

Software Simulation of DSP algorithms implemented in Binaural Hearing Aids

Vivek Dalal*, Aayush Bansal**, Pratik Bhandari***, Unmesh Barhate****

* (Department of Electrical Engineering, VJTI, Matunga, Mumbai, India)

** (Department of Electrical Engineering, VJTI, Matunga, Mumbai, India)

*** (Department of Electrical Engineering, VJTI, Matunga, Mumbai, India)

**** (Department of Electrical Engineering, VJTI, Matunga, Mumbai, India)

ABSTRACT

Hearing aids are devices used by hearing impaired persons to counterbalance the hearing loss. They cannot completely overcome the perceptual distortions caused by a hearing loss but assist the user to interpret speech. Continuous efforts are being made to overcome these distortions and improve the hearing experience. The recent advancements include the introduction of digital binaural hearing aids which consist of state of the art DSP algorithms and cutting-edge hardware. The focus of this paper is on software simulation of various DSP algorithms used in binaural hearing aids. Noise cancellation, controlled amplification, frequency transportation, spatial estimation and power compression algorithms are explained and simulation results for the same are specified.

Keywords – Binaural hearing aids, Noise cancellation, amplification, frequency transportation, spatial estimation, power compression, LMS, MATLAB®.

1. INTRODUCTION

The human ear detects and analyzes sound waves by transduction (converts sound waves into electrochemical impulses) [1]. The wave's frequencies indicate the pitch of sound, the wave's amplitude indicates the loudness of sound and the timbre of sound is detected by various frequencies that make up a complex sound wave [2].

The normal hearing of a person may get affected due to various factors including age, noise, genetic disorders illness, physical trauma etc. resulting in a hearing loss.

The hearing aid processes sounds reaching the ear. Essentially, it consists of three parts: microphone, processing unit and receiver. Microphone converts sound into electrical impulses, processing unit modifies the electrical impulses, and receiver decodes impulses back to sound.

The two types of hearing aids available are analog hearing aids and digital hearing aids. Analog hearing aids work in the continuous domain and amplify all sounds, both speech and noise, in the same manner. Some analog hearing aids are

programmable, having a provision to switch between modes and select the one which best suits the user.

They have evolved over the years with the advent of digitization and DSP algorithms.

Digital hearing aids convert the sound waves into digital domain signals and produce an exact duplication of sound. It allows for more complex processing like piecewise linear amplification, adaptive noise cancellation, frequency transportation, dynamic feedback cancellation, telecoil and integration with mobile devices [3]. Due to these reasons, digital hearing aids are being preferred over their analog counterparts.

Another classification of hearing aids is done based on whether the loss is in one ear (known as monaural hearing loss) or both the ears (known as binaural hearing loss).

The brain integrates sensory information coming from both the ears while interpreting the sound. If a hearing impaired person has hearing loss in both the ears then binaural (both ears) hearing aids are selected over monaural (one ear) hearing aids due to the following advantages. Binaural hearing aids provide better spatial estimation of source of sound and greater speech understanding in noisy environment. It also reduces the need for amplification because binaural listening provides a boost to the signal. Studies have shown that without constant auditory stimulation the brain loses its capability to process the sound information. Person using a hearing aid in one ear may lose the speech recognition ability of the other ear. Thus, it is better to use binaural hearing aids to avoid deterioration of the ear [4].

The paper consists of the following sections. Firstly, DSP algorithms implemented in binaural hearing aids are explained in Section 2. This is followed by the simulation results of the algorithms in MATLAB® in section 3. The last section concludes the paper.

2. SYSTEM

The system explained below is a simulation of digital binaural hearing aids. It comprises of two microphones (one for each ear), a

digital processing unit (simulated in MATLAB®) and two speakers (one for each ear).

