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A16BANDRECONFIGURABLEHEARINGAIDUSINGVARIABLEBANDWIDTHFILTERS

# A 16-Band Reconfigurable Hearing Aid using Variable Bandwidth Filters

James T. George [α] & Elizabeth Elias [σ]

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## I. Introduction

The main task of a hearing aid is to provide audibility, namely to amplify the required input signals to the levels that the hearing impaired can hear. An audiogram with frequencies on the X axis and hearing threshold on Y axis indicates the hearing capability of a person for various frequencies. An ideal hearing aid device should include features such as adjustable magnitude response on arbitrary frequencies, low power consumption for battery life, low processing delay to take advantage of lip-reading, linear phase to prevent the audio signal from distortion, small in size and weight for cosmetic appeal, noise reduction, programmability etc. (Yong and Ying, 2005). All these can be achieved by using the digital signal processing approach compared to the analog signal processing for implementing the hearing aid (Luo and Horst, 2002). Even though, the signal processing algorithms have dramatically improved in recent years, common hearing aid fitting procedures are still focusing on frequency bands and corresponding gain of the filters to match the audiogram of the specific user with individual hearing loss (Kuo et. al., 2008). So the basic requirement of a hearing aid is to provide precise adjustment of the gain versus frequency, which is also called audiogram matching/fitting. Sometimes, relatively minor changes may have a significant influence on performance, in particular with regard to speech intelligibility.

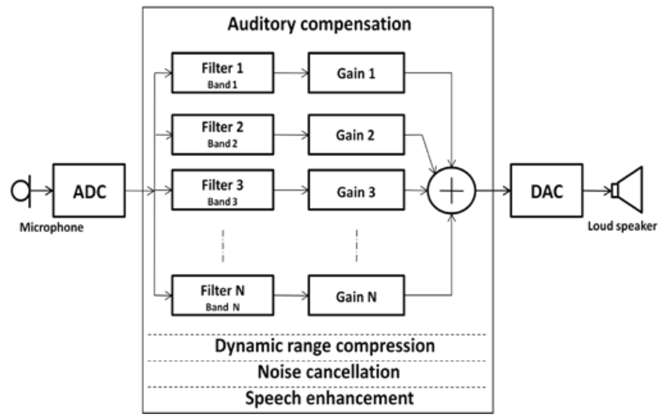


*Figure 1 :* Block diagram of digital hearing aid

Figure 1 shows the block diagram of generic hearing aid, including functional blocks such as auditory compensation, echo (feedback) cancellation, noise reduction/suppression, speech enhancement etc. (Kuo et. al., 2008). Auditory compensation is the main function in hearing aids, which performs frequency shaping to compensate for the hearing loss. An individual filter in the filter bank decomposes the input signal into (uniform or non-uniform) frequency bands. A low pass filter outputs the lowest frequencies, a high pass filter outputs the highest frequencies and a set of band pass filters output the remaining intermediate frequencies as shown in Figure 2, so that the prescribed gain based on the audiogram or hearing threshold can be applied to the different frequency bands to suit the needs of the hearing impaired people, or in other words the amplitude response of the filter bank should equalize or match the audiogram (McAllister et. al., 1994).

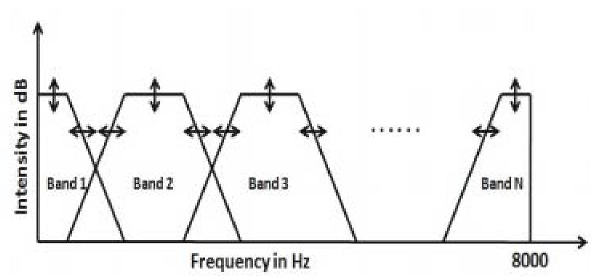


*Figure 2 :* General filter structure

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Most of the currently available hearing aid designs provide the filter bank with fixed bands (uniform or non-uniform). Thus the patients cannot take the full advantage to improve their specific auditive performance by using the hearing aid with limited number of fixed bands. This reduces the potential flexibility in fitting/matching of hearing loss with steeply sloping audiograms. One method of improving the same is to use an instrument with higher number of frequency bands for matching the audiogram with minimum matching error. But such an implementation would not only require a large amount of processing power but also increases the cost. Modern hearing aids with more than/up to 32 bands are available in the market. Therefore it would be very useful to design a filter structure with less number of bands and that can be easily customized with minimum change in parameters, for any hearing impaired person.

