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Efficient variable bandwidth filters for digital hearing aid using Farrow structure

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A B S T R A C T

Design of a digital hearing aid requires a set of filters that gives reasonable audiogram matching for the concerned type of hearing loss. This paper proposes the use of a variable bandwidth filter, using Farrow subfilters, for this purpose. The design of the variable bandwidth filter is carried out for a set of selected bandwidths. Each of these bands is frequency shifted and provided with sufficient magnitude gain, such that, the different bands combine to give a frequency response that closely matches the audiogram. Due to the adjustable bandedges in the basic filter, this technique allows the designer to add reconfigurability to the system. This technique is simple and efficient when compared with the existing methods. Results show that lower order filters and better audiogram matching with lesser matching errors are obtained using Farrow structure. This, in turn reduces implementation complexity. The cost effectiveness of this technique also comes from the fact that, the user can reprogram the same device, once his hearing loss pattern is found to have changed in due course of time, without the need to replace it completely.

Introduction

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Hearing loss patterns differ according to the anatomical and sensorineural differences. For example, Presbyacusis is an age related hearing loss. It usually affects the high frequencies more than the low frequencies [1]. The softest sound that can be recognized in the frequency range 250–8000 *Hz* is represented in an audiogram. Any sound that is heard at 20 dB or quieter is considered to be within the normal range [2]. For the patients with hearing losses, certain kinds of hearing aids are required to improve the quality of hearing. An important unit of a digital hearing aid consists of the digital filters that can tune the amplitudes selectively to a person's particular pattern of hearing loss. In case of Presbyacusis, simple amplification merely makes the garbled speech, sound louder [2]. They usually need a hearing aid that selectively amplifies the high frequencies. Thus, the filtering unit should be able to provide gain selectively to different frequency bands. This allows the filter response of the hearing aid to have minimum matching error response relative to the audiogram, within a tolerance limit. 3 dB can be taken as the limit, as most people are not sensitive to lower errors [3].

A good amount of flexibility, minimum hardware, low power consumption, low delay and linear phase (to prevent distortion) are the required characteristics of any digital

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hearing aid. Significant amount of study is available on the bank of filters designed for audiogram matching. Initial approaches were based on uniform subbands. Since, humans perceive loudness on a logarithmic scale, non-uniform filter banks are better suited, so that the matching can be achieved with minimum number of sub-bands, if possible. Some of the methods used to generate non-uniform subbands for digital hearing aid application, as found in the literature, are as follows.

A frequency response masking technique using two prototype filters [4], is employed to generate an 8-band non-uniform FIR digital filter bank. Matching errors are reported to be better compared to 8-band uniform filter bank and the number of multiplications is lower since half-band filters are used. However, the delay introduced is large and delays more than 20 ms may hamper with lip-reading [5]. This problem was addressed by using a similar method, but with three prototype filters generating 16 bands by Wei and Lian [5]. Still, for lower matching errors, better precision in designing the filters and their cascade and parallel placements, are to be taken care of, which would increase the design cost. An approach using variable filter-bank (VFB) that consists of three channels having separately tunable gains and band edges, is considered by Deng [3]. The method has increased flexibility, but the use of infinite impulse response (IIR) digital filters introduces overall non-linear phase to the system. Wei and Liu [6] give a flexible and computationally efficient digital finite impulse response (FIR) filter bank based on frequency response masking (FRM) and coefficient decimation. The frequency range is divided into three sections and each section has three alternative subband distribution schemes. The decision on selecting the sections for each sub-band for the selected audiogram has to be made wisely and the flexibility of the system is limited by this selection.

A change in the design methodology can be found in the approach by James and Elias [1], where, a variable bandwidth filter using sampling rate conversion technique, is used for the digital hearing aid application. The filter order or filter coefficients need not be altered to obtain the variability in the bandwidth. A fixed length FIR filter is designed initially, whose characteristic bandwidth is then changed by modifying the bandwidth ratio, given as input to an interpolation filter. Using this filter structure and by varying the bandwidth ratio, a bank of filters that processes different subbands, is realized. However, the hardware complexity of the structure is seen to be high.