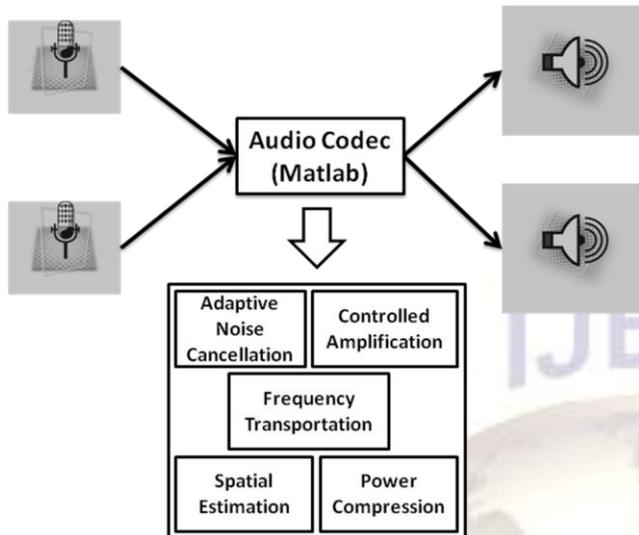


Figure 1: System Block Diagram

The user's (hearing impaired person's) audiogram is taken as an input. An audiogram is a graph that shows the audible threshold for standardized frequencies as measured by an audiometer [5]. The Y axis represents intensity of loss measured in decibels and the X axis represents frequency measured in Hertz. The person's response is measured and the loss in dB is marked on the audiogram.

Firstly, the audiogram inputs taken are linearly interpolated to form straight lines with deterministic slopes. Consequently, the dB loss at each frequency is registered into a matrix.

The sound signal received at each of the microphones is given as an input to the system (simulated in Matlab) which stores it as two separate mono sound files. These signals are in time domain. They are first converted to frequency domain by taking n-point Fast Fourier transform (FFT) where n is the next higher power of two greater than the length of the signal. The amplitude at each frequency is converted in dB scale by using the logarithmic relationship for power:

$$\text{Amplitude (in dB)} = 10 * \log_{10}(|\text{Amplitude}|) \quad \text{---(1)}$$

The following algorithms have been implemented successfully in MATLAB®.

2.1 ADAPTIVE NOISE CANCELLATION ALGORITHM:

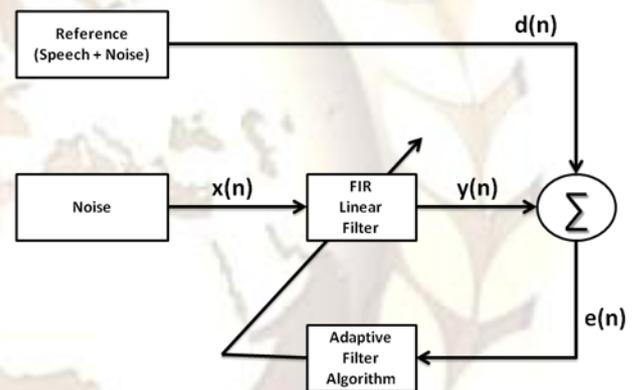
Speech is a random time varying signal. Arrangements have to be made for adjusting filter coefficients according to the statistical parameters

of the signal. In digital filters the coefficients are fixed and time invariant.

Adaptive filters are FIR filters of finite length whose coefficients are converging in nature. In adaptive filtering, two processes, the filtering operation and the weight adjustment, run simultaneously. It enables the system to adjust to the changing statistical conditions of the environment unlike digital filters. Thus, an adaptive filter is preferred over digital filters in binaural hearing aids for effective noise cancellation.

2.1.1 LMS ALGORITHM:

The Least Mean Square (LMS) algorithm, introduced by Widrow and Hoff in 1959, is a robust adaptive algorithm which uses a gradient-based method of steepest decent producing least mean squares of the error signal [7]. LMS converges the filter coefficients according to the error signal. Its organization and pipeline structures are easy to build. It neither requires correlation function calculation nor matrix inversions.



d(n): Reference Signal	y(n): Output Signal
x(n): Input Signal	e(n): Error Signal

Fig. 2: LMS Algorithm Block Diagram

This algorithm calculates the gradient from the available statistical data and adjusts the linear filter coefficients $w_0, w_1 \dots w_n$ accordingly, where n is the number of coefficients. LMS incorporates an iterative procedure that makes successive corrections to the coefficients in a direction opposite to that of the gradient.