In this paper, a new method is proposed to realize a uniform as well as non-uniform filter structure for the hearing aid application with the facility for changing the band edges and gain to match the audiogram without any change in the number of bands and filter coefficients of the fixed filters associated with each band. Variable bandwidth filter with facility for the reduction and enhancement of the bandwidth without changing the filter order or filter coefficients, based on changing the sampling frequency of the input signal, is used for achieving the same (Fred, 2009, George and Elias, 2012). It is accomplished by interpolating the input series to obtain an intermediate time series at a new sample rate proportional to the required bandwidth. The interpolated series are processed by the fixed length FIR filter, now operating at the changed sample rate. The output of the filtered time series is then interpolated back to the original input sample rate, which effectively changes the bandwidth. The change in bandwidth of the particular band can be achieved by modifying only the bandwidth ratio given as input to the sample rate converter. All the circuits associated with each band can be implemented in fixed hardware also. With the help of the variable bandwidth filter approach, a wide range of hearing loss compensation can be achieved by using the same fixed hardware. In this case, the fixed filter and the associated sample rate converters can be implemented as fixed hardware in application specific integrated circuit (ASIC) and the programmable portion of the processing can be implemented in a digital signal processor/ micro-controller, thereby achieving small chip size and low power consumption (McAllister et. al., 1994).

The paper is organized as follows. Section 2 gives a brief overview about the audiogram and the common hearing loss patterns. Section 3 presents a review of the variable bandwidth filter. Section 4 gives the design example of the fixed filter and the structure of a 16 band non-uniform filter bank. The performance evaluation and comparison of the results are given in Section 5 and Section 6 concludes the paper.

## II. Overview of Audiogram and Hearing Loss

An audiogram is a graph the hearing test is marked on, that shows the softest sound a person can hear at different pitches or frequencies. The Y axis represents the intensity or response of the ear measured in decibels (dB) corresponding to the frequency in hertz (Hz) marked on the X axis. An `O' often is used to represent the responses for the right ear and an `X' is used to represent the responses for the left ear. Curves displayed in decibels generally describe the individual hearing threshold of a person compared to the normal hearing average, which lies around 0 dB. Due to individual differences, all thresholds up to 20 dB are considered as normal. Figure 3 shows the audiogram of a normal hearing person. The threshold between 21 to 40 dB is considered as mild hearing loss, 41 to 55 dB is considered as moderate hearing loss, 56 to 70 dB as moderately severe, 71 to 90 dB is considered as severe hearing loss and greater than 90dB is considered as profound hearing loss.

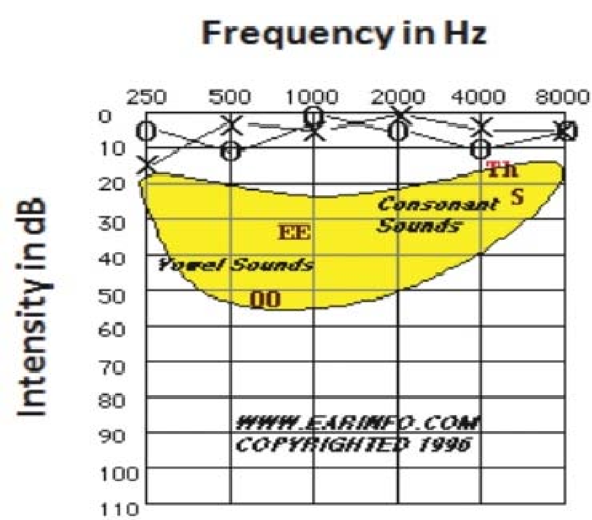


*Figure 3 :* Audiogram for normal hearing(www.earinfo.com)

