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Reconfigurable Multiband Dynamic Range Compression-based FRM Filter for Hearing Aid

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Abstract— In this research, we present an innovative method for enhancing the performance of hearing aids using a Multiband Dynamic Range Compression-based Reconfigurable Frequency Response Masking (FRM) Filterbank. First, a unform16-band reconfigurable filter bank, which is reconfigurable, is designed utilizing the FRM scheme. The strategic arrangement of each sub-band within the proposed filter bank is meticulously prepared to optimize the matching performance. Based on the hearing characteristics of patients, the sub-bands can be distributed in low, medium, and high-frequency regions. Also, the gain can be adjusted per the patient's hearing profile from their audiogram for better auditory compensation.

Further, the Multiband Dynamic Range Compression (MBDRC) technique is applied to address the specific needs of individuals with different frequency-dependent hearing impairments. It involves using dynamic range compression independently to different frequency sub-bands within a filter bank. In MBDRC, the compression parameters, such as compression threshold and ratio, can be adjusted independently for every subband. It allows for a more tailored approach to address the specific hearing needs of different frequency regions. If an individual has more severe hearing loss in high-frequency regions, higher compression ratios and lower compression thresholds can be applied to those subbands to amplify and improve audibility for high-frequency sounds. Once dynamic range compression is applied to each sub-band, the resultant sub-bands are reassembled to yield the ultimate output signal, which can subsequently be transmitted to the speaker or receiver of the hearing aid. A GUI can be helpful for better visualization and parameter control, including gain adjustment and compression parameters of this entire process. With this aim in mind, a GUI has been developed on MATLAB. Different audio files can be imported, and their frequency response can be generated and observed. Based on a person's audiogram, the control parameters can be set to low, medium, or high. Their sub-band distribution in low, medium, and high-frequency regions can be visualized. Further, the filter bank makes automatic gain adjustments, as seen in the GUI. The gain points for each band can also be manually adjusted according to users' hearing characteristics to minimize the error. Also, the compression parameters can be set separately for each subband as per the hearing requirement of the patient. Further, the processed output can be visualized in the output frequency response tab, and the input and output audio signals can be analyzed.

Keywords- Adaptive; Filter; Frequency Response Masking; Hearing impaired; MDRC.

I. INTRODUCTION

The problem of hearing loss among people is increasing day by day. Rehabilitation is needed for 5% of the world's hearing-impaired population. Disabling hearing loss is generally defined as more significant than 35 decibels (dB) [1]. The prevalence of hearing impairments globally has given rise to the dire need for improved auditory assistive devices. Historically, hearing aids have been static, offering limited personalization. Many users frequently reported dissatisfaction with the inability of these devices to adjust to diverse environments or their varying hearing profiles. Thus, improving hearing and speech comprehension for people with hearing loss damages the inner ear, i.e., hair cells[2]. To address the challenges faced by individuals with hearing loss, hearing aids have become a standard solution. Hearing aids (HA) are the devices that hearing-impaired (HI) people need. These are small electronic devices designed to amplify sound, allowing people with hearing loss to better engage in daily activities, communicate, and understand speech. Reconfigurability is the need of the hour to ensure the same hearing device can help various hearing problems. For this, the

Filterbank of HA needs to be reconfigurable [3]. Also, the filter bank should provide sharp transitions with minimum filter order and reduced complexity. So, FRM is an apt technique to meet these HA requirements [4]. HA is sought by people who are hard of hearing to hear sounds clearly in various environments, including quiet and noisy settings.

To achieve this, hearing aids employ various signal processing algorithms, including Dynamic Range Compression (DRC) [5]. DRC is a critical strategy used in signal processing because the damaged ear often has a smaller dynamic range than a normal ear. The dynamic range is the intensity difference between faint and loud sounds. DRC aims to amplify the input signal at specific frequencies in a frequency-dependent manner. Once amplification is applied, DRC reduces or compresses the increase in gain for sounds that exceed a certain threshold, ensuring that the amplified sounds are within a comfortable and audible range for individuals with hearing loss. This is particularly important because people with hearing loss may have difficulty hearing or understanding softer sounds.

Their reception to louder sounds remains relatively similar to that of an average person not hearing impaired. DRC produces the output audible to HI people, which is above the discomfort level of loudness or which is not distorted. Since dynamic range compression typically operates on a single frequency band, multiple dynamic range compressions operate on multiple frequency subbands simultaneously. This allows for more precise amplification and compression across different frequencies, improving speech comprehension and sound quality for individuals with hearing loss.

II. RELATED WORK

An exhaustive literature survey suggests much work is done toward designing filter banks with uniform subband distribution. Very minimal work has been done on the reconfigurable nonuniform filter bank where the distribution of the band is done non-uniformly according to user requirements. Nonuniform filter banks gained popularity due to their ability to replicate the characteristics of critical bands of the human ear.

