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Recurrent Neural Network-Based Variable Bandwidth Filter For Digital Hearing Aid

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Abstract: Assistive listening technologies play a crucial role in enhancing human communication, particularly for individuals with auditory impairments. Hearing aids, compact electronic devices, amplify sound waves and facilitate speech recognition by utilizing microphones to pick up sound waves, processing them through sophisticated signal processing algorithms, and delivering amplified audio via speakers. The evolution of hearing aids has led to significant improvements in size and functionality, enabling the development of more effective devices that can augment the auditory system's capacity for detecting pure tones. Recent advances in artificial intelligence and machine learning have enabled the design of sophisticated digital hearing aid (DHA) systems, which employ advanced filtering techniques, such as those developed using MATLAB programming language and recurrent neural networks. These systems offer improved sound quality, reduced implementation complexity, and enhanced cost-effectiveness through programmability, allowing users to adapt their devices as their hearing loss patterns change over time. The filtering process for digital hearing aid system includes noise reduction filter, frequency-dependent amplification and amplitude compression.

Index Terms—Electromyographic (EMG) signals, Hand pose identification, Computational complexity, Artificial Intelligence techniques.

I. INTRODUCTION

A substantial proportion of the global population suffers from some form of hearing loss, with approximately 10% of the world's population affected by this condition. However, only a fraction of these individuals utilize assistive listening technologies, such as hearing aids, likely due to a combination of factors including system design limitations, customer dissatisfaction with performance, and high upfront costs associated with premium solutions.
Historically, digital signal processing (DSP) algorithms have been developed using specialized hardware components, including signal processing chips (DSP chips) for lower-bandwidth applications. For higher-bandwidth requirements, application-specific integrated circuits (ASICs) were employed to process real-time signals. However, the advent of field-programmable gate arrays (FPGAs) has revolutionized DSP design, offering a more cost-effective and functionally superior alternative.
One key advantage of FPGAs is their ability to provide low power consumption and high processing speeds, making them an ideal platform for implementing real-time speech signal processing algorithms in digital hearing aids. The use of FPGAs as a hardware platform also enables significant reductions in system design cycle times, particularly for applications such as audio, video, and image processing. Furthermore, the flexibility and reconfigurability **of** FPGA-based systems enable efficient implementation of adaptive filtering techniques, which

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are critical for improving speech recognition performance in noisy environments.

A Digital Hearing Aids

Human hearing is a complex process that involves the conversion of sound waves into electrical signals, which are then processed by the auditory system. In contrast, digital hearing aids mimic this process using sophisticated signal processing algorithms and hardware components. A typical digital hearing aid consists of three primary components: the directional microphone, impedance matching network, and amplifier. The outer ear receives ambient noise, while the directional microphone directs it towards the middle ear. The middle ear acts as an acoustic filter, dividing the incoming sound into multiple frequency bands using a technique called spectral shaping. This process is crucial in reducing noise and improving speech clarity. In digital hearing aids, impedance matching networks and amplifiers work in tandem to achieve this goal. Currently available analog hearing aids suffer from several limitations, including inadequate spectral shaping, narrow operating bandwidth, and limited noise reduction capabilities. These deficiencies result in suboptimal performance, leading to poor speech perception in noisy environments. Moreover, analog hearing aids are rigidly fixed devices that cannot be customized to individual patient needs.

Digital hearing aids overcome these limitations by providing full-bandwidth processing, fine-grained spectral shaping, and enhanced noise reduction. As software-driven devices, they offer flexibility and customization capabilities that enable doctors to tailor the hearing parameters of a patient on the same device without replacing components. This contrasts with analog hearing aids, which require separate component replacement. The inner ear functions as a complex system, encoding signals at various frequencies, amplifying low-amplitude signals, compressing high-frequency signals beyond human hearing range, minimizing noise power, and transmitting short pulses to the brain via neural stimulation. In digital hearing aids, these functions are emulated using sophisticated signal processing algorithms, including adaptive filtering, feedback cancellation, dynamic range compression, and inverse fast Fourier transform (IFFT) followed by digital-to-analog conversion (DAC). The resulting frequency domain signal is converted back into a continuous-time signal and delivered to the speaker."