The LMS algorithm initiated with arbitrary value of filter coefficients converge appropriately when the particular condition is satisfied.

$$0 < \mu < 1/\lambda_{\max}$$

where μ is the convergence factor of correlation matrix and λ_{\max} is the largest eigen value of the correlation matrix [8].

The algorithm performs the following steps iteratively.

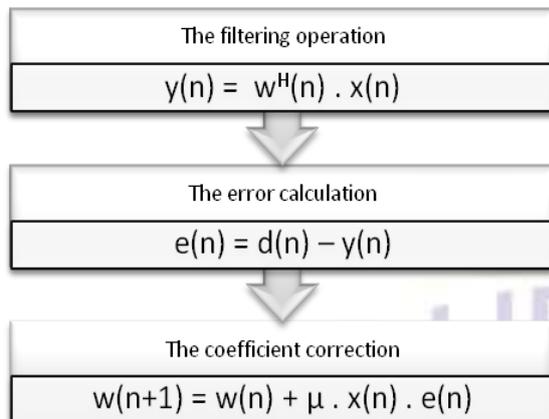


Fig. 3: LMS algorithm

The convergence factor is inversely proportional to the eigen value. If the eigen values are widespread, then convergence may be slow. The values are calculated by determining the ratio of the largest eigen value to the smallest eigen value of the matrix. Initially, a larger value of μ is taken for faster convergence. After a few iterations, a smaller value of μ is taken for getting the correct filter coefficients.

After performing the iterations it tends to converge to w_{opt} (optimum filter).

$$\lim_{n \rightarrow \infty} w(n) = w_{opt} \quad \text{---(2)}$$

Infinite iterations cannot be taken, so the algorithm implements:

$$\lim_{n \rightarrow p} E [w(n)] = w_{opt} \quad \text{---(3)}$$

where, E and p are the expected value of the filter coefficients and the number of iterations respectively.

This eventually leads to the minimum mean square error and noise is filtered out resulting in a noise free speech signal.

2.2 FREQUENCY TRANSPORTATION:

There are many users who have a hearing loss in a particular frequency range. Most of the times amplification algorithms are capable of increasing the loudness of sound to the required level but in certain cases where there is extensive damage, even amplification cannot provide required loudness.

Two common types of hearing loss are Presbycusis and Meniere's disease. Presbycusis is an age-related hearing loss that occurs in the high frequency range (4000 Hz to 8000 Hz) [9]. It is a common disorder associated with aging. Meniere's

disease causes sensorineural hearing loss in the low frequency range (125 Hz to 1000 Hz) [10]. Frequency transportation (shifting) shifts the signal from the frequency range with extensive loss to the range where the user can hear normally. In the case of Presbycusis, the frequencies from 4000Hz to 8000Hz are shifted to lower frequencies whereas in the case of Meniere's disease, the frequencies from 125Hz to 1000Hz are shifted to higher frequencies.

2.3 CONTROLLED AMPLIFICATION OF AUDIO SIGNAL:

The amplification applied to an audio signal depends upon the configuration of the hearing loss. The configuration, or shape, of the hearing loss refers to the degree and pattern of hearing loss across frequencies (tones) [11]. The system incorporates the following modes of amplification. The left and the right ear responses are amplified individually depending on the degree of hearing loss in the respective ear.

2.3.1 Low frequency amplification mode:

The frequencies ranging from 20Hz to 500Hz are termed as low frequencies [12]. The hearing impaired person has hearing loss only in this range i.e. a dB loss (of more than 15dB) is shown on the audiogram. For such a user only the low frequencies are amplified by adding an amplification factor (in dB). The remaining frequencies are passed without any amplification.

