

# Evaluation and Personalization of Noise Reduction Algorithms in Digital Hearing Aids

G.Jayanthi<sup>1</sup> and Latha Parthiban<sup>2</sup>

<sup>1</sup>Department of Electronics Engineering, Sathyabama University, Tamilnadu.

<sup>2</sup>Department of Computer Science & Engineering, Pondicherry University Community College, Puducherry.

\*Corresponding author: E-Mail:lathaparthiban@yahoo.com.

## ABSTRACT

Hearing aids are prescribed to the people, whenever, there is a difficulty in understanding the speech, in the presence of competing background. The people seem to be easily distracted and would go for frequent requisition of repetition. Mainly if they have difficulty in understanding the speech in reverberant acoustic environment, they need to have a hearing aid. The objective of the present study is to develop an algorithm which identifies the frequency region in which the noise is separated out from the speech signal and preserves the speech audibility and improves the listening comfort of the hearing aid user in an affordable price.

**KEY WORDS:** Hearing loss, Hearing aids, occlusion effect, feedback effect, filter-banks, channels, bands, vents

## 1. INTRODUCTION

Hearing aids are prescribed to the people, whenever, there is a difficulty in understanding the speech, in the presence of competing background. The people seem to be easily distracted and would go for frequent requisition of repetition. Mainly if they have difficulty in understanding the speech in reverberant acoustic environment, they need to have a hearing loss testing (David Ayllon, 2014).

The Pure tone audiometry is the primary test, performed by the audiologist in a sound treated room. The degree of loss for a particular person is in terms of magnitude of stimulus needed for him to respond to it. The smallest intensity of a sound that a person needs to detect its presence is called his threshold for that sound. (Ying Wei, 2015)



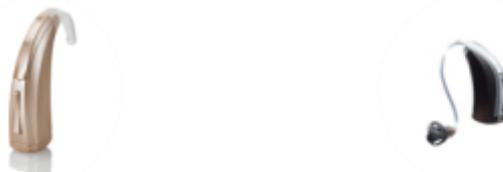
**Figure.1. Pure Tone Audiogram**

The test sounds used to determine the degree of hearing loss are usually pure tones of different frequencies at various intensities (Alexander Schasse, 2015). For example, when a patient has a hearing loss of 55dBHL, it means that his ear problem has caused his threshold to be elevated to the extent that it is 55dB higher than 0dBHL.

Test signals are presented and thresholds are obtained for both air-conduction and bone-conduction. Air-conduction testing involves presenting the test signals from standard audiometric (supraaural) earphones. Bone conduction is tested by applying a vibratory stimulus to the skull, which is transmitted to the cochlea. A comparison between the two sets of results for both ears, enable us to distinguish between different kinds of hearing losses (Jens Brehm Bagger Nielsen, 2015).

A chart is prepared, named pure tone audiogram as shown in Figure 1, provide the scope for the recognition of hearing loss. When the bone conduction is better than air conduction, then the patient is said to have conductive deafness (Umar, 2013). If the air conduction is better, then it leads to sensor neural deafness. If the problem persists in the outer ear and middle ear, then the loss leads to conductive or mixed hearing loss. If it is in the inner ear, then the loss is called sensor neural hearing loss.

Degrees of Hearing loss can be mild (25- 40dB), moderate (40 – 70dB), severe (70 – 90 dB) and profound (91 or more). Based on the degree and type of hearing loss (conductive, sensor neural or mixed), severity of hearing loss is diagnosed. Hearing aids are the assist devices, useful in hearing, when the patient has got mild, moderate, severe and profound hearing loss. Binaural ear fittings are more preferable, as it is used to get the differences in the information from both the ears (Nicolas Ellaham, 2013). Binaural systems provide the support for the improvement of speech intelligibility in noise; some of the other ear fittings in market are shown in figure 2:



**Figure.2. Hearing Aids (a) Behind-The-Ear (b) Receiver-In-Canal**

Behind-the-ear (BTE) hearing aids offer more flexibility and used for severe and profound hearing losses. It is considered as a classic one, as it contains all basic functions and much affordable in figure 2(a).

In-the-ear (ITE) can be fitted in the pinna of the ear. It is of larger in size, so that it can hold the battery for longer life.

