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Enhanced Hearing Aid Performance with an African Buffalo Optimization-Based Frequency Response Masking Filter

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ABSTRACT

Hearing aid systems rely on reconfigurable filters to selectively amplify desired signals while suppressing unwanted noise. However, the complexity of the hardware and the associated power dissipation present substantial challenges, particularly in systems where adaptability and scalability are crucial. These systems require advanced auditory compensation to accommodate the diverse auditory profiles of individuals with hearing impairments. In this study, an African buffalo-inspired optimization algorithm is introduced to fine-tune a Frequency Response Masking Reconfigurable Filter (AB-FRMRF). This novel approach ensures High-Q filtering with narrow transition bands while maintaining a low filter order, thereby enhancing selectivity and minimizing complexity. The optimization algorithm adjusts its search strategy based on solution fitness, optimizing the filter's coefficient values. Performance parameters such as matching error and group delay are meticulously tuned for optimal auditory compensation. Utilizing distributed arithmetic for hardware realization and a Brent Kung adder for the summation of dynamic random-access memory (DRAM) partial products, the hardware design deviates from conventional pipeline adder trees, culminating in significant power savings and improved processing speed. Modeled in MATLAB and implemented on a Kintex 7 FPGA Genesys 2 board, the AB-FRMRF model exhibits a reduced matching error down to 1.2 dB and a minimized delay to 2.5 msec for high-frequency sounds in cases of mild hearing loss. These metrics are favorable in comparison to existing reconfigurable filters reported in the literature. The AB-FRMRF model not only demonstrates a 35% improvement in power consumption but also outperforms traditional methods in computational complexity, auditory compensation, delay, and power metrics, making it highly suitable for hearing aid applications. The proposed filter design signifies a substantial advancement in auditory assistance technology, aligning with the necessity for high-performance, low-power hearing aid devices.

1. INTRODUCTION

Hearing is one of the most sophisticated and delicate functions of the human body, with abnormalities in the auditory system's nerves leading to hearing loss [1]. Sensorineural hearing loss (SNHL), often resulting from aging [2], infections, ototoxic medications, and exposure to excessive noise [3], necessitates the use of hearing aids to improve auditory perception. The severity of SNHL is frequently linked to the degree of deviation from a fixed threshold across frequencies [4]. Contemporary hearing aids employ dynamic multiband compression schemes, which separate the input signal into multiple bands [5]. This multiband approach was developed to address the limitations inherent in single-band compression systems [6]. Nevertheless, such a multiband compression module often results in reduced modulation and spectral efficiency in communication signals [7]. The advent of advanced digital signal processing (DSP) techniques [8], including noise reduction, band coding, echo cancellation [9], speech enhancement, and auditory repair strategies, are now repeatedly applied across various platforms [10]. Digital hearing aids utilize an array of DSP techniques to ensure speech clarity in the absence of noise [11], despite these innovations, modern hearing aids continue to encounter design challenges related to achieving sharp transition bandwidth, enhanced speech intelligibility, reduced chip size, and low power consumption.

In hearing aids, the filter bank is a pivotal component, with reconfigurability signifying the ability to tailor filters to the unique auditory profiles and environments of users with hearing impairments. Yet, existing filter designs are not without their shortcomings. They are often complex, resulting in excessive power usage, and the rate of adjustment to frequency and gain changes can be restrictive. Moreover, the compact nature of hearing aids poses additional challenges in accommodating reconfigurable filters within such limited spaces, affecting sound quality, user comfort, device aesthetics, and battery life. Auditory repair is especially critical for those suffering from a reduced range of hearing and audibility [12].

Recent developments in hearing aid technology have introduced filter bank models with a limited number of bands, coupled with active compression to mitigate hearing loss [13].

The signal processing involved is primarily geared towards noise reduction and is integral to hearing aid systems. Precise audio environment characterization can greatly enhance the design of automatic programs within hearing aid devices [14], necessitating advancements to minimize noise interference [15]. The literature reveals various design approaches for reconfigurable filter banks tailored to hearing aid applications [16], with Field Programmable Gate Arrays (FPGAs) frequently employed in their design. Initially, the speech signal is processed through an A/D converter, segmented into multiple subbands by the filter bank, each with its gain factor, before being reconverted by a D/A converter. A fixed bandwidth is commonly used in hearing aids to compensate for hearing loss.

However, an exhaustive literature review has highlighted several potential improvements for filters used in hearing aid design, which served as the impetus for this research. Increasing filter order typically leads to a rise in power consumption and complexity. Although various digital approaches, such as 2-level audio decompression [17], mono reconfigurable structures [18], and non-uniform filter band models [19, 20], have been implemented to improve hearing aid performance, they often encounter issues related to power usage and latency. Furthermore, there is a recognized need to minimize matching errors to provide better auditory compensation, a critical performance metric. This paper proposes an optimized filter bank design to ameliorate delay and power consumption while enhancing the overall performance of hearing aids.

The key contributions of this research are as follows:

- A comparative analysis of the key performance parameters of hearing aids against different reconfigurable filter-bank designs.
- The design and optimization of a Frequency Response Masking (FRM) filter using MATLAB, employing the African Buffalo optimization algorithm.
- The adoption of a distributed arithmetic approach for optimal hardware realization, and the introduction of the Brent Kung adder for adding DRAM partial products, as opposed to the traditional pipeline adder tree, with implementation on an FPGA.
- A comparative analysis of optimized performance parameters such as matching error, group delay, and power consumption against existing designs.

The organization of this paper is as follows: Section 2 reviews related work on hearing aid systems. The system model and problem statement are detailed in Section 3. Section 4 presents the proposed methodology in depth, with results discussed in Section 5. The paper concludes in Section 6.