This paper proposes the design of a bank of digital filters that can provide reasonably good matching with the set of audiograms considered. A variable bandwidth (VBW) filter, whose bandwidth can be varied dynamically, is implemented using Farrow structure. All the required bandwidths for the set of selected audiograms are derived from the VBW filter. These filters are then tuned separately to the optimum center frequencies and bandwidths to match each of the audiogram. Thus, once the VBW filter is designed using the proposed technique, the instrument can be tuned by the manufacturer to individual user audiogram characteristics. This results in an efficient method to realize reconfigurable digital hearing aid. A primitive form of this work is done by us for a single audiogram and is published in a conference proceeding [7].

An adjustable hearing aid helps the user to adjust the device according to the change in hearing loss pattern with time or age. Yet another advantage is that the vendors of hearing aid can design an instrument to suit a set of hearing loss patterns. Here, it can be customized for any of its users, using a small set of tuning parameters. The proposed method aims to design a reconfigurable filter structure to suit a set of hearing loss patterns. Consequently, the cost of the instrument can be lowered without compromising on the quality.

Section "Methodology" explains how Farrow based variable bandwidth filters can be used in digital hearing aid. In Section "Results and discussion", the efficiency of the method is verified on a set of audiograms by comparing with an existing method. The method is also applied to audiograms of real patients in the same section. Section "Conclusion" concludes the paper.

Preliminaries – Farrow structure

The design of the subbands in the digital hearing aid scenario given in this paper, is based on a variable bandwidth filter. There are many ways in which filters with adjustable band-edges are approached in the literature [8].

We propose the Farrow structure implementation for the set of variable bandwidth filters used in the digital hearing aid. In the Farrow structure, the overall response is derived as a weighted linear combination of fixed subfilters as shown in Fig. 1 [9]. The weights control the tunable bandwidths.

The Farrow structure was initially derived as a digital delay element, where the desired impulse response is approximated using $(L+1)$th- order polynomials of a delay parameter, d, [10]. Later, modified Farrow structure was proposed by Johansson and Lowenborg [9], where the subfilters are designed to have linear phase (symmetric coefficients), which also reduces the overall implementation complexity. Farrow structure is an efficient way to realize tunable filter characteristics such as variable fractional delay [9,11,12], sampling rate conversion (SRC) [13,14] and variable cut-off frequencies [15]. In a variable fractional delay filter, all the input samples are delayed by a factor, whereas in SRC, every input sample is delayed by varying factors.

An ideal frequency response of an FIR filter, $A_{ideal}(e^{j\omega})$ of order N can be written such that the magnitude and phase responses are expressed with polynomial coefficients of ω as given by Luo et al. [16],

$$A_{ideal}(e^{j\omega}) = \left(\sum_{n-0}^{N} a_n\omega^n\right) e^{-j[(N/2)\omega+\sum_{m=1}^{M} b_m\omega^m)} \quad (1)$$

where M is the order of phase response and $\sum_{m=1}^{M} b_m\omega^m$ is the fractional delay, d, in a Farrow structured fractional delay filter. This can be rewritten with unity magnitude as,

$$A_{ideal}(e^{j\omega}) = e^{-j[(N/2)\omega+\sum_{m=1}^{M} b_m\omega^m)} \quad (2)$$

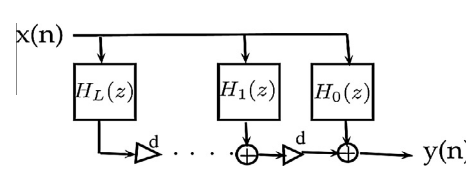


Fig. 1 The Farrow structure [9].

The frequency response can be controlled by adjusting the polynomial coefficient b_m. Each polynomial phase component can be approximated [16] using Taylor series of ω, with an error ϵ,

$$e^{-jb_m\omega^m} = \sum_{p=0}^{P} \frac{(-jb_m\omega^m)^p}{p!} + \epsilon \quad (3)$$

where P is the order of Taylor series for each polynomial phase component and the Taylor approximation error ϵ. Thus, the approximated frequency response for the fractional delay filter is,

$$\begin{aligned} A_{approx}(e^{j\omega}) &= e^{-j(N/2)\omega} \prod_{m=1}^{M} \sum_{p=0}^{P} \frac{(-jb_m\omega^m)^p}{p!} \\ &= e^{-j(N/2)\omega} \sum_{q=0}^{Q} c_q \omega^q \end{aligned} \quad (4)$$