2.3.1.1 High frequency amplification mode:

The frequencies ranging from 2000Hz and above are termed as high frequencies [13]. The high frequency components of the audio signal are amplified by adding an amplification factor and the rest are passed without any amplification.

2.3.1.2 Ubiquitous amplification mode:

If the audiogram of the user shows considerable loss over the entire frequency range or varied parts of the frequency range then amplification is performed over the entire range.

Table 1: Classification of hearing loss [6].

Degree of Hearing loss	Hearing loss range (dB HL)
Normal	-10 to 15
Slight	16 to 25
Mild	26 to 40
Moderate	41 to 55
Moderately severe	56 to 70
Severe	71 to 90
Profound	91+

2.3.2 Determination of the amplification factor:

Once the frequency range over which there is a loss, is determined, the adding of a

suitable amplification factor is done. The amplification factor is determined by the degree of hearing loss at that particular frequency (table (1)). As the severity of hearing loss increases the amplification factor added goes on increasing. The amplification factor also depends upon the maximum sound amplitude that humans can perceive without causing damage to ear. This value is around 140dB SPL [14]. In this system, the amplification factor varies from 10dB for slight hearing loss to 40dB for severe and profound hearing loss.

2.3.3 User selective modes:

If the user is not satisfied by the performance of the system in the default mode (in terms of amplification), amplification factor applied to the signal can either be increased or decreased by the user to attain the desired output.

2.4 SPATIAL ESTIMATION:

A sound wave originating from a source falls on both the ears at the same time, but with a difference in amplitude level. For a person having no hearing loss, the small difference in amplitude is detected and the direction from which the sound has originated is easily understood. But for a hearing impaired person, the difference cannot be distinguished by the user as the hearing aids do not provide very high sensitivity. They have maximum sensitivity only for sound originating directly in the front and behind the user [15]. Decent spatial estimation can be achieved either by using directional microphones or DSP algorithms.

The system incorporates the spatial estimation algorithm explained below. The statistical data of the speech signal is estimated. The amplitude is higher for the microphone which is closer to the sound source and lower for the other. The response of the left and the right ear are compared and the one having higher amplitude is increased by 20% and the other is decreased by the same amount. Thus, there is a noticeable difference in amplitude between the two microphone outputs which can be sensed by the ears. Depending on the intensities of sound received at both the ears, the person will be able to judge the direction of sound. Once the user is aligned along the sound direction, binaural hearing aid has its maximum sensitivity of detection. Thus spatial estimation helps in locating the origin of sound source and increases speech intelligibility.

2.5 POWER COMPRESSION:

One of the key performance measuring aspect of any electronic system is the power consumed by it. All the hearing aids used today are battery powered. Therefore, it is essential to reduce the power consumption as much as possible to have prolonged battery life.

Power compression algorithm aims at reducing the power consumption by attenuating insignificant frequency components of the signal and setting a maximum power level limit for the entire frequency range. A lower level threshold (P_{low}) in dB is set, such that all the frequency components having a power level less than P_{low} are attenuated and brought down to 10% of its original value. Also, an upper power limit (P_{sat}) in dB is set, such that any frequency having power level greater than P_{sat} are made equal to P_{sat} . The values of P_{low} and P_{sat} can be set as per the requirement of the system. This reduces the total power of the signal. Different modes available in the system are explained below:

2.5.1 Low frequency mode(Only speech):

If the user wants to hear only speech signals then this mode can be activated. The speech signal (taken from 20Hz to 4000Hz) is passed through the system by applying the power compression algorithm and the remaining frequency range (up to 20000Hz) is compressed to 10% of its original value.

2.5.2 Default operating mode:

The system is in the default operating mode when it is first turned on. Power compression algorithm is applied to all the frequency components (20Hz to 20000Hz).