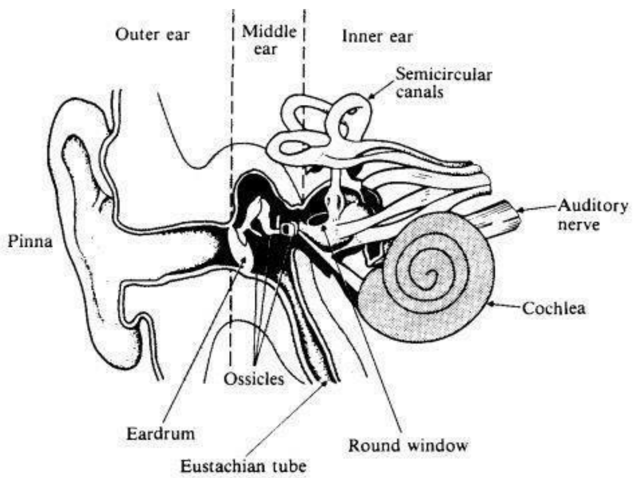

Figure 1.1: Structure of Human Ear

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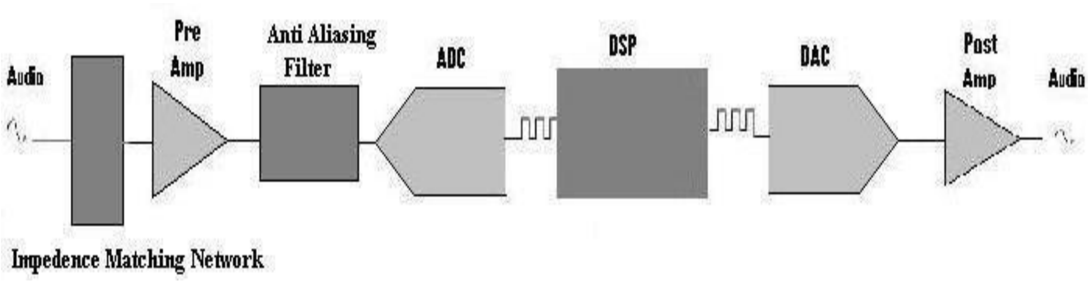

Figure 1.2: Structure of Digital Hearing Aid

Advances in digital signal processing have enabled the development of sophisticated hearing aids that effectively amplify millions of distinct sound signals, thereby significantly enhancing the auditory abilities of individuals with hearing impairments. The inaugural digital hearing aids emerged in the mid 1980s, but these initial models were characterized by limitations such as bulkiness and impracticality. Following a decade of refinement, digital hearing aids became increasingly prevalent, with compact devices integrated into earmolds or discreetly positioned behind the ear. Today, digital technology has become an integral component of daily life, ubiquitous in households with various digital products, including telephones, video recorders, and personal computers

Table 1: Different degree of Hearing Loss

Classification of Hearing Loss	**Hearing level**
Normal Hearing	-10 dB – 26 dB
Mild Hearing Loss	27 dB - 40 dB
Moderate Hearing Loss	40 dB - 70 dB
Severe Hearing Loss	70 dB - 90 dB
Profound Hearing Loss	Greater than 90 dB

The advent of digital signal processing has also yielded significant benefits for hearing aids. One key advantage is hands-free operation, facilitated by real-time noise reduction algorithms. These devices automatically adjust volume and pitch parameters to optimize speech intelligibility in noisy environments. Moreover, they perform thousands of adjustments per second, thereby minimizing background noise, improving sound quality, and offering multiple programmable settings tailored to specific listening situations. Users can seamlessly switch between these programs as needed, ensuring optimal auditory comfort and performance."

1.1.1. *B* Working of digital hearing aid system

. "Digital hearing aids employ sophisticated signal processing algorithms, such as digital signal processor (DSP) techniques, to convert analog sound signals into crisp, distortion-free digital sounds,

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thereby enhancing speech intelligibility and clarity. The integration of advanced microelectronics has led to the development of compact, hands-free digital hearing aids that overcome physical size and power consumption limitations inherent in traditional analog devices. Recent advancements in field-programmable gate arrays (FPGA) technology have enabled the design of highly sophisticated digital hearing aid systems, incorporating adaptive beamformer noise reduction algorithms and multiband loudness correction techniques to achieve superior sound reproduction, reduced device size, and lower power consumption. These devices utilize embedded microprocessors, peripherals, and FPGAs to form a complete system on a programmable chip.