The author in [6] presented many techniques for reconfigurable filter design based on FIR filters for hearing aids [7]. This type of filter bank was designed using only one prototype filter. Later, an auto-reconfigurable filter bank was designed using 3-level octave interpolated filters [8]. A filter bank with 27 variable subband distribution schemes was designed using the FRM technique, which proved to be the most effective filter bank [9]. However, later, it was calculated that the delay offered by these filter banks with 27 subbands is comparatively more significant. This led to the invention of 16-band reconfigurable filter banks, which used the same FRM techniques [10]. Further, designs of eight subbands showed low delay, but their complexity increased. Finally, the 16-band low-power reconfigurable nonuniform filter bank was designed for hearing aid use [11]. They were ideal for the hearing aid application due to their low power consumption. These filter banks proved to be the best fit for the hearing aid application. A. Sokolova et al. [12] presented a real-time multi-rate, multiband amplification system for hearing aids. The system offering eleven frequency sub-bands and an automatic gain control block with accurate control of dynamic range compression improved the accuracy of hearing aid prescriptions. It developed an open-source software application specifically designed to research hearing loss. In [13], the author introduced mathematical tools to investigate the distortion effects of standard DRC systems with noise and to develop new nonlinear processing methods for numerous signal combinations. The author in [14] developed a hearing aid application based on smartphones. Using the developed GUI, hearing thresholds are fitted by the user, which can be used as an alternate option to the traditional HA. In [15], an introductory method of analyzing signals is proposed. The research is significant as it provides an instance for the people who plan to do signal studies and design GUIs for themselves. Authors in [16] introduced a learning tool for understanding hearing aid compression. A web-based tool has successfully imitated the DSL-v5 hearing aid fitting procedures and prescription gain in nine frequency bands for four speech intensity levels. An amplification technique that flexibly adapts the compressor properties characteristics based on the predicted speech activity in individual Time-Frequency(T-F) units is proposed in [17], where Speech-dominated T-F units with high SNR were compressed fast. In contrast, noise-dominated units with low SNR were compressed slowly. Based on the exhaustive literature review of the existing reconfigurable filter banks, it can be concluded that a reconfigurable filter bank with 16 bands is the best in terms of power efficiency and output delay. Further, we use the Frequency Response Masking technique [18] to enhance the output sound experience and reduce the matching error. These will be more effective in hearing aids owing to the adaptive hearing characteristics of people. Every hearing loss has a different hearing profile and requirements. So, we need to decompose the subbands as per their requirement. This can be done only using reconfigurable nonuniform filter banks as the subband distribution can vary in low, medium, and high-frequency regions per a person's hearing loss characteristics.

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III. METHODOLOGY

In this work, we present a unique methodology to improve the efficacy of hearing aids by designing a Multiband Dynamic Range Compression-based Reconfigurable Frequency Response Masking (FRM) Filterbank. The methodology is delineated in the following steps:

1. The Filterbank was designed employing the FRM approach to partition an incoming audio stream into many subbands.
2. The incoming audio signal is passed through the FRM filterbank decomposed into multiple subbands, each corresponding to a different frequency range. This process ensures that individual frequency components of the signal are isolated for further processing.
3. Gain is adjusted for each sub-band based on the predefined gain values provided to a system based on the dataset of 105 hard-of-hearing people.
4. Each subband signal obtained from the FRM filterbank then undergoes dynamic range compression. The MDRC algorithm adjusts the gain of the signal in each subband based on its level, ensuring that quiet sounds are amplified more than loud sounds. Parameters for MDRC included Compression ratio, Attack time, Release time and Knee point.
5. After processing through MDRC, the subband signals are recombined using a synthesis filterbank, the counterpart of the analysis filterbank from the FRM design, ensuring the recreation of an enhanced audio signal optimized for the end-user.

The proposed system model is presented in Figure 1.

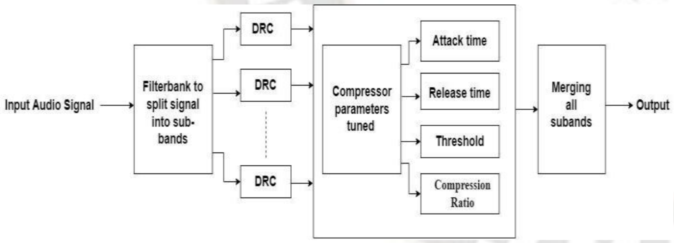

Figure 1. Proposed system model

A. DESIGN OF FRM-BASED FILTERBANK

Multi-branch frequency response masking is employed in a sixteen-band nonuniform reconfigurable filter bank. Two distinguishing factors set it apart from the standard FRM. Instead of employing a solitary band edge shaping filter, multiple band-shaping filters are utilized to create separate bands, and all frequency masking is accomplished by filters derived from a standard prototype filter. The suggested filter bank, as seen in Figure 2, comprises band-edge shaping filters $H_1(z)$, $H_2(z)$, and $H_3(z)$, together with a masking filter.