where the coefficient c_q is derived from the polynomial phase component which is related to the fractional delay d as $c_q = d^q$ [16]. The frequency response can be rewritten as

$$A_{approx}(e^{j\omega}) = \sum_{q=0}^{Q} d^q H_q(e^{j\omega}) \quad (5)$$

where $H_q(e^{j\omega}) = \omega^k \exp^{-j(N/2)\omega}$ is the linear phase FIR subfilters of the Farrow structure, shown in Fig. 1. The corresponding transfer function for $z = e^{j\omega}$ is given as,

$$A_{approx}(z) = \sum_{q=0}^{Q} d^q H_q(z) \quad (6)$$

$H_q(z)$ in Eq. (6) are the subfilters in the Farrow structure designed by means of approximation. $A_{approx}(z)$ denotes the transfer function of the system in Fig. 1. It is related to the input and output as,

$$A_{approx}(z) = Y(z)/X(z) \quad (7)$$

where $Y(z) = \sum_{n=-\infty}^{+\infty} y(n)z^{-n}$ and $z = e^{j\omega}$.

The subfilter design can be carried out for the same or different order and can be used according to the requirement. Different order subfilters are found to be better in terms of complexity [9]. Further complexity reduction could be achieved by replacing the multipliers in the implementation by means of adders and shifters [17]. This is carried out by expressing the filter coefficients as signed-power-of-two (SPT) terms.

Variable bandwidth filter using Farrow structure

Farrow structure based variable bandwidth filters were introduced very recently when compared to their use as fractional delay filters. An initial attempt to design a filter with varying cut-off frequency is done by Pun et al. [18]. Here, the FIR filters are designed using Parks–McClellan algorithm for a set of evenly spaced bandwidths within the tunable range, which is then interpolated by an Lth degree polynomial in b, denoting the bandwidth. The variability is achieved by updating the adjustable parameters, which directly depends on the bandwidth. When the multipliers in this structure are quantized, it causes high overall implementation complexity due to the roundoff noise. This could be overcome by adopting a fixed parameter, b_0 [13,15], along with the variable bandwidth factor, b. The fixed parameter is selected as the mid-point between the desired bandwidths. Thus, the approximate transfer function is written as function of z and b as,

$$A(z,b) = \sum_{l=0}^{L} (b-b_0)^l H_l(z) \quad (8)$$

where $H_l(z)$ are N_lth order linear phase FIR subfilters [15]. The error function is defined as the difference between the ideal and approximate frequency responses, $A_{ideal}(z,b)$ and $A(z,b)$ respectively and is given by $E(z)$ as,

$$E(z) = A(z,b) - A_{ideal}(z,b) \quad (9)$$

One of the techniques to minimize the squared error, which is widely used along with weights to emphasize certain frequencies, is the weighted least squares design approach. If it is desired to minimize the peak approximation error, it is suitable to use the minimax design. These approximation problems can usually be solved only by iterative techniques, such as linear programming. The required filter specifications can be stated as

$$\begin{aligned} 1-\delta_c(b) \leqslant |\, A(e^{j\omega T},b) \,| &\leqslant 1+\delta_c(b), \omega T \in [0, b-\Delta(b)] \\ |\, A(e^{j\omega T},b) \,| &\leqslant \delta_s(b), \omega T \in [b+\Delta(b), \pi] \end{aligned} \quad (10)$$

for $b_l \leqslant b \leqslant b_u$, where $[b_l, b_u]$ is the range of the desired bandwidth. $b-\Delta(b)$ to $b+\Delta(b)$ is the range of transition width at each of the designed bandwidth b. $\Delta(b)$ is half of the transition width. δ_c and δ_s are the passband ripple and stopband attenuation respectively. The weighted error function is given by,

$$E(\omega T,b) = W(\omega T,b)[A(\omega T,b) - A_{ideal}(\omega T,b)] \quad (11)$$

where $W(\omega T,b)$ is unity for passband and ratio of specified ripples $(\frac{\delta_c}{\delta_s})$ for stopband. This approximation problem can be solved to have global optimum solution in the minimax sense using linear programming [15]. The frequency range and required bandwidths are discretized initially and the problem is restated as

$$minimize\ max\ \ |\, E(\omega_i T, b_j) \,| \quad (12)$$

where i,j are the discrete points used for optimization. Eq. (12) is the objective of the optimization problem to minimize the maximum of the weighted error between ideal and the approximate transfer function response of the variable bandwidth filter. This error is not related to the matching error of the final hearing aid, which is the difference between audiogram and the response of the bank of filters with appropriate magnitude gain and frequency shift.