2. LITERATURE SURVEY

Hearing aid devices primarily serve two key functions: masking and generating passband frames. Fine-tuning these parameters has led to significant improvements in system performance. Amir et al. [17] introduced a 2-level Audio Wave Decompression (AWD) framework within a reconfigurable device to enhance functionality and auditory performance. This design was hardware-validated and succeeded in reducing hardware complexity, although it exhibited a considerable range of matching errors. Amir et al. [18] developed a mono-reconfigurable structure catering to various hearing impairments across patients, with FPGA-based implementation results indicating low power consumption and reduced complexity. The filter parameters were refined using an optimization model.

Shrivastav and Kolte [19] proposed a reconfigurable Frequency Response Masking (FRM) filter bank, optimized with a heuristic technique, yielding over a 60% improvement in auditory compensation. Wei et al. [20] introduced a non-uniform filter bank (NF) model with 16 bands for hearing aid systems, requiring only three multipliers for executing filter band operations. This model demonstrated a lower delay ratio, though it was complex in design. Shrivastav et al. [21] designed a reconfigurable FRM filter utilizing three decomposition levels and optimized it with a swarm-based algorithm. Fan et al. [22] created a deep learning-based application for hearing aids capable of automatically recognizing five types of audio effects, achieving a 98.8% accuracy rate with thirty audio blocks for training, albeit with a longer execution time.

Zhang et al. [23] designed a unified modulated filter bank aiming to reduce complexity by minimizing the use of multipliers, but this approach resulted in lower hardware accuracy. Tiwari et al. [24] presented a sliding-band compression model for sensorineural hearing loss, offering a distinct amplitude range for various compression ratios but with high processing times that affected performance. Zheng et al. [25] introduced an FRM-based non-uniform filter bank for hearing aids, achieving desired stopband attenuation with optimal word length for filter coefficients, which simplified hardware design. Bhuyan et al. [26] proposed an area- and power-efficient reconfigurable filter using interpolated subband distribution and a LUT-less DA method. Devis and Manuel [27] developed a cost-effective three-level filter bank from a single FIR filter using the Parks McClellan algorithm and fractional interpolation, aiming for minimal hardware requirements. Lastly, Parameshappa and Jayadevapp [28] proposed an FRM-based uniform digital filter bank utilizing an interpolated half-band FIR filter, which achieved an acceptable range of matching error and computational complexity.

This extensive review of existing reconfigurable filter designs underscores a significant opportunity for enhancements in matching error, delay, power consumption, and hardware complexity. The primary aim of the proposed research is to address these gaps and provide hearing aid users with a more effective and user-friendly experience. The proposed approach begins with the input of the original speech analog signal, followed by the development of a novel African buffalo-based Frequency Response Masking Reconfigurable Filter (AB-FRMRF) using MATLAB and the Xilinx tool. The filter bank's characteristics, such as matching error, delay, frequency response, area, and power consumption, are meticulously optimized using the African buffalo optimization algorithm. The proposed model is then thoroughly evaluated against existing literature on parameters including delay, matching error, area, and power, ensuring a comprehensive analysis of its efficacy.

3. SYSTEM MODEL AND PROBLEM STATEMENT

Hearing aid (HA) is one of the necessities for people who are hard of hearing to live an everyday life. So, several models are implemented on diverse software platforms to enhance hearing aid functions. However, these models are complex in hardware architecture because of their complicated and broad design. Moreover, numerous existing approaches have consumed much power and attained high delay measures. Statistics from the World Health Organisation show an enormous gap between the number of people who need hearing aids and those who are using them. Better audiogram matching with reduced power requirements, narrow transition bandwidth, and less complexity are the primary concerns in design to prevent hard-of-hearing patients from using hearing devices. Addressing these burning issues that degrade the system's performance is the primary goal of this research work. The FRM technique provides a narrow transition bandwidth and does not increase filter order as the size increases; hence, it is chosen for the proposed filter design. The basic block diagram of the FRM filter is detailed in Figure 1 [29].

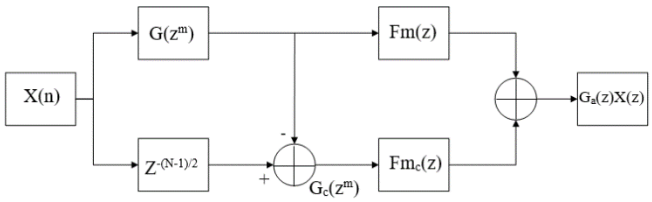

Figure 1. System model for FRM filter

The FRM sub-filter's transfer function is mentioned in Eq. (1).

$$\begin{aligned} T_a(z) &= \sum_{l=0}^{N-1} t_l z^{-l} \\ T_{ma}(z) &= \sum_{l=0}^{N_a-1} t_l^a z^{-l} \\ T_{mc}(z) &= \sum_{l=0}^{N_c-1} t_l^c z^{-l} \end{aligned} \quad (1)$$

The group delay is mentioned in Eq. (2), where the subfilters of the FRM have linear phase response through N_a and N_c.

$$D = \left[\frac{N-1}{2}\right] m + \max\left\{\left[\frac{N_a-1}{2}\right], \left[\frac{N_C-1}{2}\right]\right\} \quad (2)$$

The low-pass filter (LPF) has the sampling factor (S), stopband edge as ω_a and passband edge as ω_p. The utilized subfilters are mentioned as $T_a(z)$, $T_{ma}(z)$ and $T_{mc}(z)$ respectively. Additionally, the LPF design with the edges of bands has determined for $T_a(z)$, the passband and stopband edges as θ, ϕ are mentioned in Eq. (3) and Eq. (4).

$$\theta = \omega_p m - 2m\pi \quad (3)$$

$$\phi = \omega_a m - 2m\pi \quad (4)$$

where, $m = \omega_p m \,/\, 2\pi$.