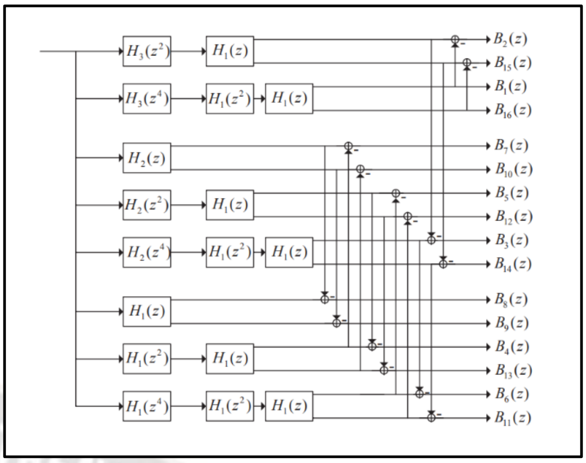

Figure 2. Proposed FRM filter bank structure

Subband allocation in the filter bank is conducted meticulously, considering the audiogram inputs. This allocation process aims to optimize the matching performance for individuals with different hearing impairments. By considering the individual's auditory attributes, modifying the filter frequency response to align with the audiogram is possible. This can be achieved by designating additional frequency bands in the low, middle, and high ranges. The hearing impairment in high frequencies due to aging can be mitigated by incorporating additional high-frequency bands, improving matching performance. The predefined subband cut-off frequencies are listed in Table II.

TABLE II. CUT-OFF FREQUENCIES OF 16 SUB-BANDS

Band	Lower 3 dB frequency	Upper 3 dB frequency
1	-	250
2	250	500
3	500	750
4	750	1000
5	1000	1500
6	1500	2000
7	2000	3000
8	3000	4000
9	4000	5000
10	5000	6000
11	6000	6500
12	6500	7000
13	7000	7250
14	7250	7500
15	7500	7750
16	7750	-

The suggested structure is constructed using a set of eight low-pass and high-pass filters viz.$\{P_j(z), j= 1, \cdots, 8\}$ and $\{Q_j(z), j = 1, \cdots, 8\}$respectively as depicted in Figure 3.

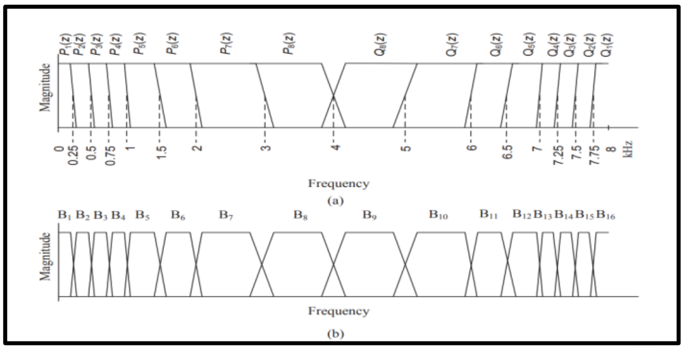

Figure 3. Distribution into 16 sub-bands

The 16 sub-bands are produced by taking the difference of magnitude responses of two consecutive filters, which are given mathematically below in (1) and (2).

$$B_j(z) = \begin{cases} P_j(z) & j = 1 \\ P_j(z) - P_{j-1}(z) & j = 2, \cdots, 8 \end{cases} \quad (1)$$

$$Bj(z) = \begin{cases} Q_{16-j+1}(z) & j = 16 \\ Q_{16-j+1}(z) - Q_{16-j}(z) & j = 9, \cdots, 15 \end{cases} \quad (2)$$

The set {Bj(z), i = 1, ⋯, 16} denotes a collection of 16 subbands. The low pass filters' transfer functions are denoted as Pj(z), where i ranges from 1 to 8. Similarly, the transfer functions of the high pass filters are denoted as Qj(z), where j ranges from 1 to 8.

$$H_{1c}(z) = z-(N_1-1)/2 - H_1(z) \quad (3)$$

$$H_{2h}(z) = H_2(-z) \quad (4)$$

where N_1 is filter length of $H_1(z)$ whereas $H_{1c}(z)$ and $H_{2h}(z)$ are the complement of $H_1(z)$ and the mirror image of $H_2(z)$ around $\pi/2$, respectively.

The sample frequency has been configured to 16 kHz. The determined cut-off frequencies of $H_1(z)$, $H_2(z)$, and $H_3(z)$ are presented in Table III.