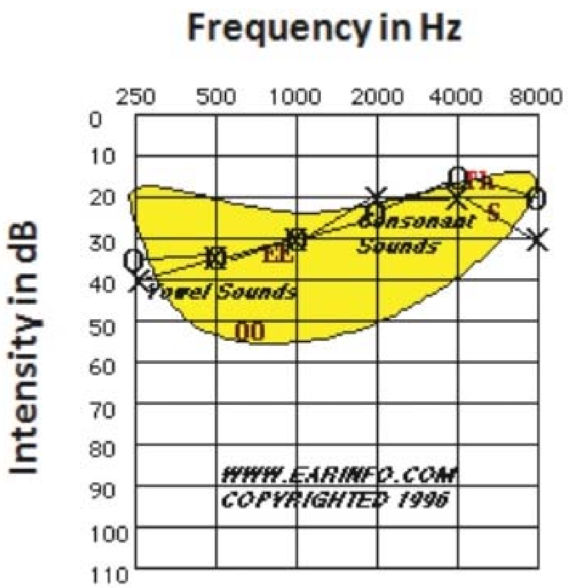


*Figure 4 :* Audiogram for mild to moderate hearing loss at low frequencies (www.earinfo.com)

Loudness level and frequencies of different speech sounds are also presented in audiograms. Because of the shape of the speech area, it is also referred to as `speech banana'. Vowels are low frequency sounds with a higher volume than consonants, which are soft high frequency sounds. The vowels carry the loudness impression of speech whereas consonants carry the meaning.

Hearing level measurements are done at each octave 250/500/1k/2k/4k/8k in a standard audiogram, which suggests that the uniform filter bank may face difficulties in matching the audiogram at all frequencies (Yong and Ying, 2005). A typical hearing loss shown in Figure 4, caused by aging, occurs at low frequencies. To achieve a better compensation, narrower bands with gain control need to be allocated at frequencies where

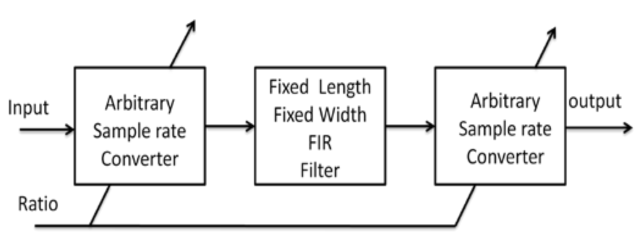


*Figure 5 :* Functional block diagram of variable bandwidth filter

the slope is steep. Therefore, a non-uniformly spaced digital filter bank structure with varying band edges using variable bandwidth filter becomes very useful. Several methods are described in the literature to realize the filter structure for hearing aid applications using combinations of techniques such as infinite impulse response (IIR), Finite impulse response (FIR), frequency response masking (FRM), frequency transformation, uniform and non-uniform filter banks etc. (Ito et. al., 2010, Deng, 2010, Brennan and Todd, 2001, Chong et. al., 2006, Lim, 1986, Ying and Yong, 2006, Ying and Debao, 2011), but all of them have fixed band edges with particular design. The next section explains the operation of the variable bandwidth filter for the hearing aid application with provision for re-configurability in band edges (George and Elias, 2012).

## III. Review of Variable Bandwidth Filter

There are several methods (Oppenheim et. al., 1976, Roy and Ahuja, 1979, Hazra, 1984, Carson et. al., 2002, Johansson and Lowenborg, 2004, Mahesh and Vinod, 2011) available for the design of variable bandwidth filters for signal processing and communication application. Most of them require changes in the filter coefficients for achieving the revised frequency characteristics. For hearing aid application, the FRM based technique also requires modified structural design for getting the new frequency characteristics. But the fractional variation of the bandwidth (reduction as well as enhancement) of the fixed bandwidth prototype filter, without any change in the coefficients can be achieved by changing the sampling rate of the input signal (Fred, 2009, George and Elias, 2012).

In the case of constant form factor FIR filter, the length of the filter is inversely proportional to the transition bandwidth, which is proportional to the bandwidth of the filter. Thus if the bandwidth of the filter is changed by a factor, the length of the filter is also changed by the same factor.

In order to realize a variable bandwidth filter without changing the number of coefficients or coefficient values, we can utilize the relationship between the sampling rate and bandwidth (Fred, 2009). If the number of taps of the filter is fixed, one method of changing the bandwidth is to change the sampling rate. So we can change the absolute bandwidth of a filter by operating it at a different sampling rate. The technique proposed by us in George and Elias, 2012 is used for changing the bandwidth continuously in the decreasing and increasing fashion.