3. SOFTWARE SIMULATION RESULTS

The software simulation of the above explained algorithms in MATLAB® are given below. The input signals used are a 1000Hz frequency tone (40 dB) and a speech signal (0-40dB).

3.1 ADAPTIVE NOISE CANCELLATION:

Adaptive noise cancellation using LMS algorithm is implemented. The noise signal (shown in red) is removed and speech is retained successfully.

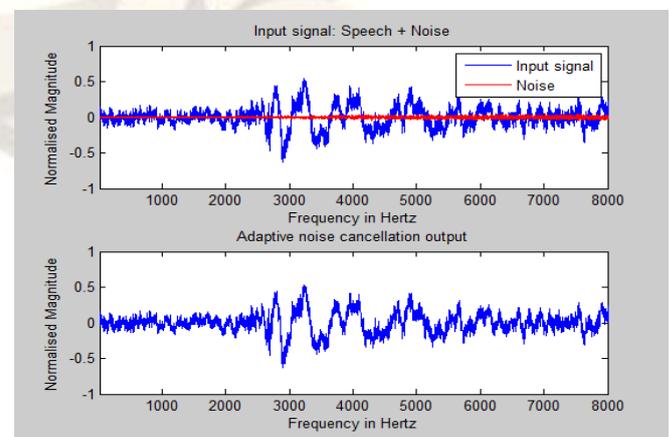


Fig. 4: Adaptive Noise Cancellation

3.2 FREQUENCY TRANSPORTATION:

It is assumed that the user is suffering from Meniere's disease and thus the signals are shifted to the high frequency range. The Fig. shows the shifting of a 1000Hz frequency tone to the high frequency range. Thus enabling the user to hear it properly.

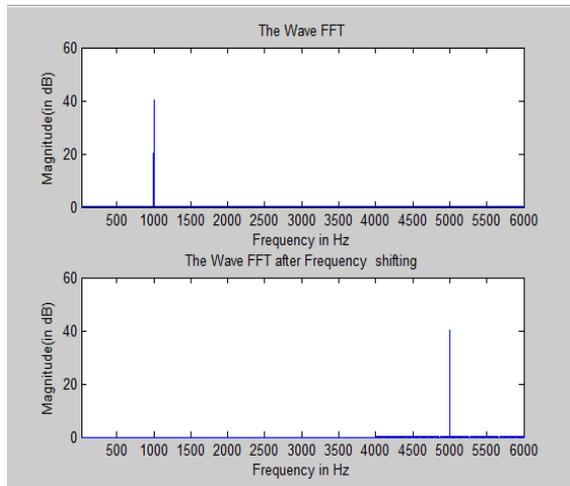


Fig. 5: Frequency Transportation

3.3 CONTROLLED AMPLIFICATION:

The ubiquitous amplification mode is activated assuming the user has loss over the entire frequency range. An amplification factor of 20dB is applied.

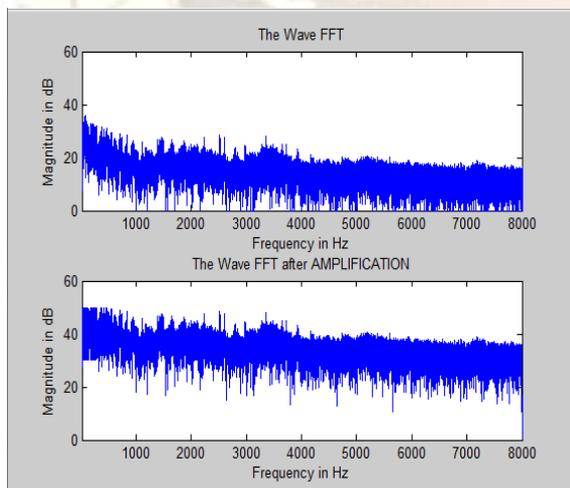


Fig. 6: Amplification

3.4 POWER COMPRESSION:

The default mode of power compression is activated. The values for P_{low} (shown in red) and P_{sat} (shown in green) are taken as 15dB and 25dB respectively for the sake of understanding the results better. As these values are reprogrammable, they can be set according to the requirement of the system.