TABLE III. CUT-OFF FREQUENCIES

Filter	Cut-off frequency	Normalized cut-off frequency
H1(z)	4KHz	0.25
H2(z)	3KHz	0.1875
H3(z)	1KHz	0.0625

B. DATASET OF HEARING IMPAIRED FOR TRAINING THE SYSTEM

Audiometric threshold data (pure-tone testing) is used to classify hearing loss. Audiometric testing uses different sound intensities over different frequencies to diagnose hearing loss and plots the results on an audiogram. Audiogram configurations and patterns can determine hearing loss etiology and severity. In the proposed research, an elaborative database of 105 hard-of-hearing patients with their hearing profile of both ears, type of hearing loss, chief complaints, and doctor recommendation, if any, is compiled and analyzed to set the filter bank's design specifications.

After analysis of the audiogram of 105 patients, the patients are categorized into normal, mild, moderate, severe, and profound hearing loss, and the system is trained based on this dataset for better audiogram matching and minimum possible matching error. A consolidated graphical representation of the audiogram database for varying hearing loss intensity is depicted in Figure 4.

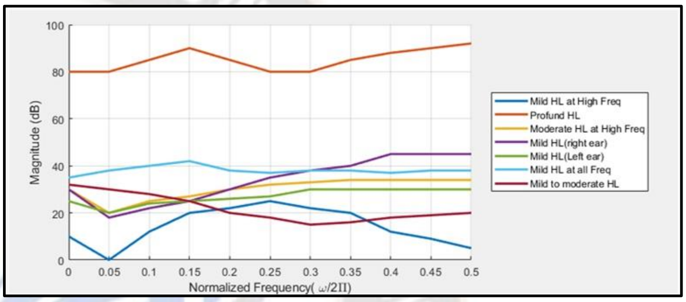

Figure 4. Consolidate graphical representation of Sample audiograms for various types of HL

Based on the database generated, the designed filter bank is trained with the following gain values, as shown in Table I.

TABLE I. SUB BAND GAIN VALUES FOR PROPOSED FILTERBANK

Sub-bands	Low-frequency hearing loss	Mid-frequency hearing loss	High-frequency hearing loss
Band 1	15 dB	25 dB	30 dB
Band 2	15 dB	25 dB	25 dB
Band 3	25 dB	30 dB	30 dB
Band 4	25 dB	30 dB	35 dB
Band 5	30 dB	35 dB	30 dB
Band 6	35 dB	30 dB	35 dB
Band 7	20 dB	25 dB	40 dB
Band 8	15 dB	30 dB	40 dB

C. MULTIBAND DYNAMIC RANGE COMPRESSION

Hearing aids use multiband dynamic range compression to improve speech and other auditory inputs for hard-of-hearing people. This method uses dynamic range compression to partition the audio signal into multiple frequency bands and compress each band separately [19]. Multiband dynamic range compression improves hearing loss patients' auditory

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perception for a more realistic listening experience. Multiband dynamic range compression applies different compression ratios to specific frequency bands to match each person's hearing impairment, unlike conventional hearing aids that enhance all sounds. A more equal and targeted augmentation of auditory output improves speech clarity and sound quality. Multiband dynamic range compression uses digital signal processing to divide the audio signal into frequency bands using a filter bank. Next, a compressor processes each frequency band. This compressor's compression ratio depends on the frequency band's sound level [20]. The number and width of sub-bands depend on the application and the user's hearing.

Once the audio signal has been divided into sub-bands, a separate compression algorithm is applied to each sub-band. The compression ratio, attack time, release time and compression threshold are adjusted independently for each sub-band, allowing for more precise and targeted control over the compression characteristics in different frequency regions [21]. For adjusting these parameters, a MATLAB graphical user interface is designed in the proposed work.

The flowchart depicting the proposed methodology is illustrated in Figure 5.