The output of the FRM filter is mentioned in Eq. (5).

$$Ga(Z) = G(Zm)Fm(Z) + Gc(Z)Fmc(Z) \quad (5)$$

The designed FRM filter bank is then enhanced using the proposed reconfigurable model and the African buffalo mechanism, which lessens the matching error and latency.

4. PROPOSED AB-FRMRF DESIGN METHODOLOGY

Typically, digital hearing aids are effective for people with hearing loss. However, the existing hearing aids have uncomfortable noise, high delays and more matching errors. The proposed study article aims to model a novel African buffalo-based Frequency Response Masking Reconfigurable Filter design (AB-FRMRF) to enhance the functionality of hearing aid devices to improve the hearing ability of hard-of-hearing people. Additionally, the African buffalo fitness function optimizes the parameters of the designed filter bank. The proposed filter bank's masking parameter also covers the full-range frequency signal. Figure 2 describes in detail how the proposed AB-FRMRF system operates.

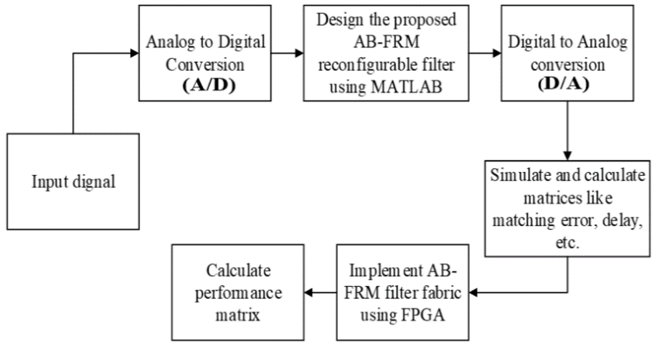

Figure 2. Methodology for AB-FRMRF design

The A/D converter is given the input audio signal. The proposed filter then processes the input to increase the effectiveness of the hearing device. The D/A is then fed the filter bank's output, and the parameters are determined. As a result, the designed approach has produced the best results while using less power and experiencing less delay.

4.1 Design of FRM reconfigurable filter model

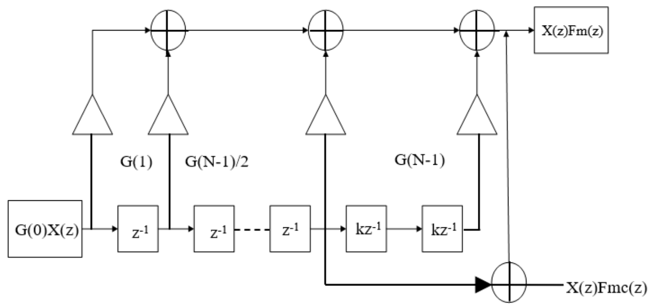

Figure 3. Proposed AB-FRMRF structure

The proposed AB-FRMRF model is developed to augment hearing aid performance. The proposed optimized design has decreased complexity and delay while achieving minimal matching error. This work builds the AB-FRMRF filter and optimizes the filter bank parameters using the African Buffalo Optimisation (ABO) [30] methodology. The FRMRF filter is hybridized with ABO to improve the filter bank characteristics. The structure of AB-FRMRF filters is paired with masking filters, the AB fitness function, and periodic model filters for achieving a narrow transition band using FIR filters with linear phase response. In this method, the proposed filter divides the entire frequency range into separate zones with variable bandwidth, utilizing the masking stage. The masking filters are represented as Fm (z) and Fmc (z). Also, the periodic model filters Ga(z) and $G_a(z^m)$ are designed by exchanging every delay component of the narrow band filter into (m) delay components by fitness (k), which are depicted in Figure 3.

Additionally, passband and stopband edges in the filter are mentioned as ϕ and θ. The matching Filter's transition width $G(z^m)$ is $(\theta - \phi)/m$. Also, these filters are mentioned as $G_c(z^m)$ and expressed in Eq. (6) as:

$$G_c(z^m) = \left(z^{\frac{-N-1}{2}k} - G(z^m)\right) \quad (6)$$

where, N denotes the impulse response length of $G(z^m)$, subsequently, the masking filters of this structure have cascaded to $G(z^m)$ and $G_c(z^m)$ and are combined to create an FRM filter. As a result, Eq. (5) is used to represent the developed AB-FRMRF filter's transfer function.

The proposed AB-FRMRF filter's frequency response is mentioned in Eq. (7).

$$G_a(z) = \mathrm{G(Zm)Fm(Z)} + \left(z^{\frac{-N-1}{2}k} - G(z^m)\right)\mathrm{Fmc(Z)} \quad (7)$$

The proposed AB-FRMRF filter also has group delay, as mentioned in Eq. (8).

$$D = \frac{(N-1)m}{2} + d \quad (8)$$

where, d = max($(N_a$-1)/2, $(N_c$-1)/2).

4.2 Methodology for optimized AB-FRMRF model design

The proposed AB-FRMRF model's methodology is described in Figure 4.