II. PREVIOUS WORK

A closed-loop adaptive filter architecture is employed to continuously cancel acoustic feedback signals, improving sound quality and gain stability in digital hearing aids. However, excessive gain can lead to deterioration of sound quality. To address this challenge, a reconfigurable digital FIR filter bank was implemented, generating 21 distinct sub-bands for optimized masking and sound amplification. This approach enables flexible distribution of sub-bands, reducing complexity and improving matching performance compared to fixed filter banks.

The use of multirate frameworks and adaptive FIR filter banks in personalized hearing assistants is investigated. These systems require the ability to decompose sound waves according to individual patient characteristics, necessitating a sophisticated understanding of acoustic feedback cancellation, signal processing, and sound wave manipulation. The integration of advanced signal processing techniques, such as closed-loop control and adaptive filtering, enables the development of highly effective and personalized hearing aids that can significantly enhance speech comprehensibility

III METHODOLOGY

A PROPOSED SYSTEM

SYSTEM DESCRIPTION

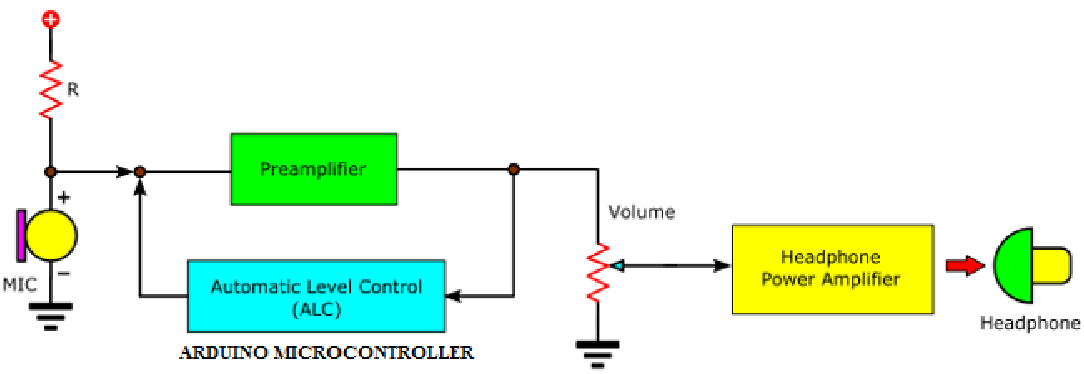

Figure 3.1 Overview of the proposed method

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Figure 3.1 shows the overview of the proposed method. The digital hearing aid comprises a suite of hardware components, including a directional microphone, impedance matching network, anti-aliasing filters, preamplifier, analog-to-digital converter, specific processor for signal processing algorithms, digital-to-analog converter, post-amplifier, and speaker. The acquired signals undergo frequency range and band value calculation, followed by the application and evaluation of various filters using statistical and qualitative methods. A flowchart illustrating the hearing aid's signal processing workflow is presented in Figure 3.2. This methodology enables the assessment of the microphone signal (HAM) through filtering, allowing for optimization of sound quality and speech intelligibility.

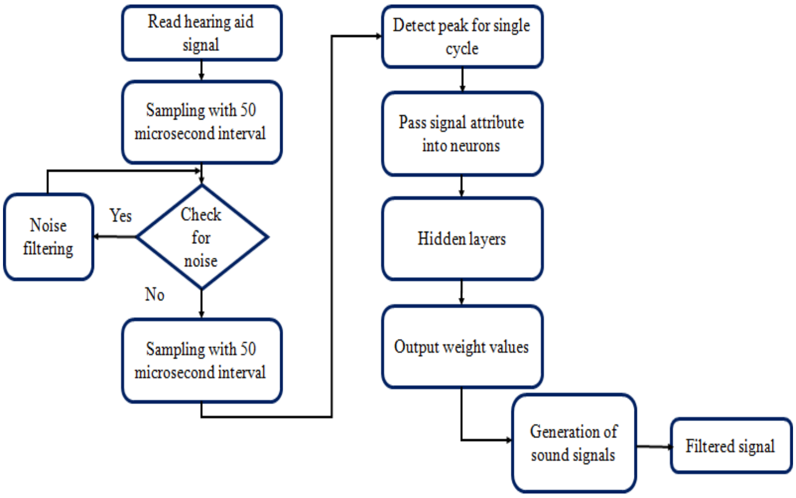

Figure 3.2 Flowchart of the proposed method

The development board components typically include :