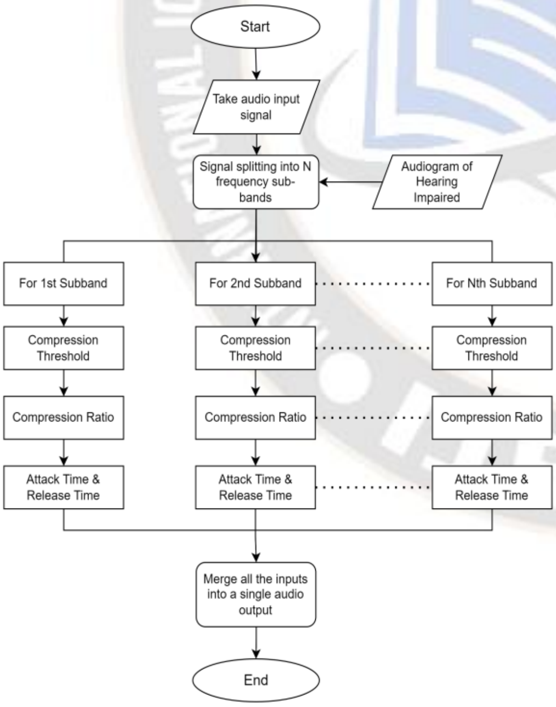

Figure 5. Complete system methodology flowchart

IV. EXPERIMENTATION RESULTS

Here, we present the experimental results of the proposed design. The loss of hearing of the impaired has been categorized into three frequency ranges viz. Low, Mid and High-frequency Hearing Loss. These three scenarios are considered for experimentation, as illustrated in Figure 6.

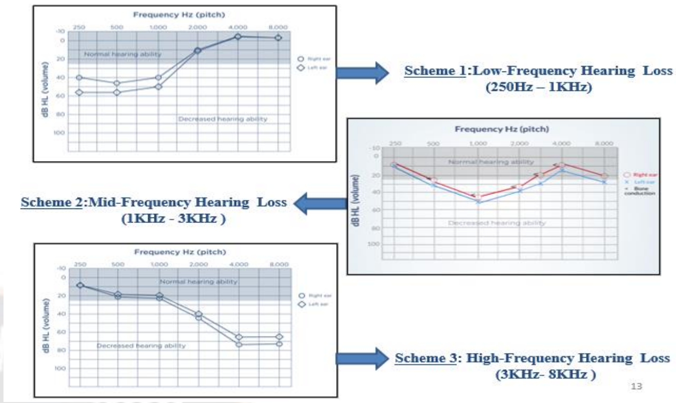

Figure 6. Categories of hearing loss for experimentation

Graphical User Interface (GUI) is designed in MATLAB, where users can load multiple input files, which can then be resampled using the Fast Fourier Transform (FFT). The resampled audio file can be displayed, and the users can hear the audio signal.

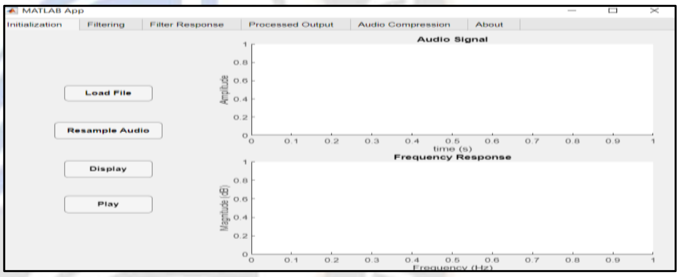

Figure 7. GUI designed for the proposed system

In the designed GUI, we can select the hearing loss category, like low, medium, or high, based on the patient's audiogram. Accordingly, the sub-bands are distributed non-uniformly in the selected frequency range, as shown in Figure 8.

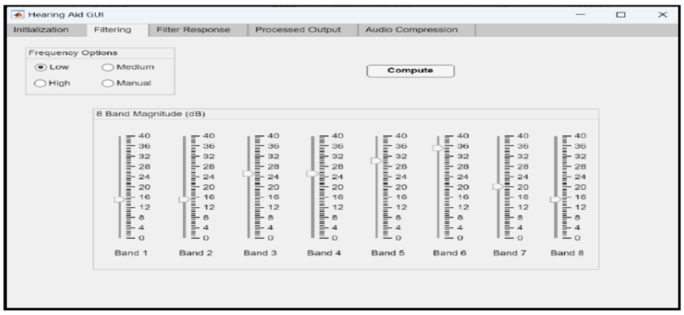

Figure 8. Adaptive gain selection for sub-bands in GUI

The gain values are automatically adjusted in GUI once the radio button of low, mid or high frequency is chosen as per the hearing profile of the patient. These gain values are the optimum values mentioned in Table I that have been fed to the system based on the dataset of 105 hard-of-hearing patients.

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To assess the efficiency and effectiveness of our approach, a sample audio file, gettysburg10.wav was chosen, encompassing a range of frequencies to emulate real-life listening scenarios. The audio file was processed using our proposed system and tested across the three distinct frequency loss categories.

Here, we elucidate the experiment results focusing on all three hearing loss categories in increasing order of frequency.

A. Validation of Results for Low-frequency Hearing Loss

This majorly affects sounds below 1000 Hz, including deeper voices, some environmental sounds, and musical bass notes. The audio file gettysburg10.wav was given as input to the system. The system generates the Frequency response of the input test signal, as shown in Figure 9.