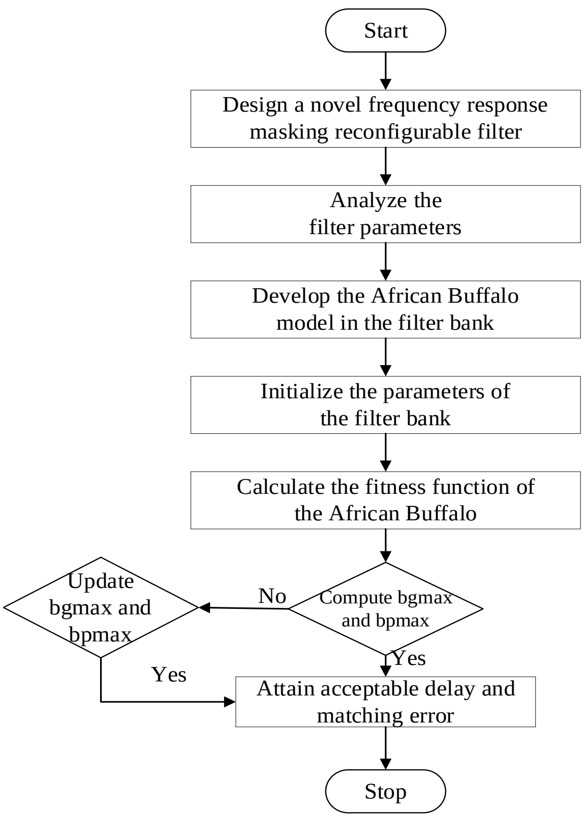

Figure 4. Flow chart of proposed AB-FRMRF algorithm

Here, the constructed filter bank's parameters, such as delay and matching error, are further tuned using the AB fitness function indicated in Eq. (9)

$$k = m.a + r_1(bg_{max} - w.a) + r_2(bp_{max} - w.a) \quad (9)$$

where, r_1 and r_2 are the learning parameters for identifying the filter bank parameters, whereas m.a and w.a denote the actual loss parameters, bg_{max} represents the matching error optimizing factor, and bp_{max} is the optimizing factor for group delay. As a result, outcomes like matching errors and delays were more effective. More sub-bands are generated because of the reconfigurable nature of the proposed filter-bank model.

4.3 Speech perception analysis

Speech perception of a person using a hearing aid is the ability to read, interpret, and understand language sounds. The input signal from the speaker includes several environmental effects like background noise, reverberation, and signal level reduction due to the distance between the speaker and the hearer. Speech perception consists of the hearer's outcome of finding an acceptable fit between the emotional input provided by the speaker and the linguistic structure intended by the speaker.

The suggested filter bank evaluates each patient's hearing loss and speech perception. Moreover, to perform speech perception analysis, various categories of input signals are given to the filter bank to identify the ability to understand the sound signals.

5. AUDIOGRAM MATCHING

The designed AB-FRMRF filter bank in the hearing aid is verified in this study using audiogram matching concerning the hearing loss pattern. The results are simulated and estimated. The proposed design uses the bands developed at middle frequencies with fitness functions to optimize the matching error.

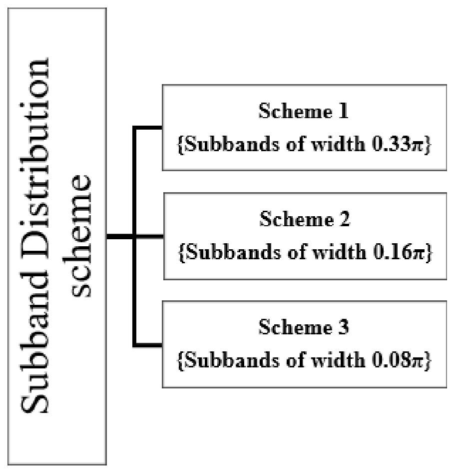

Figure 5. Proposed sub-band distribution scheme

As elaborated in the methodology section, the proposed FRM filter-bank decomposes the complete frequency range from 250 Hz to 8K into multiple subbands using the subband allocation scheme defined in Figure 5. Here, three schemes are generated for allocating the subband: schemes 1, 2, and 3.

Scheme 1 contains a subband in the bandwidth range of π/3; the subband in Scheme 2 was allocated in the bandwidth range of π/6, and the subband in Scheme 3 was given in the bandwidth range of π/12. A set of sample audiograms of patients with varying hearing losses in different frequency bands is taken as a data set. The system is trained accordingly to provide desirable amplification in the frequency subbands not audible to patients. Thus, better audiogram matching is verified from experimental results using the proposed AB-FRMRF filter bank. Hereafter, the FPGA models of the proposed filters are designed, and their performance is validated with different input signals.

The hardware architecture of the proposed reconfigurable filter is designed by employing the Distributed Arithmetic (DA) technique. The proposed architecture comprises processing elements (PE) and pipeline shift add trees, as shown in Figure 6.

The processing element comprises a SIPO Shift Register, a set of DRAM-based generators of partial products (DRPG), a parallel prefix adder, and a shift accumulator, as shown in Figure 7. In literature, the Pipeline Adder Tree [PAT] adds partial products. PAT can reduce the delay associated with adding large operands by breaking them into smaller chunks that can be added in parallel. However, this comes at the expense of increased hardware complexity due to the need for multiple adders and additional control logic. We used Brent-Kung Adder, a parallel prefix adder with high performance and low hardware complexity, in our proposed design. The Brent-Kung Adder has a smaller depth than the PAT, further reducing propagation delay. Also, it is faster than the pipeline adder tree for large operands. Thus, in the present research work, adding Brent Kung's adder has improved results by utilizing less area and less delay time. Also, using a shift accumulator in the FPGA module improves communication sharing by maximizing the throughput ratio.