1. Power circuit (typically, set up to run from a 9V/12V power supply).
2. Arduino UNO board.
3. Basic output circuitry, such as LEDs.
4. Speaker and microphone with pre-amplifier.

A power supply connected to a suitable source of electrical power.

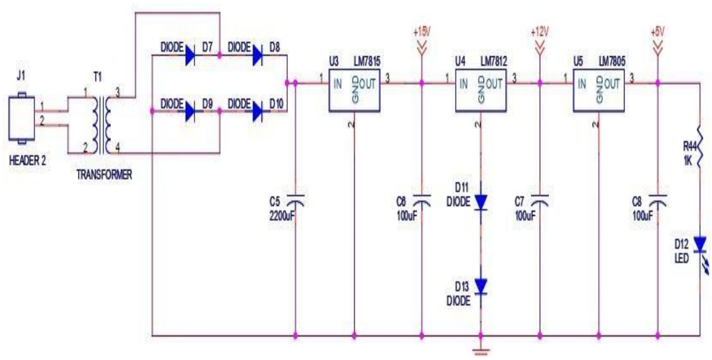

Figure 3.3 Circuit diagram of the power supply

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The power supply circuit diagram is presented in Figure 3.3. This circuit serves as a primary power source for all components, converting AC voltage to DC voltage. The transformer (230V) is used to step down the voltage to 12V AC, which is then rectified by the diode (1N4007). A 1N4007 diode with a forward voltage drop of 0.7V is selected for this purpose.

To filter out unwanted AC components, two capacitors are employed: an AC capacitor that charges and discharges to ground, and a low-pass capacitor that filters the remaining AC signal. The LM7805 and LM7812 regulators are used in conjunction with these capacitors to maintain a stable DC voltage supply. The output of this regulator is then filtered by a low-pass capacitor before being applied to the load circuit, consisting of an LED (Vf = 1.75V) and a resistor.