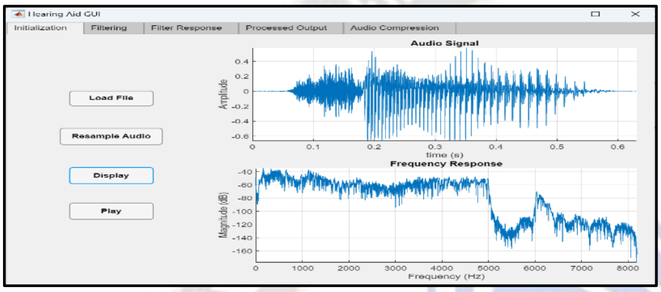

Figure 9. Frequency response of input test signal

The signal was then non-uniformly distributed into 16 sub-bands, specifically targeting different segments of the low-frequency range. The bandwidth of individual sub-bands varied based on the spectral content, and more sub-bands concentrated around the low-frequency region, as depicted in Figure 10.

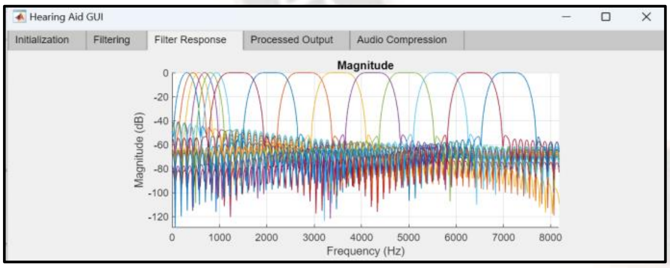

Figure 10. Sub-band distribution in the low-frequency region

Appropriate gain values were provided to each sub-band in the low-frequency region for optimum audibility results, as illustrated in Figure 11.

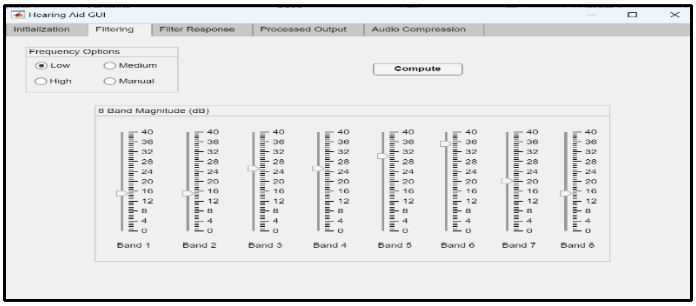

Figure 11. Gain adjustment in low-frequency range

The system then generated the Frequency response of the processed output, as shown in Figure 12.

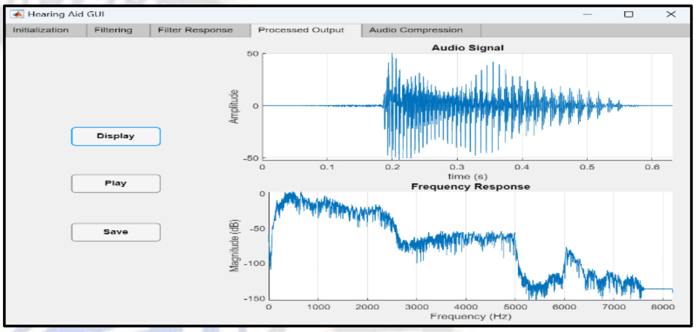

Figure 12. Frequency response of the processed signal

Compression parameters such as Compression Ratio, Compression Threshold, Attack Time, and Release Time can then be adjusted or assigned to each sub-band as per the hearing profile and requirement of the patient [22]. The signal undergoing processing is subsequently compressed, generating a final compressed output signal. The frequency response of this signal can be viewed, and the resulting audio signal can be heard, as depicted in Figure 13.

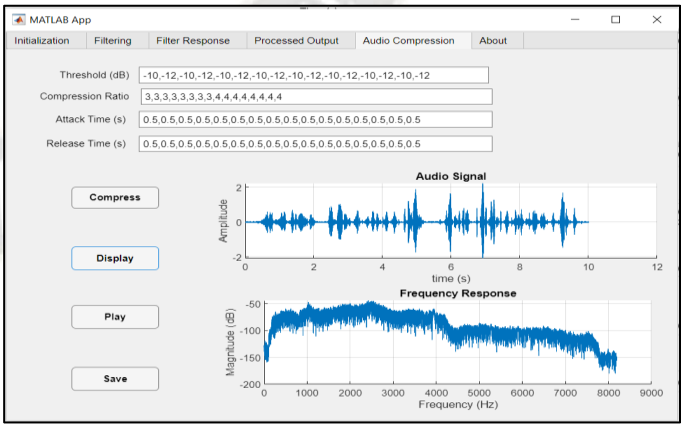

Figure 13. Adjusting the compression parameters on the GUI

Figure 14 illustrates the compression of each sub-band based on the provided inputs. The uncompressed and compressed waveforms are visually represented using two distinct colors, red and blue, with amplitude plotted over time.