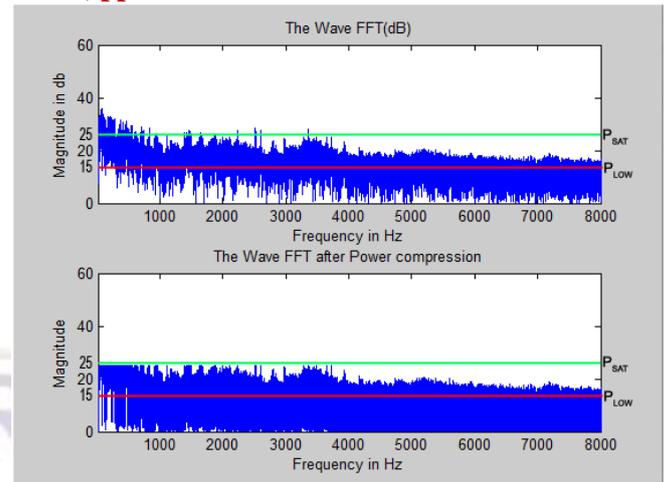


Fig. 7: Power Compression

4. CONCLUSION

We discussed the various DSP algorithms used in binaural hearing aids and showed the simulations for the same in MATLAB®. The individual algorithms explained above should be performed in a sequential order to correctly simulate binaural hearing aids. First, Adaptive noise cancellation should be performed followed by frequency shifting and controlled amplification. These should be followed by spatial estimation and power compression. Features such as dynamic feedback cancellation, environment adaptive algorithms for amplification and power compression, telecoil, directional microphones, integration with smartphones etc. can be incorporated in the system. Also, these algorithms can be tested on a DSP processor to realize their practical constraints and further improve on them.

5. ACKNOWLEDMENT

The authors are deeply indebted to Prof. Dr. Alice N. Cheeran, Department of Electrical Engineering, VJTI for her cheerful guidance, valuable suggestions, constant encouragement and supervision during this project work.

References

- [1] <http://www.britannica.com/EBchecked/topic/175622/human-ear>
- [2] <http://www.physicsclassroom.com/class/sound/u1112d.cfm>
- [3] <http://www.fda.gov/MedicalDevices/ProductsandMedicalProcedures/HomeHealthandConsumer/ConsumerProducts/HearingAids/ucm181470.htm>
- [4] http://www.dineenandwestcott.com.au/docs/pdf/aid_technology.pdf

- [5] <http://www.babyhearing.org/HearingAmplification/HearingLoss/audiogram.asp>
- [6] <http://www.asha.org/uploadedFiles/AIS-Hearing-Loss-Types-Degree-Configuration.pdf>
- [7] http://etd.lib.fsu.edu/theses/available/etd-04092004-143712/unrestricted/Ch_6lms.pdf
- [8] <http://www.ece.utah.edu/~mathews/ece6550/chapter5.pdf>
- [9] http://www.isa-audiology.org/periodicals/1971-2001_Audiology/1979,%20%20Audiology,%20%20Vol.%20%2018/No.%204%20%20%28265-352%29/Robinson%20Sutton,%20%20Audiology,%20%201979.pdf
- [10] http://en.wikipedia.org/wiki/Sensorineural_hearing_loss
- [11] <http://www.asha.org/uploadedFiles/AIS-Hearing-Loss-Types-Degree-Configuration.pdf>
- [12] <http://www.teachmeaudio.com/production/mixing/4-techniques/10-audio-spectrum>
- [13] <http://www.dspguide.com/ch22/1.htm>
- [14] <http://www.dspguide.com/ch22/1.htm>
- [15] <http://www.eurasip.org/Proceedings/Eusipco/Eusipco2010/Contents/papers/1569292685.pdf>