Figure 3.4 Circuit diagram of the controller circuit with display

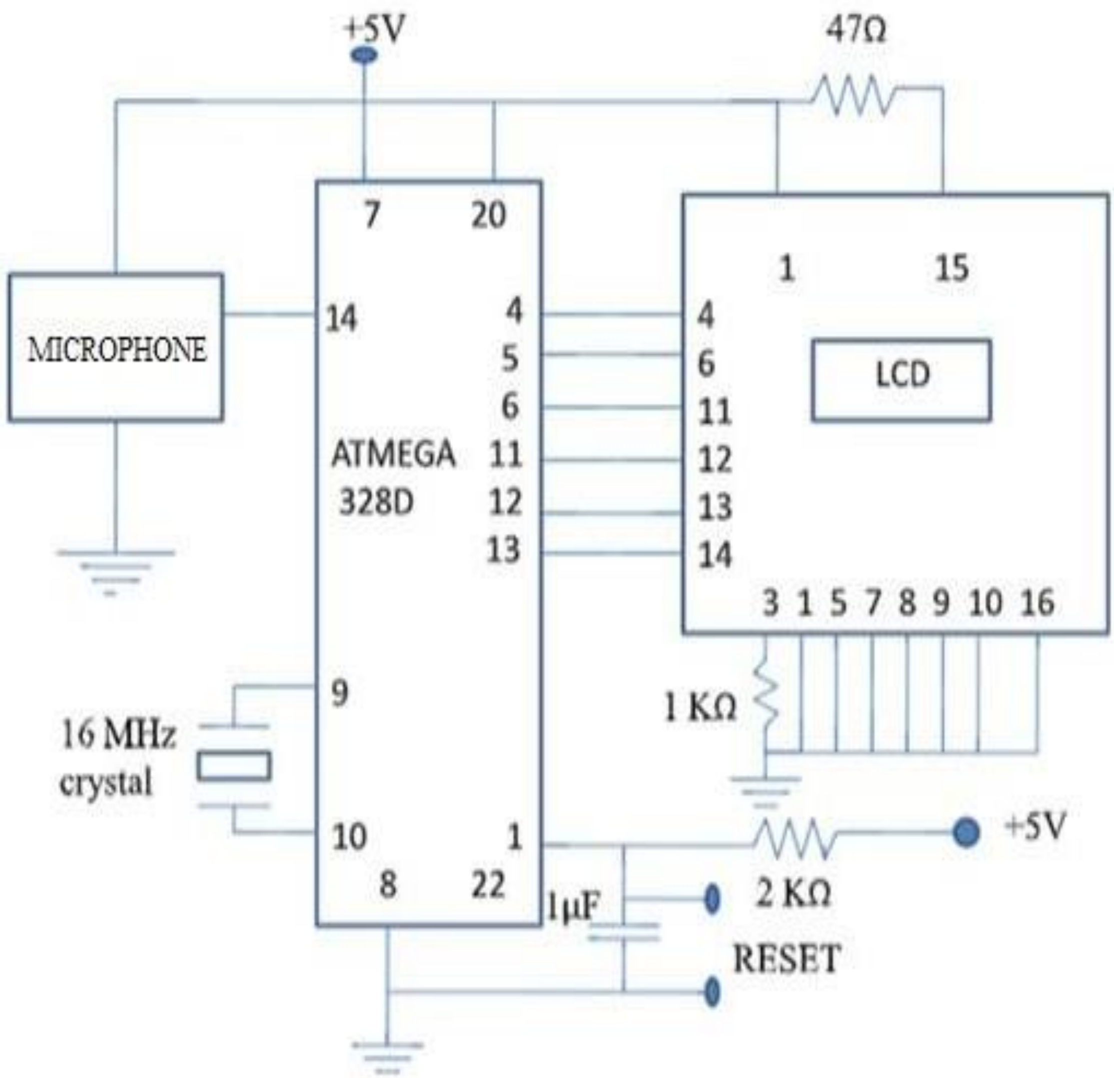

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Figure 3.4 illustrates the circuit diagram of the controller circuit.

The negative feedback regulation is achieved through the LM7905, ensuring the circuit maintains a stable voltage supply despite variations in load or input conditions. This regulated DC voltage powers all components within the digital hearing aid system.

The controller circuit integrates an Atmega328P microcontroller, a high-performance, low-power CMOS 8-bit processor optimized for efficient data processing and storage. Equipped with 32KBytes of in-system flash program memory, the microcontroller is programmed for optimal performance. It features six channels of pulse width modulation (PWM), one of which controls the LCD backlight, adjusting its intensity based on PWM width variations. Additionally, the Atmega328P includes six 10-bit analog-to-digital conversion channels, enabling precise detection of voltage changes across light-dependent resistors.

To address hearing loss and mitigate noise distortion, the system incorporates three key processing stages: **noise addition**, **noise reduction filtering**, **frequency shaping**, and **amplitude compression**.

Noise Addition

The input speech signal for the system begins as a clean signal. To simulate real-world conditions, Adaptive White Gaussian Noise (AWGN) and random noise are introduced using MATLAB functions. AWGN has a continuous, uniform frequency spectrum over a specific frequency range with equal power per Hertz. It comprises all frequencies at equal intensity and follows a Gaussian probability density function.

3.1 Noise Reduction Filter

A major concern for individuals with hearing loss is the ability of a hearing aid to distinguish intended speech signals in noisy environments. To address this, the system employs a wavelet-based noise reduction filter function to suppress unwanted noise effectively.

3.2 Frequency Shaper

Hearing aid users often complain that the device amplifies all signals indiscriminately rather than focusing on desired sounds. Many hearing-impaired individuals struggle with high-frequency sounds. To address this, the frequency shaper corrects for hearing loss at specific frequencies, particularly high frequencies, ensuring better speech clarity.