The implementation of the process is represented in Figure 5. An input signal which is initially oversampled is applied to an arbitrary sample rate converter (up or down). The modified signal is signal is processed by the fixed length, fixed bandwidth FIR filter and the output of the filter is then converted back to the original input sampling rate by using another arbitrary sample rate converter (down or up). This effectively changes the bandwidth.

Figure 6 shows the implementation of an arbitrary sample rate converter (Fred, 2004 ). Three $M$ - path polyphase filters are used for calculating the sample values of the interpolant and the sample

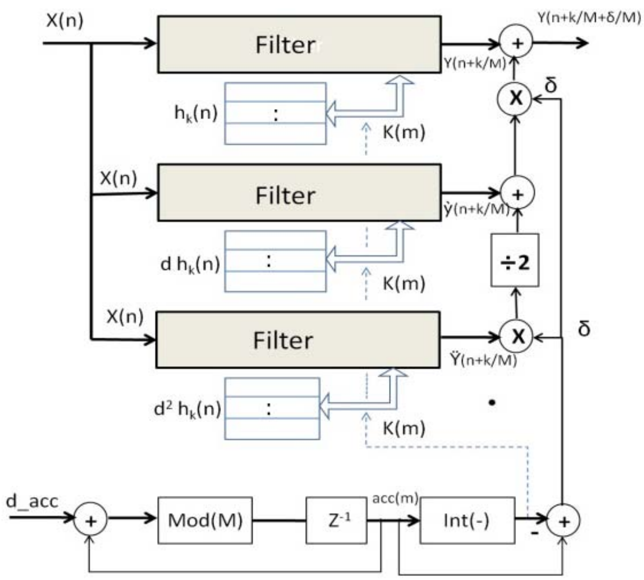


*Figure 6 :* Implementation of arbitrary sample rate converter

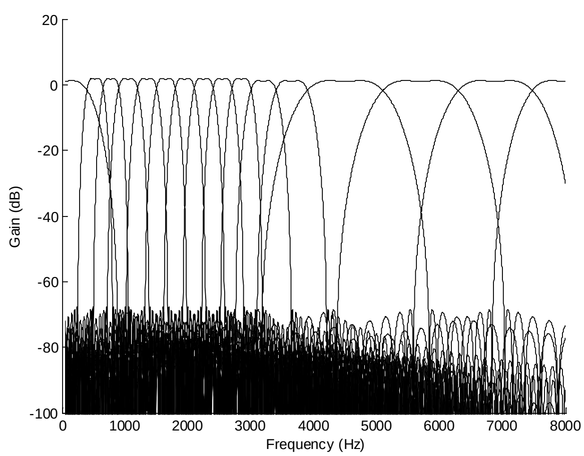


*Figure 7 :* Frequency response of the uniform filter bank

derivatives at the offset position $k/M$ from the interpolating output phase centre. The computed output is formed from a local Taylor series as given by Equation (1).

$$y\left(n+\frac{k\delta}{M}+\frac{}{M}\right)\cong y\left(n+\frac{k}{M}\right)+\acute{y}\left(n+\frac{k}{M}\delta\right)+\frac{^2}{2!}\ddot{y}\left(n+\frac{k}{M}\right)+\ldots \quad (1)$$

The increment *d_acc* satisfies the following relation shown in Equation (2) and the required bandwidth can be achieved by just changing the parameter *d_acc* as shown in Equation (3).

$$\frac{d\_acc}{M}=\frac{T_{out}}{T_{in}}=\frac{fs_{in}}{fs_{out}}=\textit{Arbitrary sampling ratio} \quad (2)$$

$$d\_acc=M\ *sampling\left(bandwidth\right)ratio \quad (3)$$

Where

$M$ - Number of polyphase filters

$fs_{in}$ - Input sampling rate to interpolator

$fs_{out}$ - Output sampling rate of interpolator

## IV. Design Example

Finite impulse response (FIR) filters are widely accepted in many areas of signal processing, communication and bio-medical applications including hearing aid due to their exact linear phase and high stability under certain conditions. In the design of the variable bandwidth filter, we should preserve the finite length and linear phase for a desired set of frequency characteristics. Therefore priority should be given for the optimal design of the fixed bandwidth FIR filter.