Methodology

In order to design the non-uniform bandwidth filters, we propose to initially design a VBW filter using Farrow structure as described above. The filter structure shown in Fig. 1 can be designed to meet the specification for each of the variable bandwidth parameters, b, such that there is complete control on the desired specifications and performance. As mentioned in the introduction, this approach to design the sub-bands for digital hearing aid is relatively unattempted. In the work of James and Elias [1], tuning of the designed fixed filter is carried out by means of sampling rate conversion (SRC) filter. Using

Farrow structure in this approach, is so far not reported in the literature.

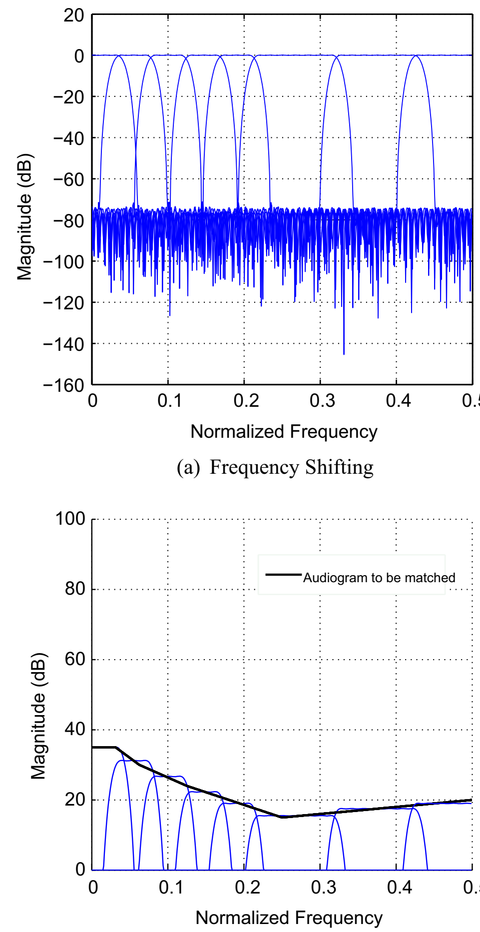
Initially, from the selected hearing loss patterns, a set of bandwidths, b_{set}, that could be used to fit the audiograms, is chosen. A variable bandwidth filter is designed to realize these bandwidths (b_{set}) using Farrow structure. The subfilters in this paper are designed only once and is a fixed hardware implementation for a set of bandwidths for which the system is designed. The variability is achieved only by altering the variable factor, b, for each implemented filter. The coefficients of the filter are fixed. The fixed parameter, b_0 can be chosen to be the midpoint between the minimum and maximum bandwidths from the selected set. The order of the Farrow subfilter is dependent on the specified frequency response characteristics. The optimum transition bandwidth of the VBW filter is selected such that all the audiograms under consideration can be matched within a tolerable error limit. It is observed that some audiograms are better fitted with wider transition bandwidths. Also, the number of subfilters required, depends directly on the number of bandwidth points selected for the design. The filters $H_l(z)$ are obtained by means of linear programming, such that the overall transfer function $A(z, b)$, achieves the specifications within tolerable limits. Fig. 2 shows an example response obtained when designed for the frequencies 500 Hz, 750 Hz and 1000 Hz normalized to 8000 Hz. The filter specifications for this variable bandwidth filter are:

Passband Ripple = 0.05 dB.

Stopband Attenuation = 80 dB.

The bands, thus obtained using VBW filter, are to be shifted appropriately using the spectrum shifting property [7]. The proper magnitude gain is provided for each band by trial and error approach until it matches with the given audiogram. The maximum of the overall response forms an approximation of the audiogram. If proper shifts are used, this would consist of only the passbands of the shifted filter responses. As an example, an audiogram of mild hearing loss at all frequencies is selected and matched using the above bands. This is shown in Fig. 3. If any change occurs to the hearing characteristics of the user, the audiologist records the new audiogram. The bandwidth of each of the frequency bands is altered within the range b_{set} for all the filters. Also, proper gain can be provided to the filters by the audiologist.