3.3 Amplitude Compression

Amplitude compression controls the overall gain of the speech amplification system, ensuring the amplified signal remains within the saturation power limit. Saturation power is the point at which the sound signal begins to distort. This function prevents such distortion by dynamically managing the signal's gain.

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These processing stages collectively enable the system to adapt to varying environmental conditions and address a wide range of hearing loss scenarios, ensuring a balanced and user-friendly auditory experience.

3.4 Advantages

- Effectively compensates for losses caused by varying distances and environmental conditions between the source and the hearing aid, accommodating different types of signals.
- The proposed filter addresses diverse hearing loss cases, providing an acceptable frequency range tailored to user needs.

IV. RESULTS AND DISCUSSION

4.1 Simulation Results

The proposed methodology accesses the microphone of the host computer, capturing audio input and plotting the response in real time. The following plots and figures summarize the simulation results:

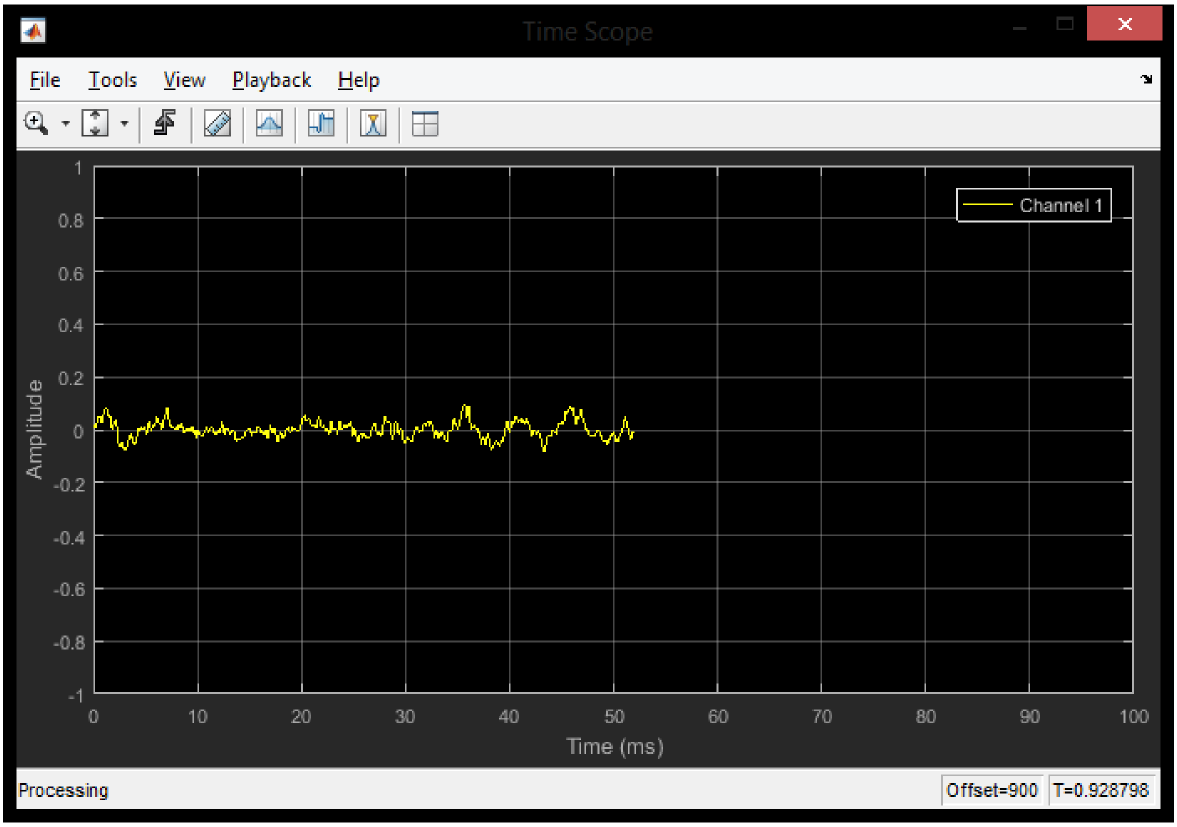

Figure 4.1: Time Response Plot

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The time response plot illustrates amplitude variations within the range of -0.1 to 0.1. This plot is dynamic, updating instantly with real-time audio data.