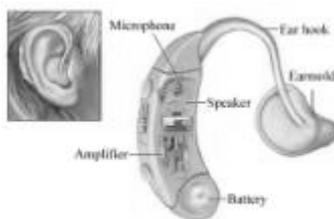
In-the-canal (ITC) is smaller in size and hence less noticeable. Persons having visibility issues can choose ITC. The additional features in ITC are directional microphones are included, in order to remove the undesired sound sources, in the various noisy environments. They have fixed directivity pattern, not adapted to directivity in the changes of acoustic environments (Soon-Suck Jarng, 2013).

Completely-in-canal (CIC) is kept completely inside the ear canal. Hence provide comfort and retention, since it is deeply seated in the ear canal. Hence the problem of adaptation and the notice ability of the device can be resolved.

**Problems in Hearing Aid:** Hearing aids provide assistance in hearing, still the user feel uncomfortable, while wearing it. Apart from the problem of adaptation, notice ability of the instrument or improper functioning of the device, they may also get other major problem termed occlusion or feedback effect. An occlusion effect occurs, when the hearing aid is worn continuously. The cartilaginous section of the ear canal is occluded by the HA device, but not when the bony portion is blocked. This can be used clinically help to determine whether a conductive impairment is present or not. (Uriz, 2013) In the normal persons, these vibrations will be escaped through an open ear canal. But, for the HA user, this becomes a burden while wearing it.

The second major problem in HA is the feedback effect. When the probe of the hearing aid is not fitting properly, then the user hears a squealing sound, apart from the amplified sounds. Normally, the input sound is amplified by the microphone, transmitted by the receiver, passes through the ear canal and reaches the ear drum (Parmod Kumar, 2009). The part of this signal is escaped through the ear probe of the hearing aid and again picked up by the microphone and sent it back to the ear canal. This process is repeated again and again, forming a feedback loop and presents the squealing sounds inside the ear. This feedback effect is mainly due to the passing of high frequency signals, which can be blocked by usage of ear moulds.

**Hearing Aid with Ear Moulds:** Ear moulds are the external fittings in the ear through which the hearing aid tubing is inserted. The impression of ear canal of the user is taken and an ear mould is prepared shown in figure 3. Hence while using the HA, the sounds escaping through the ear probe can be blocked and the problem of hearing squealing sounds can be eliminated.



**Figure.3.Hearing Aid with Ear mould**

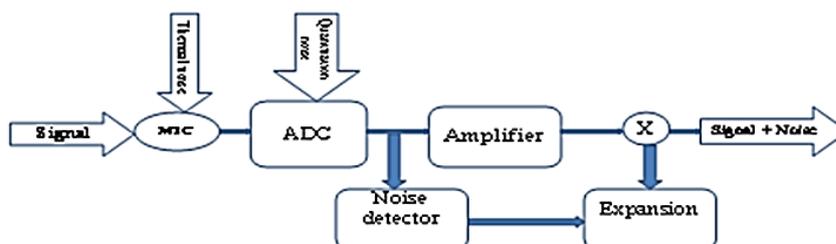
Ear moulds can be hard or soft moulds, among which the soft moulds are used for severe to profound hearing loss and the hard moulds, are used for mild, moderate and severe hearing loss (Seiji Nakagawa, 2013). Since ear moulds are completely shuttering the ear canal, it may leads to occlusion effect to HA user. To avoid that, vents are provided in the harder ear moulds. Vents are of small holes of size in the range of 1, 1.5, 2, 2.5mm, in order to pass the low frequency signals and to provide proper ventilation to the ear canal.

When an ear mold is used, the ear canal is blocked and hence the vibrations are reflected back toward the eardrum, which in turn, increases the loudness perception of their own voice. Nowadays, the Open fit hearing aids are introduced, which can be fitted without ear moulds (Jonathan, 2008). These open fit HAs are used to reduce the occlusion effect and comparatively smaller than BTE. The Receiver-in-canal (RIC) hearing aids are one of the open fit HA, shown in figure 2(b). In RIC, the receiver is separated from the housing and kept at the ear tip. Because of this, the size of the device is reduced and amplification of feedback signals can also be minimized.

**Noise in Hearing Aids:** The basic function of hearing system is to amplify the sounds selectively and then transfer the processed signal to the ear. Hearing aids are the most common form of treatment for mild to severe hearing loss. However, adaptation of hearing aids by patients is low (15-20%), due to relatively poor performance and the usability

of device. The main complaints of hearing aid users are: the sound qualities are poor and particular voices or sounds in the noisy environment are difficult to differentiate.