Let us consider a typical example for designing a prototype fixed bandwidth FIR filter for a 16 band hearing aid to support up to 8 kHz. The design specifications of the low pass FIR filter are shown below.

*Bandwidth : 269 Hz*

*Maximum Pass band ripple: 0.05dB*

*Minimum stop band attenuation: 65dB*

*Sampling frequency: 16kHz*

*Table 1 :* Bandwidth, ratio and frequency shift of sub-bands

| Band | Uniform | | | Non-uniform case-1( [12]) | | | Non-uniform case-2 | | |
|---|---|---|---|---|---|---|---|---|---|
| | Band-width (Hz) | Shift (Norma-lised) | Band-width Ratio | Band-width (Hz) | Shift (Norma-lised) | Band-width Ratio | Band-width (Hz) | Shift (Norma-lised) | Band-width Ratio |
| 1 | 269 | 0 | 1 | 250 | 0 | 0.9375 | 400 | 0 | 1.5001 |
| 2 | 533 | 0.0335 | 1 | 250 | 0.0234 | 0.4688 | 250 | 0.0328 | 0.4688 |
| 3 | 533 | 0.0668 | 1 | 250 | 0.0391 | 0.4688 | 250 | 0.0484 | 0.4688 |
| 4 | 533 | 0.1001 | 1 | 250 | 0.0547 | 0.4688 | 300 | 0.0656 | 0.5625 |
| 5 | 533 | 0.1334 | 1 | 500 | 0.0781 | 0.9375 | 300 | 0.0844 | 0.5625 |
| 6 | 533 | 0.1667 | 1 | 500 | 0.1094 | 0.9375 | 300 | 0.1031 | 0.5625 |
| 7 | 533 | 0.2000 | 1 | 1000 | 0.1563 | 1.8750 | 300 | 0.1219 | 0.5625 |
| 8 | 533 | 0.2333 | 1 | 1000 | 0.2188 | 1.8750 | 300 | 0.1406 | 0.5625 |
| 9 | 533 | 0.2667 | 1 | 1000 | 0.2813 | 1.8750 | 300 | 0.1594 | 0.5625 |
| 10 | 533 | 0.3000 | 1 | 1000 | 0.3438 | 1.8750 | 300 | 0.1781 | 0.5625 |
| 11 | 533 | 0.3333 | 1 | 500 | 0.3906 | 0.9375 | 400 | 0.2000 | 0.75 |
| 12 | 533 | 0.3666 | 1 | 500 | 0.4219 | 0.9375 | 500 | 0.2281 | 0.9375 |
| 13 | 533 | 0.3999 | 1 | 250 | 0.4453 | 0.4688 | 1200 | 0.2813 | 2.25 |
| 14 | 533 | 0.4332 | 1 | 250 | 0.4659 | 0.4688 | 1200 | 0.3563 | 2.25 |
| 15 | 533 | .4665 | 1 | 250 | 0.4766 | 0.4688 | 1175 | 0.4305 | 2.203 |
| 16 | 269 | 0.5 | 1 | 250 | 0.5 | 0.9375 | 525 | 0.5 | 1.968 |