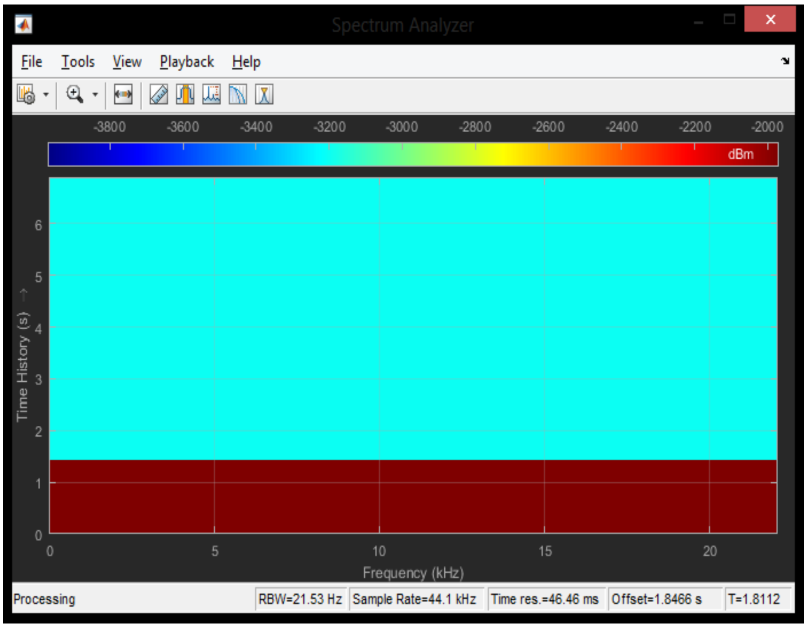

Figure 4.2: Frequency Spectrum

The frequency spectrum demonstrates the distribution of frequency bands along the x-axis, with corresponding time scales on the y-axis. Using a sampling frequency of 44 kHz, the maximum frequency captured is approximately 25 kHz.

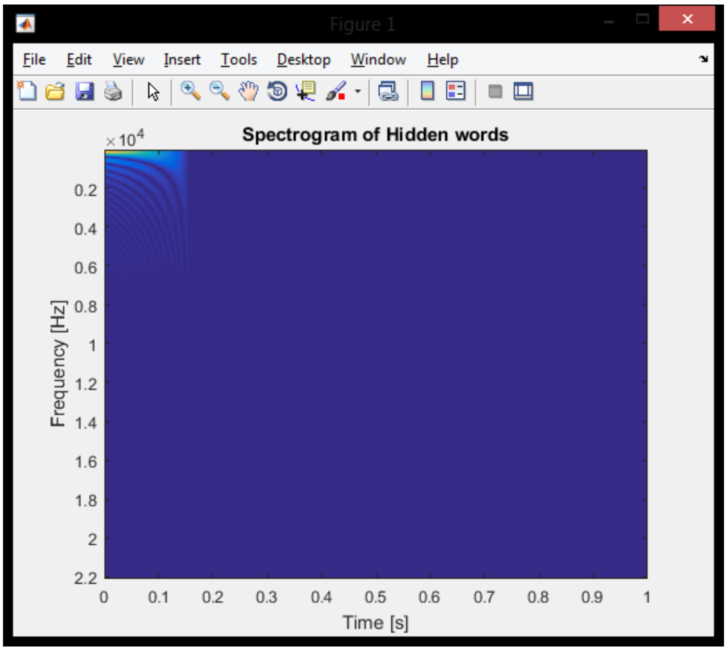

Figure 4.3: Spectrogram Plot (20 kHz)

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The spectrogram visually represents the temporal evolution of spectral features, showing filtered signals across time and frequency. It highlights distinct words and patterns emerging after environmental noise is removed.

In **Figure 4.3**, the spectrogram of a user-provided audio sample at 0.1s predominantly features frequency bands concentrated within 20 kHz. These frequency distributions vary depending on the speaker, reflecting natural variability in human speech.