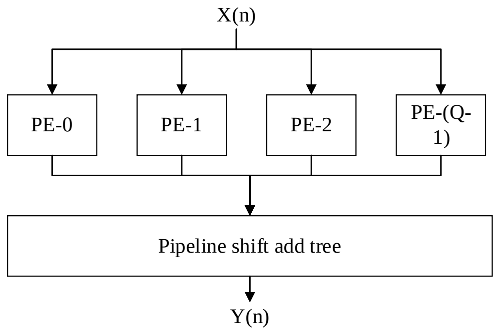

Figure 6. Hardware architecture of the proposed filter

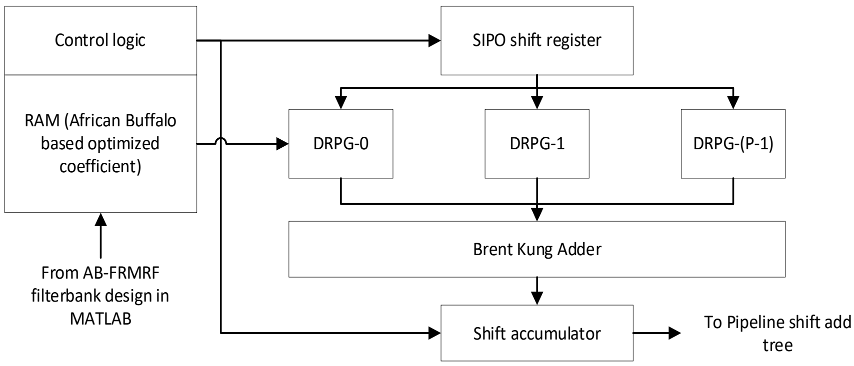

Figure 7. Detailed structure of processing element (PE)

6. RESULTS AND DISCUSSION

Table 1. Different audio signal dataset

Sr. No	Audio Signal	Samples	Frequency (Hz[1])
1	Speech 1	256733	44090
2	Speech 2	41874	26453
3	Frogs	12254	14000
4	Birds	145671	14000
5	Ducks	74568	14000
6	Traffic	123478	14000
7	Piano	154567	14000
8	Speaker in restaurant	145983	14000
9	Beethoven music	129875	14000
10	Clarinet	113234	14000

Note: [1]Hertz

A novel filter bank is presented in this paper to augment the performance of the HA system. The Xilinx/Kintex and MATLAB frameworks are used to develop the proposed AB-FRMRF replica. The FPGA device model is XC7K325T-2FFG900C, and parameters, including matching error, delay, frequency, area, and power, are determined. Additionally, the obtained findings are contrasted with other widely used techniques for performance parameters, viz., matching error, frequency, delay, power, and area. Table 1 displays the various audio signal datasets that are used.

Initially, the proposed filter is designed mathematically in the MATLAB environment. The filter's performance was analyzed by evaluating key metrics like matching error and delay. Hereafter, the speech signal is converted to digital form and trained as input for the Xilinx FPGA model. Here, the novelty in the FPGA module is Brent Kung's adder. It used less area to run the process. The RTL schematic model of the proposed system designed for Vivado 2018 is depicted in Figure 8. After FPGA synthesis, the filter coefficients are again imported into MATLAB to optimize the filter coefficients. The optimized filter coefficients are then given to Xilinx, and the optimized results are validated. The proposed filter reports a delay of 4.443 ns in the FPGA module against the typical reconfigurable filter delay, which was high up to 6.008 ns in the literature.

The Simulation results of the proposed filter design are shown in Figure 9.

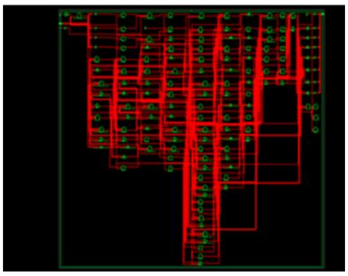

Figure 8. RTL schematic model of the proposed filter

Figure 9. Simulation result of the proposed design

6.1 Validation

Hearing loss can cause significant harm to human life, relationships, work, and physical and emotional well-being, making it one of the most critical health problems affecting human life. In this research, the novel AB-FRMRF filter is designed to enhance the efficiency of hearing aids to improve the hearing ability of hearing loss patients. This scheme includes digital filters, which tune the amplitude based on the patterns of hearing loss (HL). The hearing loss types are also mentioned as usual: mild, moderate, and severe. Moreover, the developed filter separates the sound signals into various subbands. Selective gain has been provided to every subband to match the audiogram of the particular person.

The average hearing range of humans is 0–20 dB. The threshold range of the hearing loss levels is mentioned in Table 2. In this work, the hearing ability of hearing loss patients is improved using a hearing aid system designed using a novel AB-FRMRF model. The audio signals are initially taken for the system's input and are collected under various environmental conditions, like a human voice and music. If the distance between the hearing aid and the source is considerable, the noise and errors in the audio signal are high. Eight sub-bands of input were separated using the specified filter. Specifications of the designed filter bank are shown in Table 3.

Table 2. Threshold range of hearing loss

Degree of Hearing Loss	Threshold Range (dB[1])
Slight	16-25
Mild	20-40
Moderate	41-55
Moderately severe	56-70
Severe	71-90
Profound	>90

Notes: [1]Decibel

Table 3. Specification of the designed filter

Specification	Values
Transition bandwidth	0.05
Passband frequency	6.4KHz[1]
Sampling frequency	16kHz
Passband ripple	0.0001 dB
Stopband reduction	120 dB

Note: [1]KiloHertz

Only thirteen multipliers are used when the developed filter satisfies these requirements. Finally, the hearing aid system employs the designed filter to compute the performance metrics. With the aid of this AB-FRMRF model, the hearing aid system's effectiveness is increased while matching error, frequency, power, and delay are all optimized.