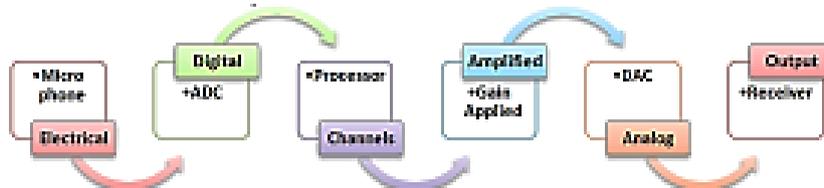
Usually the human brains receive sensory information from both the ears. As it is known, the sounds arrive at different times with various intensities, before sensing the multitude of sounds and voices in a crowded room, which can be achieved only with balanced hearing. (Apte, 2010) The binaural hearing helps in determining the origin of voice. While with a single hearing aid, if the sound comes into the ears are unbalanced, then the user may not be able to determine the location of the speaker immediately. Though the size and quality of the hearing aids are improved, still there exists a problem of background noises, comes with the voices to be heard. This may be due to noises from the environment, or due to feedback of signals or the noise created by the components itself, as shown in figure 4.



**Figure.4. Block diagram showing possible noise entry in the digital hearing aid**

Maximum all hearing aids using digital technology, in which a computer chip convert the incoming sounds into digital codes. It is then been analysed and adjusting the sound, to suit to particular person's hearing loss, listening needs and level of sounds around him. In the digital hearing aids, the analog sound signal is transformed into the digital signal by an ADC. This digital signal is divided into sub-band signals within different frequency banks by a filter bank. The amplified sub band signals are then synthesized and fed into DAC as shown in figure 5. Lots of paperwork has been done, in order to sweep out the possible noise entry. Some of the works have been shown.

Earlier uniform filter-banks have fixed sub bands, hence cannot provide flexible sound decomposition plans according to the characteristics of different types of hearing loss. The Lattice wave digital filter-banks (LWDFB) process the sound waves and DFT filter banks have multi-dimensional logarithmic number system. Both have reduced complexity than finite impulse (FIR) filter banks. But the above mentioned filter banks are not sensitive to coefficients and they have still fixed sub bands.



**Figure.5. Block diagram of sound transmission pathway**

## 2. MATERIALS AND METHODS

Design of reconfigurable digital FIR filter-bank for hearing aid, consists of multiband generation block and sub-band selection block. By changing the value of control signals, transfer function of filter bank can be changed and provides different sound decomposition methods. The output of first stage is manipulated to separate adjacent bands. Therefore, the complexity of masking stage is reduced greatly. But it is noted that the delay of this structure is relatively large for a real hearing aid system (Nathaniel, 1996).

Ying and Yinfeng proposed a flexible filter bank based on fractional interpolation, which provides the re-configurability of the number of sub bands and the locations of the sub bands. Multiple pass bands are generated after fractional interpolation and the pass bands are extracted by the masking filters which are obtained by the same prototype filter. In general, the more sub bands are used, the larger delay we get.

Alexander schasse, proposed a two-stage filter-bank system, so as to improve the frequency resolution, for the purpose of noise reduction. The first-stage hearing aid filter bank system can be used for compression and amplification, while the second stage filter- bank is for the lower frequency bands only and it resolves the harmonics of speech. Still it is unable to improve the noise reduction performance by adopting the number of first stage frequency channels to the signal, in terms of a voiced / unvoiced sound detection.

A novel sound source separation algorithm is used for binaural speech enhancement based on supervised machine learning and time-frequency masking. The algorithm considers the power restrictions in hearing aids, computational cost of algorithm and the transmission rate.

Selecting an algorithm, that uses fewer resources, to be implemented on a hearing aid device, based on a low cost DSP, is a challenging effort. However the noise methods require higher memory levels, which make the device size to be uncomfortable. The signal-to-noise ratio (SNR) is the measure of power of a desired signal to the power of the background noise. To improve the SNR of a signal, the signal is decomposed to obtain its Eigen values. From which, it is considered, the higher Eigen values related to desired signal while lower Eigen values are related to noise. Then the signal is reconstructed using highest Eigen values. Hence it has less noise than the original. The difficulty in obtaining the Eigen values is by Toeplitz matrix, which requires considerable number of instructions and a significant amount of memory, usually 58KB or more (Arne Leijon, 2009).

Calculating the SNR (signal to noise ratio) determines the presence or absence of speech and thereby determines the gain required. The outputs obtained, seem to have minimal distortions and appear clear enough for practical purposes. In the spectral subtraction method, the DFT of the input noisy speech samples are calculated. Subtracting the estimate of noise spectrum is obtained from the signal measured, when there is no presence of speech activity. It is noted that if the noise spectrum is estimated too steep, some musical noise is generated, on contrast, if it is estimated too low, the input noise speech is distorted. Therefore optimal parameters should be used for a varying noisy atmosphere (Arne Leijon, 2009).