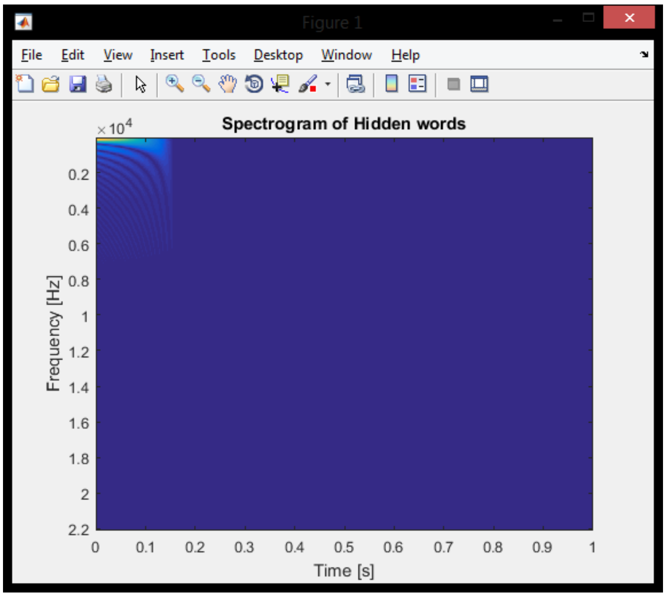

Figure 4.4: Spectrogram Plot (1 kHz - 20 kHz)

This plot displays the spectrogram for a user pronunciation recorded between 0.1s and 1s, where the frequency bands range from approximately 1 kHz to 20 kHz.

Hardware Simulation Results

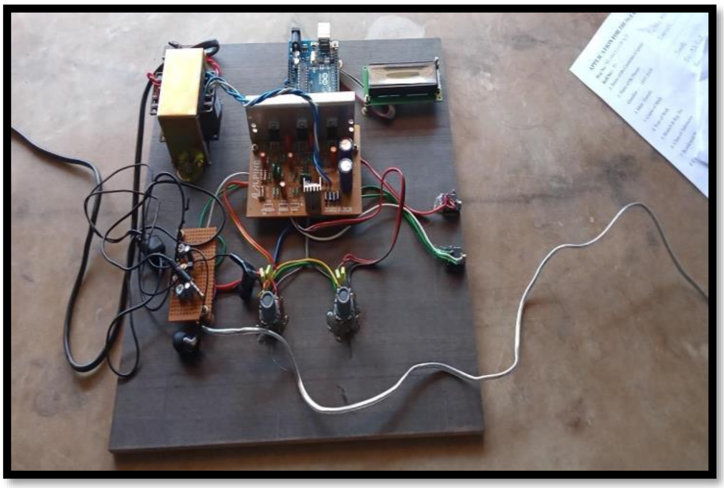

Figure 4.5: Hardware Hearing Aid

The hardware implementation results are displayed, showing the physical hearing aid device integrated with the proposed methodology. This device demonstrates the practical application and real-world functionality of the developed system.

V. CONCLUSION

Our proposed system offers an efficient and cost-effective digital hearing aid solution with variable bandwidth filtering capabilities, tailored to individual patient needs. The primary goal of this research is to develop a comprehensive digital signal processing platform for digital hearing aids, addressing the diverse requirements of patients with hearing impairments. The system design accommodates the specific hearing profiles of users, enabling customized tuning for enhanced auditory support. Advanced speech processing algorithms, such as dynamic range compression and feedback cancellation using adaptive filtering, have demonstrated significant improvements in speech intelligibility, even under high signal-to-noise ratio (SNR) conditions. Real-time processing constraints, such as signal processing latency, memory limitations, and processing power, are carefully considered in the design. To address these challenges, techniques like overlap-add and overlap-save algorithms are employed. Additionally, strategies for de-correlating pre-filters are integrated to address correlation issues in adaptive feedback cancellation (AFC). The feedback path model plays a critical role in effective AFC implementation, particularly in resolving correlations between near-end signals and loudspeaker outputs. By incorporating de-correlating pre-filtering techniques within the AFC algorithm, the system effectively mitigates correlation artifacts, enhancing overall performance. This research contributes to the development of a robust digital hearing aid platform that adapts to real-world conditions, improves speech intelligibility, and provides an optimized auditory experience for individuals with hearing impairments.

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