6.2 Performance metrics

The proposed AB-FRMRF approach effectively designs the filter bank and tunes the parameters such as matching error, delay, frequency, power, and area. Initially, the audiogram for various HL categories, such as mild, moderate, mild to severe, and profound HL for left and right ears, is determined.

An audiogram is a graph representing the patient's hearing ability based on various frequencies. A consolidated representation of a set of sample audiograms of varying hearing loss profiles is presented in Figure 10. Additionally, different hearing loss categories are considered for audiogram matching. The subbands are formed during the breakdown of an input signal. Based on the hearing capability of each person, amplification is provided in every subband using the gain block.

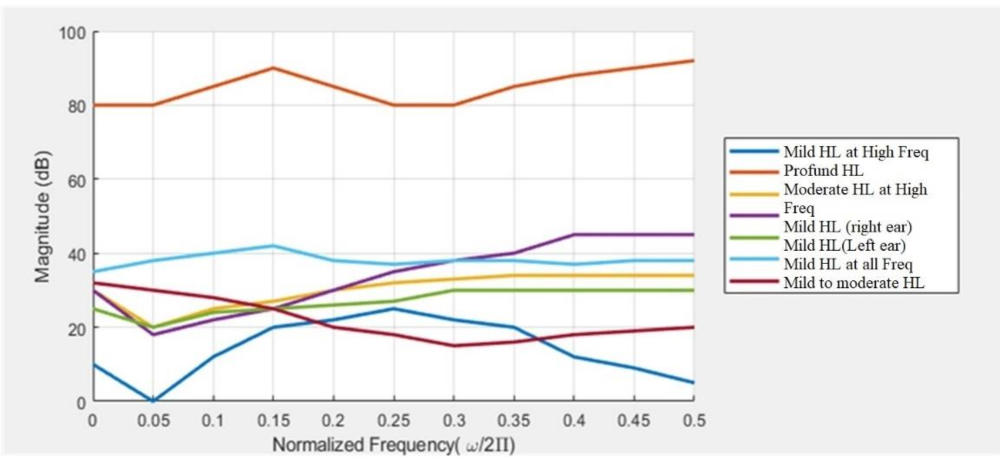

Figure 10. Sample audiogram for various types of HL

The number of bands, frequency range, bandwidth, and gain for each audiogram are analyzed, and these design parameters are given in Tables 4-7, respectively.

Table 4. Design parameters of AB-FRMRF for mild HL

Bands[1]	Frequency (Hz)	Bandwidth (Hz)	Gain (dB)
1	Up to 1000	1500	25
2	800-2200	1500	28
3	2100-3100	1000	30
4	2900-3900	1000	32
5	3700-4700	1000	28
6	4500-6000	1500	30
7	5500-7500	1500	32
8	7000-8200	1500	34

Note: [1]Subbands

Table 5. Design parameters of AB-FRMRF for moderate HL

Bands	Frequency (Hz)[1]	Bandwidth (Hz)	Gain (dB)
1	Up to 800	1000	56
2	500-1500	1000	50
3	1200-2200	1000	58
4	1900-4400	2500	53
5	4200-5200	1000	59
6	5000-6000	1000	60
7	5500-7000	1500	62
8	6500-8100	1500	64

Note: [1]Different frequencies

Table 6. Design parameters of AB-FRMRF for severe HL

Bands	Frequency (Hz)	Bandwidth (Hz)[1]	Gain (dB)
1	Up to 1050	1500	12
2	700-2200	1500	48
3	1900-2900	1000	60
4	2600-3600	1000	78
5	3300-4300	1000	82
6	4100-6100	1900	85
7	5600-7500	1900	70
8	7100-8600	1500	64

Note: [1]Different bandwidth

Table 7. Design parameters of AB-FRMRF for profound HL

Bands	Frequency (Hz)	Bandwidth (Hz)	Gain (dB[1])
1	Up to 1050	1500	85
2	700-2200	1500	88
3	1900-2900	1000	90
4	2600-3600	1000	92
5	3300-4300	1000	82
6	4100-6100	1900	80
7	5600-7500	1500	86
8	7100-8600	1500	88

Note: [1]Decibel

Additionally, the magnitude responses of the utilized filters are represented graphically. The periodic model, masking, and narrowband low-pass filters employed in the proposed FRMRF produce the optimal output signal. Figure 11 to Figure 15 display the obtained audiogram for various hearing impaired with mild to profound hearing loss.