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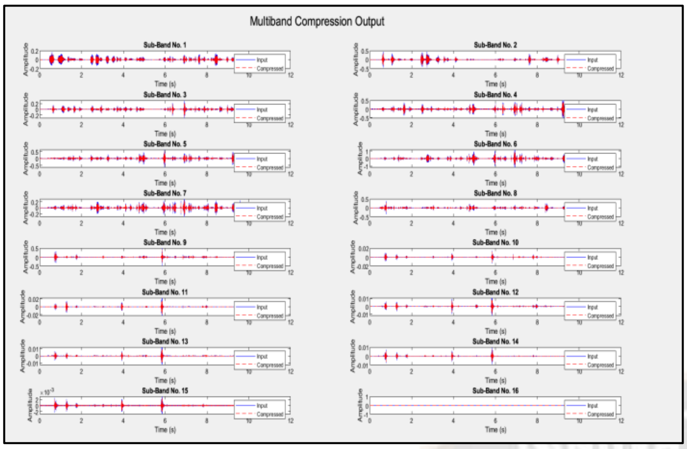

Figure 14. Multiband Compressed Output of all sub-bands for low-frequency hearing Loss

Figure 15 is a representation of the compressed output of one of the sixteen sub-bands that were subjected to MRDC compression.

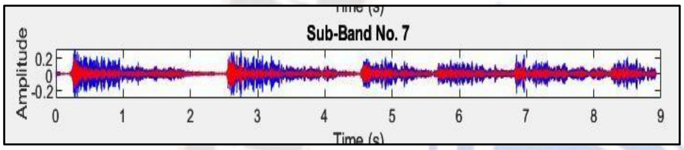

Figure 15. Compressed output of a single sub-band

B. Validation of Results for Mid-frequency Hearing Loss

This predominantly affects sounds ranging from 1000 Hz to 3000 Hz, which encapsulates certain speech elements, music tones, and environmental sounds. For this experiment, the audio sample 'CantinaBand60.wav' was particularly dense in mid-frequency sounds, ensuring a comprehensive testbed was given as input to the system. The graphical user interface (GUI) results for this particular scenario are presented in Figure 16.

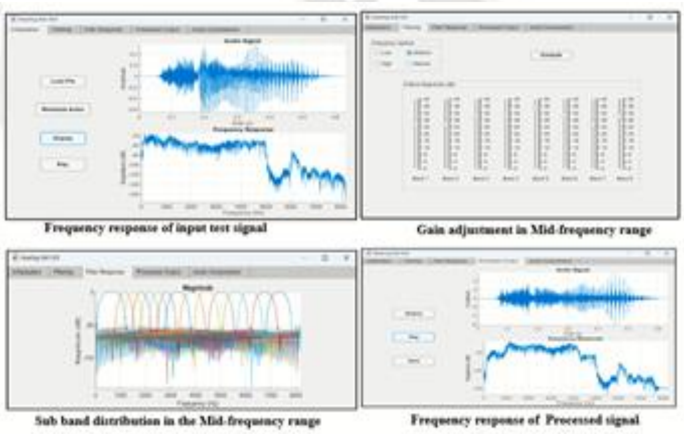

Figure 16. GUI results for Mid-Frequency Hearing Loss

In line with the hearing profile and patient need, compression parameters are set for each sub-band, and the compressed output is generated as shown in Figure 17.

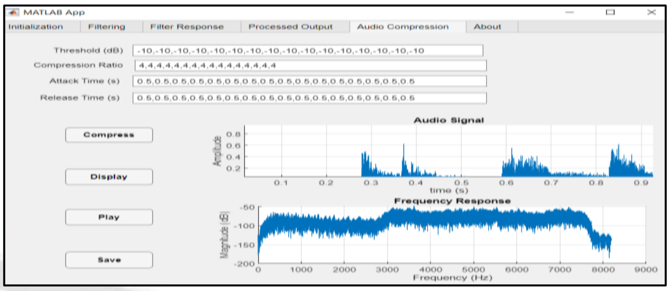

Figure 17. Adjusting the compression parameters on the GUI

Each sub-band is then individually compressed, as illustrated in Figure 18.

