptive filter bank equalizer for speech enhancement, Signal Process., Vol. 86, No. 6, pp. 1206-1214, Jun. 2006.
[20] Zheng Gong and Youshen Xia, "Two Speech Enhancement-Based Hearing Aid Systems and Comparative Study" 5th International Conference on Information Science and Technology (ICIST), 2015.