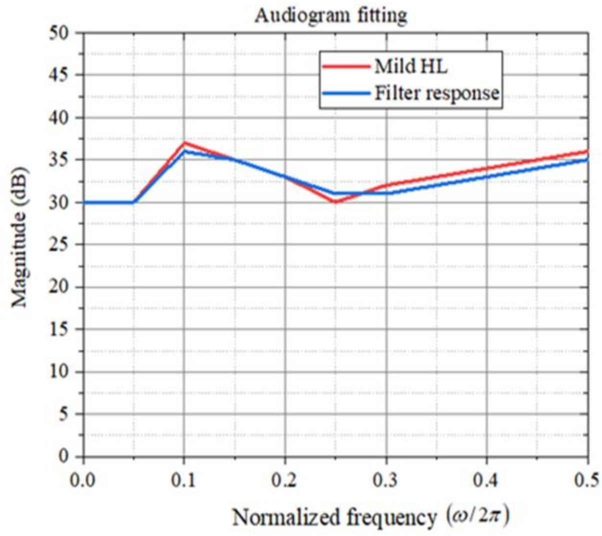

Figure 11. Audiogram fitting for mild HL

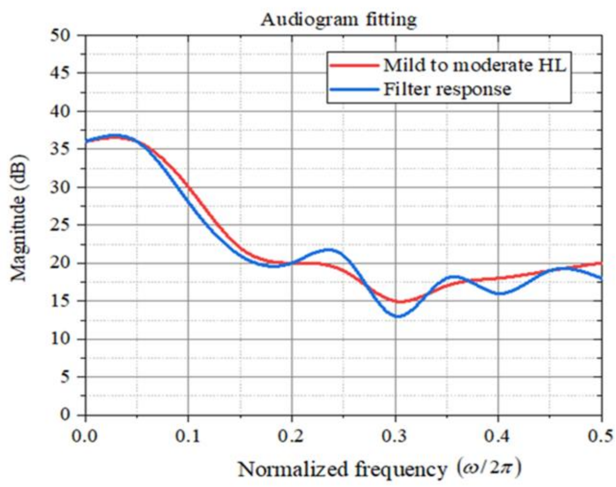

Figure 12. Audiogram fitting for mild to moderate HL

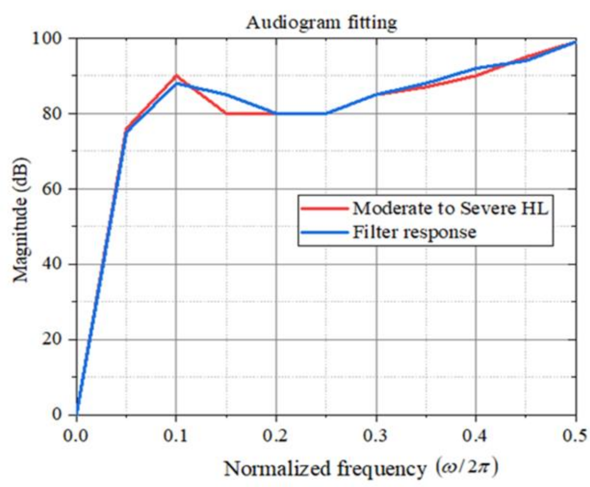

Figure 13. Audiogram fitting for moderate to severe HL

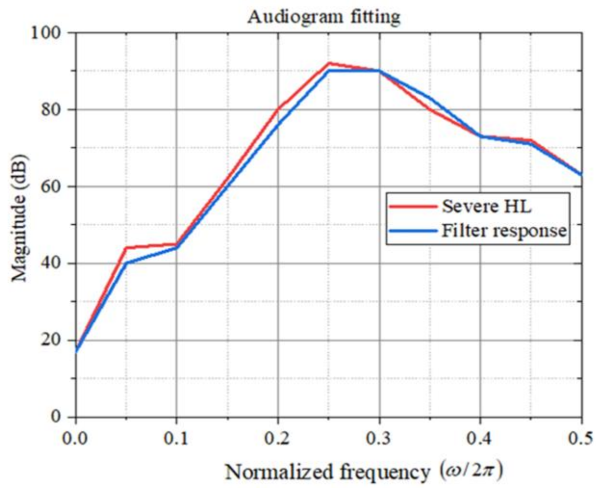

Figure 14. Audiogram fitting for severe HL

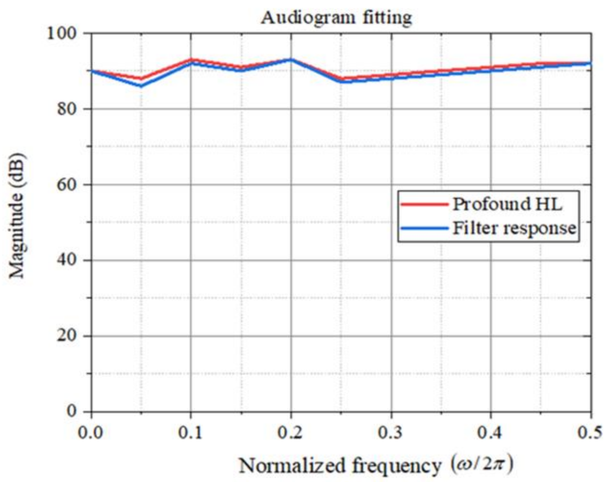

Figure 15. Audiogram fitting for profound HL

6.2.1 Matching error

Matching error (ME) is the measure of variation between the audiogram curves that are matching and the hearing loss curves. The matching error of the proposed approach is compared with existing methods and is shown in Figure 16.

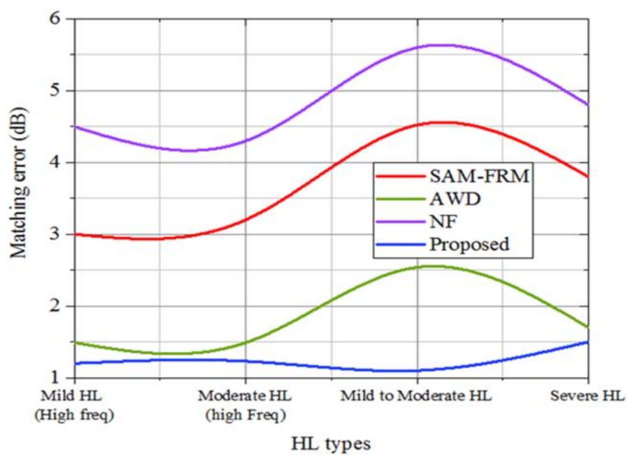

Figure 16. Comparison of ME curves for various methods

The proposed model is compared with existing models like the Signed Approximate Multiplier FRM (SAM-FRM) filter [31], AWD [15], and Non-uniform Filter Bank (NF) approach [18] for evaluating the performance of the proposed model. The comparison of matching errors for different HLs is detailed in Table 8.