This forms an approximation model of the audiogram and can be altered during simulation until a minimum matching error is obtained. Matching error is the overall error between the filter output and the audiogram [7]. The advantage of the proposed method is that, the hardware overhead in realizing the non-uniform frequency bands is minimal and depends on



(a) Frequency Shifting

(b) Separate gain to each band

Fig. 3 The variable bandwidth filter is shifted and provided gain to match an audiogram of mild hearing loss at all frequencies.

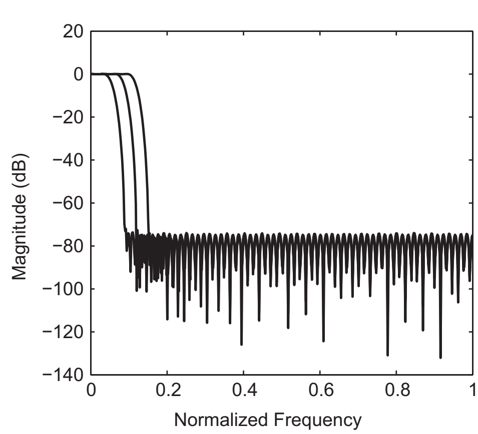


Fig. 2 Farrow structure based VBW for adjustable factor $b - b_0$.

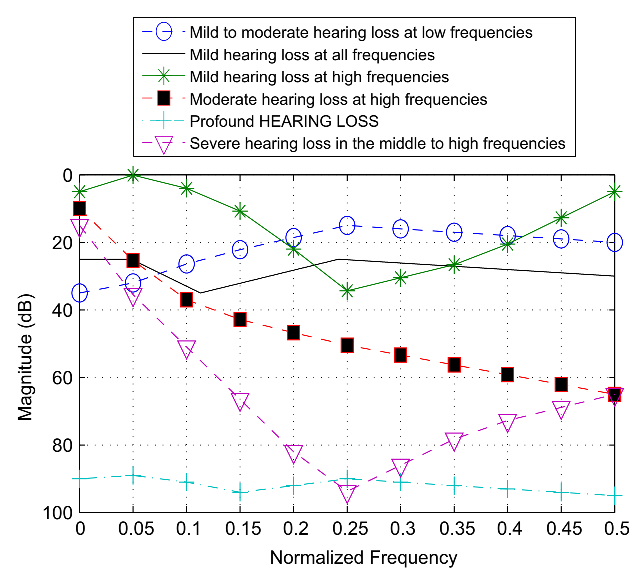


Fig. 4 Sample audiograms used for verification.

Table 1 VBW parameters with 10, 8, 6 and 4-band hearing aid for audiograms in Fig. 4.

No. of bands	Bandwidths (Hz)	Transition width (Hz)
10	500, 750, 1000, 2430, 3100	311.1
8	800, 1000, 1500, 1900, 2500, 3160	311.1
6	800, 1000, 1500, 1700, 2000, 2400, 2800, 3700	339.4
4	1000, 2000, 3000, 5000	622.2

Table 2 Comparison of minimum matching errors with 10, 8, 6 and 4-band hearing aid for various audiograms.

Sl. no.	Type of hearing loss	Number of bands and maximum matching error in dB			
		10	8	6	4
1	Mild to moderate hearing loss at low frequencies	1.48	1.35	1.87	2.05
2	Mild hearing loss at all frequencies	1.24	1.27	1.6	2.00
3	Mild hearing loss at high frequencies	1.86	2.00	2.49	2.81
4	Moderate hearing loss at high frequencies	1.76	2.57	2.62	3.70
5	Profound hearing loss	2.52	2.51	2.88	2.90
6	Severe hearing loss in the mid to high frequencies	2.44	2.9	3.00	4.3

the number of unique bands required. The number of unique bands required to match a particular audiogram, is found by a number of trials to fit it with minimum number of bands and minimum matching error.

Results and discussion

The aforementioned design is used to obtain audiogram matching on various types of hearing losses. Sample audiograms that are used here are adopted from the Independent Hearing Aid Information [1,19], a public service by Hearing Alliance of America. These are as given in Fig. 4. Using the proposed method, the audiogram fitting is tried for 4, 6, 8, and 10 bands on the sample audiograms. The matching error comparison is made in Table 2.