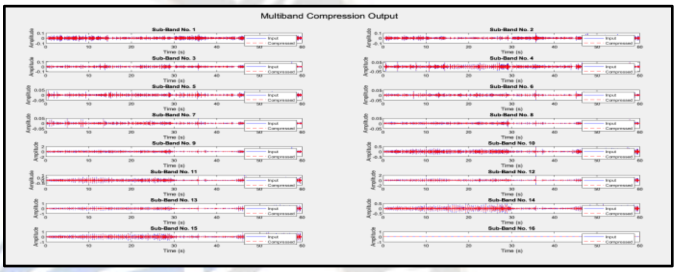

Figure 18. Multiband Compressed Output of all sub-bands for Mid-Frequency hearing loss

C. Validation of Results for High-frequency Hearing Loss

High-frequency hearing loss impacts frequencies above 3000 Hz, encompassing certain speech sibilants, high musical notes, and subtle environmental sounds such as bird chirps. For this experiment, our chosen audio sample was of the word 'please' in a high-pitched sound in women's voices. The input sample was rich in high-frequency content, serving as an ideal platform for evaluation. The graphical user interface (GUI) outcomes for this specific case are depicted in Figure 19.

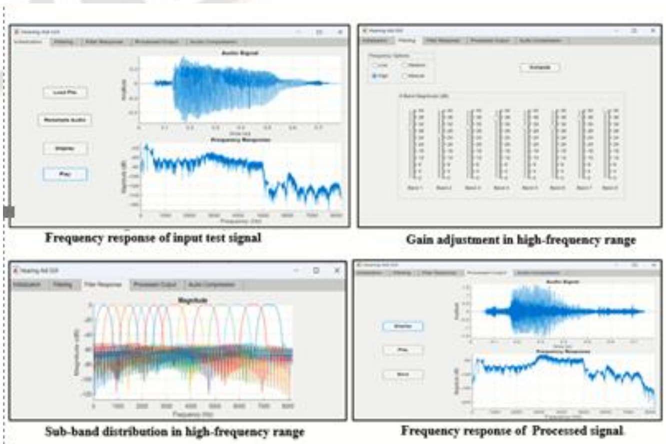

Figure 19. GUI results for High-Frequency Hearing Loss

MDRC is applied individually to all the sub-bands. The resulting original and compressed signals can be observed in Figure 20.

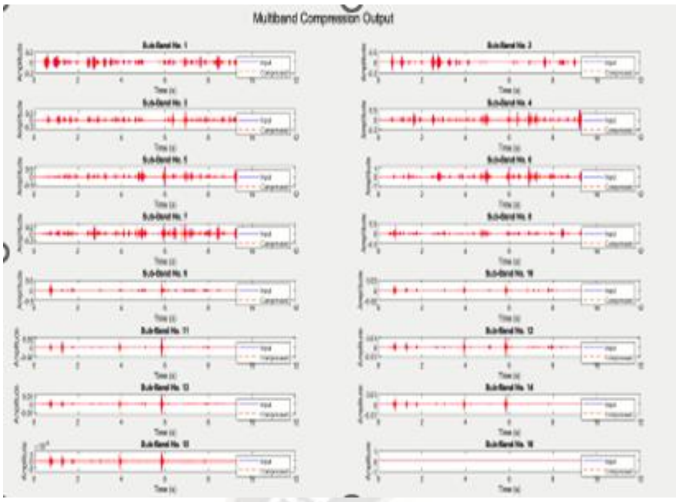

Figure 20. Multiband Compressed Output of all sub-bands for High-Frequency hearing loss

V. CONCLUSION

In the research, we have presented an advanced and nuanced approach to address the challenges the hearing-impaired face across a spectrum of frequency loss types. The basis of our investigation is established on the fundamental principles of Frequency Response Masking (FRM) and Multiband Dynamic Range Compression (MDRC). Our methodology entailed decomposing a sample audio file into 16 nonuniform subbands using the FRM technique, which proved beneficial in isolating individual frequency components, thereby enabling targeted processing. This was followed by the application of MDRC across each of these subbands. The entire process was evaluated across three distinct frequency loss categories: low, mid, and high. By harnessing a comprehensive database of 105 audiograms, our system can automate the gain settings across all subbands. This database-driven approach reduces the need for manual calibrations and offers an efficient, patient-specific adjustment. The use of real-world data ensures the applicability and efficiency of the system in addressing diverse hearing impairments. The Multiband Dynamic Range Compression, when applied across all subbands, showcases a significant enhancement in the auditory experience for the users. By compressing the dynamic range of each subband, we amplify the softer sounds and ensure that loud sounds do not become overpowering, striking a balance crucial for people who are hard of hearing. Our research is complemented by a user-friendly Graphical User Interface designed in MATLAB. This GUI is an accessible platform for audiologists and users to interact with our system, adjust settings, and obtain real-time feedback. Such a tool underscores the practicality and user-centric design of our solution. Thus, our research has put forth a promising step forward in hearing aid technologies. Through a synergistic combination of FRM, MDRC, and a database-driven approach, we took the next step to provide a more natural, transparent, and tailored auditory experience to people who are hard of hearing with the incorporated MATLAB-based GUI for further ease of use and adaptability.

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