*Sampling frequency: 16kHz*

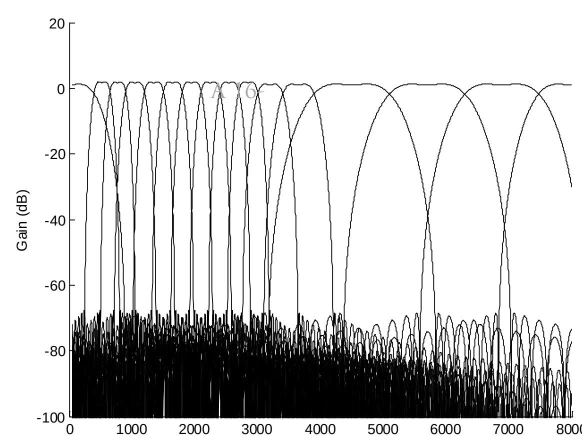


*Figure 8 :* Frequency response of the non-uniform case-2 filter bank

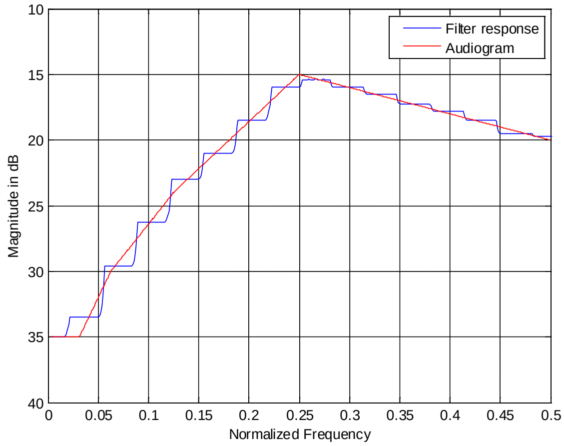


*Figure 9 :* Audiogram fitting for the uniform filter bank

The bandwidth, corresponding shift in centre frequency and bandwidth ratio for the various bands in the uniform and two examples/cases of the non-uniform bandwidth selection are given in Table 1. The frequency response of the filters in the uniform and non-uniform cases based on Table 1 are shown in Figure 7, 8 and 9 respectively.

The bandwidth of the particular band can be changed by modifying the bandwidth ratio *d_acc* and the corresponding centre frequency shift. The centre frequency shift can be achieved by using the spectrum shifting property of the Fourier Transform (Vaidyanathan, 1990 ).

## V. Audiogram Matching/Performance Evaluation/Result and Discussions

In order to evaluate the performance of the 16 band filter bank for the hearing aid application, the structure is tested by using MATLAB 7.10.0 on a Toshiba satellite L750 laptop with Intel (R) core (TM) i5 2410M processor operating at 2.3 GHz. Various audiograms for the common types of hearing loss are used for evaluating the effectiveness of the filter bank. It is observed that the matching error between the filter output and the audiogram is very small.

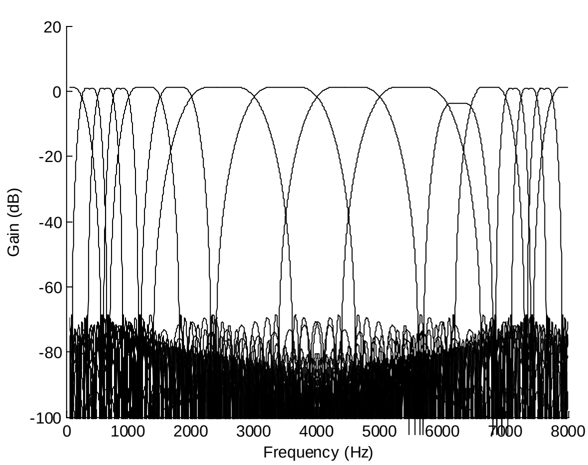


*Figure 10 :* Frequency response of the non-uniform case-1 filter bank

The results corresponding to the audiogram shown in Fig 4 are used for presenting the output. The purpose in selecting the mild to moderate hearing loss at low frequencies is that the audiogram contains steep slope up to the middle of the audio range. The audiogram fitting/ matching with the 16 band uniform and non-uniform cases based on the data given in Table 1. along with the selected audiogram are given in Fig 10, 11 and Fig 12 respectively. The corresponding errors between the audiogram and the magnitude responses of the filter banks in all the three cases are given in Fig 13. The advantage of the proposed method is that, the same hardware setup can be configured as uniform or non-uniform frequency bands based on the selection of the bandwidth ratio and frequency shift. The proper selection of the bandwidth or bandwidth ratio and frequency shift will decide the quality of the output.

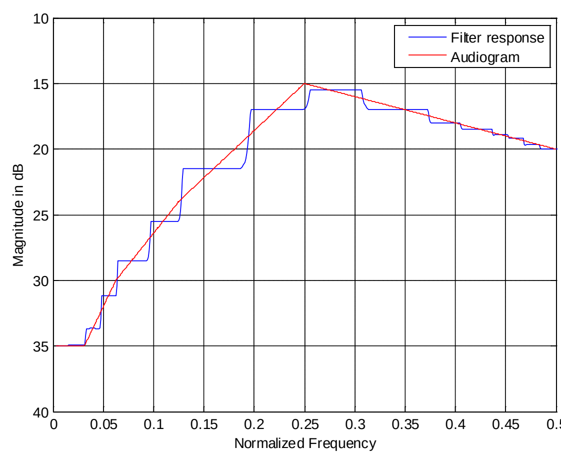


*Figure 11 :* Audiogram fitting for the non-uniform filter bank (case-1)