Design example

A bank of digital filters are to be designed to match each of the audiograms of Fig. 4. Optimal sub-band bandwidths for matching these audiograms are decided by first simulating them individually for minimum matching error. For the example in Fig. 3, minimum number of bands for best matching for the audiogram with mild hearing loss at all frequencies, is obtained by trial and error approach, and is found as 7. For the design Example 4.1, a trial is carried out to find the minimum number of bands, among 4, 6, 8, 10 bands, to obtain minimum matching error with respect to all the 6 audiograms in Fig. 4. The comparison is provided in Table 2. Consider 8-bands of filters to be used, each having a maximum deviation in passband and stopband respectively as follows,

$$\delta_c = 0.0058$$
$$\delta_s = 0.00056$$

The optimum transition bandwidth for this example is obtained, by trial and error for the chosen set of audiograms, as 311.1 Hz. A set of 8 different bandwidths is to be obtained using the variable bandwidth filter, as described in Section "Results and discussion" and shown in Fig. 2. This is realized using the proposed method, where the variable bandwidth filter is a linear phase Type I low pass filter with varying bandedges.

The method is then repeated for realizing the bank of filters whose response is divided as 10, 6 and 4 bands. The bandwidths and the transition bandwidth for the VBW filter, to match these audiograms, for 10, 8, 6 and 4 bands realization are as given in Table 1.

Table 3 Comparison for various audiograms in terms of Hardware complexity and Matching Error.

Hearing loss type	Method in James and Elias [1]				Proposed method			
	No.of bands	Max. error	Multipliers (1 band)	Adders (1 band)	No.of bands	Max. error	Multipliers (1 band)	Adders (1 band)
Mild to moderate hearing loss at low frequencies	10	1.78	445	889	8	1.35	138	275
Mild hearing loss at all frequencies	10	1.93	445	889	10	1.24	160	319
Mild hearing loss at high frequencies	10	3.54	445	889	10	1.86	160	319
Moderate hearing loss at high frequencies	10	3.05	445	889	10	1.76	160	319
Profound hearing loss	6	2.49	445	889	8	2.51	138	275
Severe hearing loss to high frequencies	10	5.53	445	889	10	2.44	160	319

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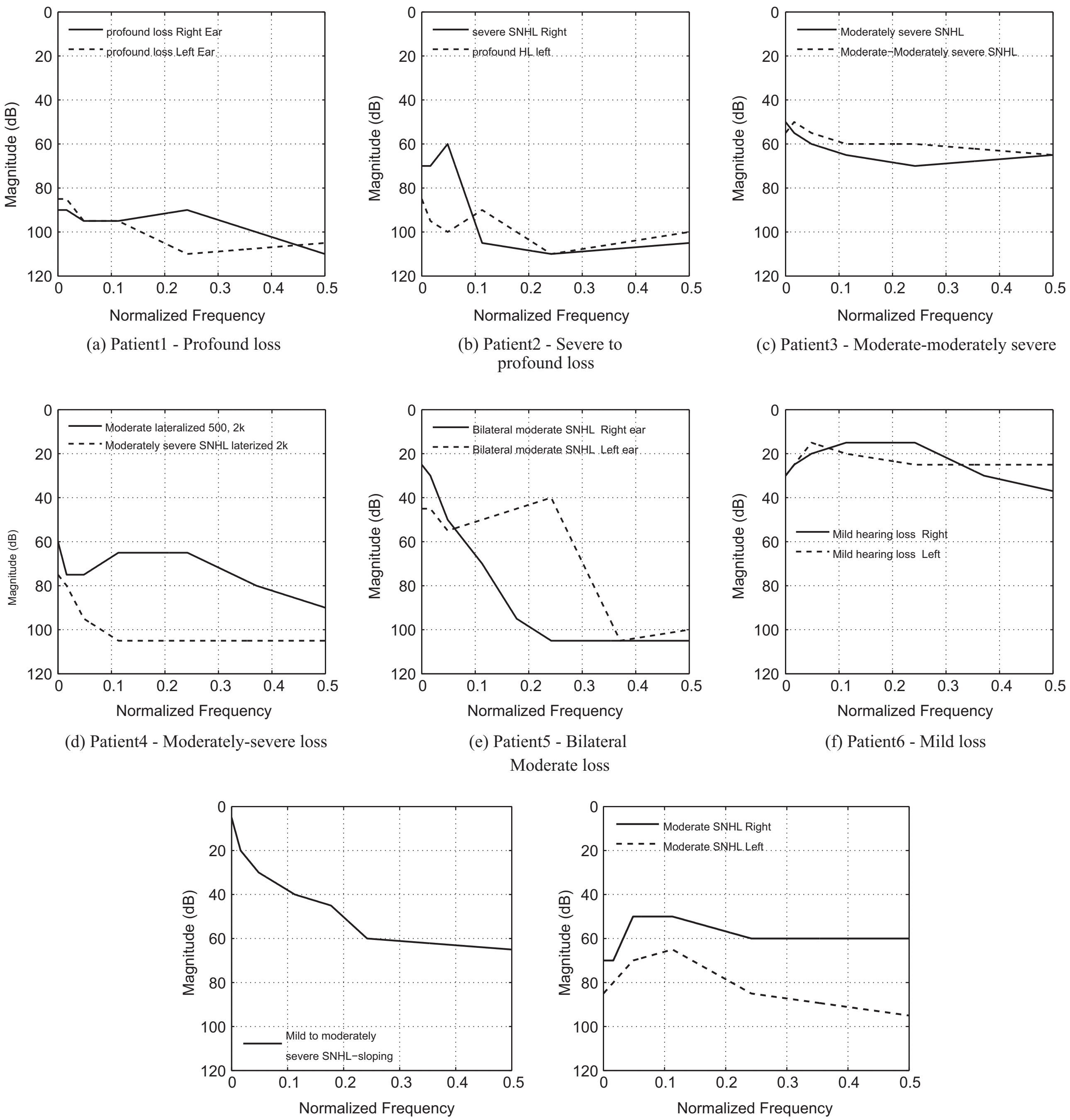