Table 8. Comparison of matching error in different methods

Hearing Loss[1]	**Matching Error (dB)**			
	SAM-FRM	**AWD**	**NF**	**AB-FRMRF [Proposed]**
Mild HL (high freq)	3	1.49	4.498	1.2
Moderate HL(high freq)	3.2	1.49	4.3	1.23
Mild to Moderate	4.52	2.54	5.6	1.1
Severe HL	3.8	1.70	4.8	1.5
Profound HL	3.5	1.86	4.3	1.45

Note: [1]Hearing loss types

6.2.2 Delay calculation

The delay calculation is an essential parameter in the HA filter bank. The delay value should be low for a better filter bank system. Generally, non-uniform filter banks have more delay than uniform filters. High delays can affect the lip-reading ability of hearing loss patients. The proposed filter design has attained a lower delay value when compared with other existing methods detailed in Table 9.

The calculation of the delay value of the proposed HA design is compared with existing methods like SAM-FRM, AWD, and NF. Here, the proposed filter model utilized the fitness function of African buffalo to optimize the delay value.

Table 9. Comparison of delay

HL	**Delay (in ms)**			
	SAM-FRM	**AWD**	**NF**	**AB-FRMRF [Proposed]**
Mild HL (high freq)	4.56	13.55	6	2.5
Moderate HL(high freq)	4.8	13.55	6	3.75
Mild to Moderate	5.34	18.61	9.4	4.8
Severe HL	5.24	9.28	7.9	3.79
Profound HL	5.2	15.24	8.75	4.68

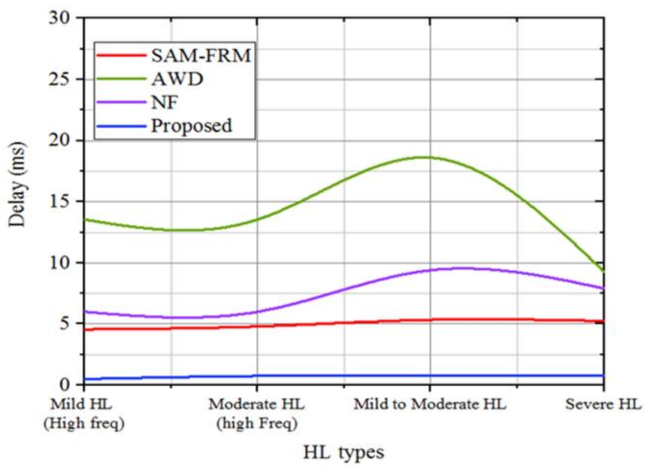

Figure 17. Comparison of delay

The proposed filter bank model has optimized the delay for various hearing loss types at multiple frequencies. The delay value of the proposed approach is compared with the existing techniques and is presented in Figure 17.

In our proposed non-uniform filter design with three sub band distribution schemes, the use of FRM technique further optimized with ABO algorithm in hybrid combination with distributed arithmetic method resulted in variable customized non-uniform sub band distribution with narrow transition bandwidth and optimized filter coefficients leading to improved matching error and delay as against the SAM-FRM [31] and AWD [15] based designs.

Table 10. Comparison of power

Frequency (MHz[1])	Proposed Method Power in µw	Conventional Reconfigurable Filter [32] Power in µw
50	1.300	1.998
100	1.307	2.006
150	1.314	2.014
200	1.321	2.002
250	1.328	2.03

Note: [1]Megahertz

6.2.3 Power analysis

Power is vital for all digital applications; better options exist if the finest technique consumes more power. Power analysis is performed in Xilinx Vivado for the target Kintex 7 FPGA family. The comparison of power analysis is described in Table 10, and its statistical comparison is depicted in Table 11. Results show that the proposed design consumes less power at all frequencies than the conventional reconfigurable filter proposed in the study [32].

In our proposed methodology, using the Distributed Arithmetic technique combined with employing Brent Kung's parallel prefix adder for adding DRAM partial products instead of the conventional pipeline adder tree has significantly improved power requirements. It is thus apt for small-size hearing aids with limited battery life.

Table 11. Statistical comparison of power consumption

Power in µw (Freq Range 50-250 MHz)	Proposed Method	Conventional Reconfigurable Filter
Standard Deviation	0.0111	0.0126
Variance	0.0001	0.0002
Mean	1.3140	2.0100
SEM	0.0049	0.0057

7. CONCLUSIONS

The research presented a novel methodology for developing a reconfigurable filter bank specifically designed for hearing aids. Due to their adaptability and scalability features, high hardware complexity and power consumption are burning issues in existing reconfigurable filter designs. Also, improvement in audiogram matching and reduced latency are a must for better hearing ability and speech perception in the hearing disabled. A novel African buffalo-based frequency response masking reconfigurable Filter (AB-FRMRF) is designed for the HA system to address this issue. Results show that the proposed design is optimized for matching error, delay, and power. Thus, this model has enhanced the performance of the HA system by reducing the variation between potential and actual HA users. The reconfigurable filter has provided flexibility for controlling and altering the sub-band distribution schemes and parameters such as frequency, bandwidth, and location of sub-bands. Therefore, the designed filter banking system has provided excellent results in performance measurements such as matching error, delay, and power in comparison to existing methods. The proposed model enhanced the filter's performance by minimizing the matching error to 1.2 dB and delays to 2.5 ms for mild hearing loss at high frequencies. The proposed hardware implementation leads to a 35% improvement in power consumption compared to conventional models. Thus, the proposed AB-FRMRF model can greatly improve the performance of hearing aids and can give impetus to the use of hearing aids by people who are hard of hearing.

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