For verifying/analyzing the performance of the proposed method, the number of bands and bandwidth corresponding to each band are selected as given in Ying and Yong, 2006 for the non-uniform case -1. A modified selection of bandwidth for reducing the matching error is proposed as case-2 in Table 1. The matching error corresponding to each case is given in Table 2. In the case of uniform bands, the only possibility for reducing the matching error is to adjust the gain of the bands. Because of the wider bandwidth provided in the mid frequency region having large slope, further reduction in the matching error is not possible with the band allocation as specified in Ying and Yong, 2006 and given as case-1 in Table 1. In the case of non-uniform case-2, the control of band gain and selection of smaller bandwidth in those positions where the slope is steep, can reduce the matching error considerably as shown in Table 2.

*Table 2 :* Matching errors of the uniform and non-uniform filter banks

| Maximum matching error in dB | | |
|---|---|---|
| Uniform | Non-uniform (case-1) | Non-uniform (case-2) |
| 1.59 | 2.1 | 1.08 |

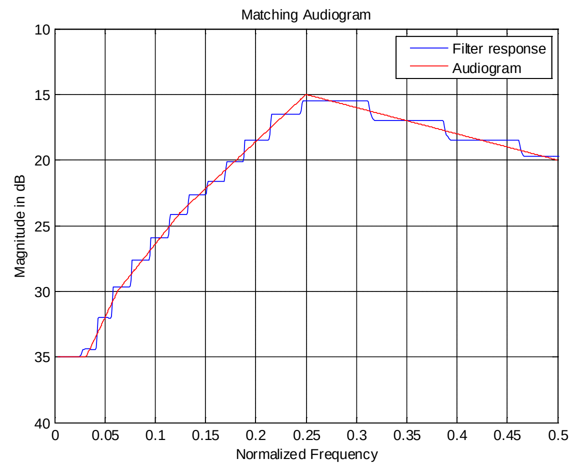


*Figure 12 :* Audiogram fitting for the non-uniform filter bank (case-2)

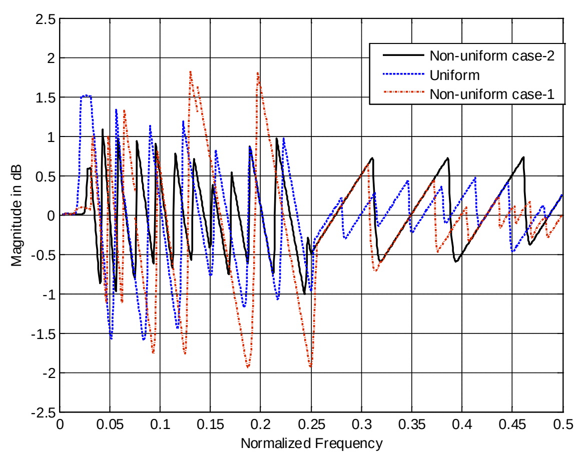


*Figure 13 : Comparison of matching errors*

## VI. Conclusion

In this paper a technique is presented to obtain a 16 band uniform or non-uniform spaced filter bank using variable bandwidth filter to suit the fitting of audiogram for the hearing aid application. Each band is implemented as a combination of two arbitrary sample rate converters and a fixed bandwidth FIR low pass filter, which can be implemented in a fixed hardware. The advantage of the proposed scheme is that, the bandwidth and the position of the band can be configured by modifying the bandwidth ratio and frequency shift, based on the audiogram of the hearing impaired, at the time of hearing aid fitting, by an audiologist. This part of the hardware can be implemented in a FPGA/DSP/micro-controller part associated with the programmable hearing aid. The performance of the filter bank is evaluated with various audiograms for common types of hearing loss and the results are comparable with available/existing ones in the literature. Miniaturization, reduction in power consumption, programmability and affordable delay can be achieved by the implementation of the major part in the fixed hardware. Hearing aid fitting becomes easier and re-configuring the bands is also possible with minimum modification in the programmable part. Hence the same hearing aid can be easily re-programmed for the changes which can happen in the audiogram due to the aging of the user. Thus the re-usability of the hearing aid can be improved by the re-configurability.

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