(a) Patient1 - Profound loss
(b) Patient2 - Severe to profound loss
(c) Patient3 - Moderate-moderately severe
(d) Patient4 - Moderately-severe loss
(e) Patient5 - Bilateral Moderate loss
(f) Patient6 - Mild loss
(g) Patient7 - Mid-to-moderate loss
(h) Patient8 - Moderate Sensorineural loss

Fig. 5 Audiograms collected from Government Medical College, Kottayam.

Matching errors for the selected set of audiograms, when matched using 4, 6, 8 and 10 bands of filters, are given in Table 2.

Hardware complexity

A digital hearing aid is to be compact and thus the amount of hardware that goes into its design is to be kept minimum. In the current scenario, we aim to minimize the number of multipliers in the filter design, which contributes toward area and power during implementation [20]. Selection of optimal number of bands and minimum order VBW filter contributes to the overall lowering of hardware complexity. Also, the Farrow based structure is mainly used for providing enhanced tunability. A comparison of the proposed method with the method by James and Elias [1] is done in Table 3. From Table 2, minimum number of bands giving minimum matching error for every audiogram is compared with the corresponding minimum error by following the method given by James and Elias [1]. The parameters of comparison have been chosen as the number of multipliers and adders for a single filter. For all the cases except that for profound hearing loss, the

Table 4 VBW parameters of 8-band hearing aid for audiograms in Fig. 5.

No. of bands	Bandwidths (*Hz*)	Transition (*Hz*)	No. of multipliers	No. of width adders
8	600, 1000, 1500, 2500	186.66	180	359

Table 5 Selected minimum matching errors for low hardware complexity for real patient data.

Patient no.	Sl. no.	Diagnosis	Maximum matching error
1	1	Profound loss Right Ear	1.99
1	2	Profound loss Left Ear	1.82
2	3	Severe sensorineural HL Right	1.96
2	4	Profound HL left	2.06
3	5	Moderately severe SNHL	1.71
3	6	Moderate to Moderately severe SNHL	1.68
4	7	Moderate lateralized 500, 2 k	1.93
4	8	Moderately severe, laterized at 2 k	1.95
5	9	Bilateral moderate SNHL Right ear	2.38
5	10	Bilateral moderate SNHL Left ear	3.05
6	11	Mild hearing loss Right	2.27
6	12	Mild hearing loss Left	1.61
7	13	Mild to moderately severe with high frequency sloping	2.39
8	14	Moderate SNHL Right	1.58
8	15	Moderate SNHL Left	2.51

proposed technique gives better matching error than those obtained using method by James and Elias [1]. For profound hearing loss, the existing method [1] and the proposed method give almost the same matching error. The former requires only 6 bands, but with 445 multipliers for each filter. Our proposed technique requires 8 bands, but with only 138 multipliers for each filter. Hence, there is a significant advantage in the number of multipliers and adders when the proposed technique is employed.

Also, in some cases, minimum number of bands is sufficient, as in rows 1 and 2 of Table 2, when the proposed method is used. For mild hearing loss at high frequencies (row 3), the matching error is as high as 3.54 dB by following the method in a paper by James and Elias [1], for 10 sub-bands and more than 10 dB obtained in the paper by Lian and Wei [4] for 8 sub-bands. This is brought down to a maximum of 2.8 dB with only 4 bands and a minimum of 1.8 dB with 10 bands, using the proposed design. The number of multipliers required to implement a single filter is 138, when designed to fit the audiogram with 8 bands. When the same is performed for 10 bands, the number of multipliers for each filter is 160, for almost the same matching error. The designer can trade-off between number of bands and the filter order.

Design for real world audiograms

The proposed method is also applied to real data of some patients.

Data collection

The data are collected from the Government Medical College, Kottayam, India, with the clearance from its ethical committee (IRB No. 35/2014). All procedures followed were in accordance with the ethical standards of the responsible committee on human experimentation (institutional and national). Informed consent was obtained from all patients for being included in the study.

These audiograms are shown in Fig. 5 and classified by the audiologist as mild, moderate, moderately severe, severe, profound sensorineural hearing losses (SNHL). The number of bands used to fit the real set of audiograms is chosen as 8. This selection is also made by individually simulating the audiograms for 4–10 bands, as done in the previous example. The parameters for the VBW filter design are given in Table 4. This filter is realized for the required bandwidth and center frequency, for the 8 bands, separately for each of the audiogram. The matching errors obtained are provided in Table 5 along with the hardware complexity for single subband implementation. It can be observed that the design is optimized in such a manner that, the maximum matching error does not exceed 3 dB for any of the data considered. The right ear audiogram for Patient 2 in Fig. 5(b) has comparatively larger slope. Still, a matching error of 1.96 dB is possible. In the case where there are laterized sections such as in Fig. 5(d), which has even slope from 2 k*Hz* to 8 k*Hz*, was matched within 1.95 dB. Also, note that the number of multipliers in this case is only 180 for this set of real audiograms. This is due to the optimal transition width used for the filter design. As mentioned in Section ''Results and discussion'', the selection of transition width according to the requirement is possible with this technique and this gives an amount of flexibility to the designer. Thus, it can be seen to have a large amount of saving in terms of hardware.

Conclusions

An efficient method for the design of digital filters suitable for digital hearing aid, is proposed in this paper. The method utilizes Farrow structure based variable bandwidth filters. The required variable bandwidth response is obtained by using a single parameter, b. A fixed number of bands are generated from the variable bandwidth filter by means of spectral shifting of the required bandwidth response. The difference in the overall response from the corresponding audiogram gives the matching error. This method is applied to a set of standard database audiograms as well as on some real hearing loss data of patients. Thus, the vendors of hearing aid can design an instrument to suit a set of hearing loss patterns, that can be later customized for any user by means of the parameter b and simple frequency shifting. These adjustments are made for each user by the audiologist. Compared to a previous sample rate conversion based method [1], this technique proves to give better audiogram matching with minimum hardware implementation complexity (mainly multipliers). The variable bandwidth based design is simple as only the shifts and required gain are to be provided. Since separate filters are used for subband selection, there is no additional delay incurred, which is a required characteristic of a good hearing aid. The proposed method uses trial and error approach to decide the minimum number of bands, their center frequencies and magnitude gain such that the matching error is minimum. But for a set of audiograms, the hearing aid is designed in such a way that the variable bandwidth filter coefficients remain fixed. The same set of filters are placed at each band with the required bandwidth at that center frequency. Thus, for all the types of hearing losses considered, the design of variable bandwidth filter using Farrow structure is a one-time job. Once it is designed, it can be reconfigured for each user, by the audiologist, for one of the type of hearing loss considered. Magnitude gain change can simply be adjusted even after the design.

Conflict of interest

The authors have declared no conflict of interest.

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