In most of the speech enhancement algorithms, the magnitudes of STFT of signals are used, while phase is kept unchanged. According to 'Apte', the magnitude of STFT of noisy speech is kept unchanged while the phase is modified. Modified complex spectrum of speech is obtained by combining unchanged magnitude spectrum and modified phase spectrum. Because of this modification, low energy components (essentially noise) of complex spectrum are cancelled out, compared to high energy components (particularly speech), which leads to the reduction of background noise in hearing aids. This method of modelling performs best for the case of white noise, compared to train and babble noise (Joseph, 2013).

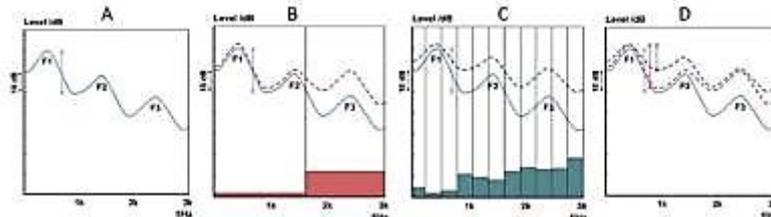
The phase inversion method provides ideal signal separation without suffering any biases at low signal-to-noise ratios. In non-linear hearing aids, distortion may be introduced unintentionally by imperfect transducers analog to digital converters and other circuitry. While evaluating the improvements due to noise reduction and compression in hearing aids, method of measurements using speech and noise mixtures can be used, in which the phase of the noise is reversed (phase inversion method). The recovered estimates are used to derive the equivalent speech and noise transfer functions, from which signal-to-noise ratio is derived. The problems in this evaluation is the comparison parameters, say compression ratio, threshold, attack and release times, number of channels and the hearing loss severity, are to be considered (Joseph, 2013).

The background noises are the noises which are distracting, confusing and simply irritating, are considered as unwanted noises. To reduce the additive noise, without impairing intelligibility, a new noise reduction method is developed based on wavelet method, by reducing correlated noise in noisy speech signals. The signal is decomposed into sub bands to give temporal and spectral resolution for the given application. Still wavelet method is useful in white noise, while for other applications, AR noise model is appropriate. Adaptive noise cancelling technique provides significant intelligibility gains, in turn; it decreases the presence of reverberation and subject head movement. Wiener filtering method increases the intelligibility for some of the hearing aid users, while reduces the intelligibility for all kind of normal hearing listeners. Spectral subtraction and sinusoidal modelling techniques remove both interfering noise and crucial high frequency cues. Therefore improves the SNR but in turn decreases the intelligibility. The single microphone approach for noise reduction methods, which is existing, are incapable of providing consistent improvements in intelligibility (Nathaniel, 1996).

The personalization of multi parameter hearing aids involves an initial fitting followed by a manual knowledge-based trial-and-error fine-tuning from ambiguous verbal user feedback. This often results in sub-optimal hearing aid setting, in which the full potential of modern hearing aid is not utilized. Jens Brehm, et al., proposed an interactive hearing-aid personalization system (IHAPS), to obtain an optimal individual setting of hearing aids from direct perceptual user feedback. The IHAPS is developed on flexible non-parametric Gaussian process model, to provide an efficient sequential design. A pitfall still occurs, if no perceptual difference exists for a large range of settings. Although there is a better chance to generalise the individual setting obtained with IHAPS, additional uncertainty is introduced (Jens Brehm Bagger Nielsen, 2015).

Hearing scientists believe that listening in complex acoustic environments requires the details of the acoustic waveform. The adaptation rate at which the amplification of sounds takes place, may determine the amount of acoustic information, passed to the auditory nerve. Also the best adaptation rate for a particular patient depends on the physiology of that given patient. Therefore, the fine structure information is also important the underlying physiology, hearing aid strategies and the resultant neural responses. Due to the sensor neural hearing loss (hearing loss due to noise exposure or aging), most of the impairment is of damaged outer hair cells, which in turn amplify low intensity sounds but apply less gain to high intensity sounds (Jonathan, 2008).

In hearing aid, the incoming voice signals are processed and different gains are applied for each channel. In every channel, suitable gain is applied, based on the audiogram of the person. This gain is chosen by the fitting formulae set in the system software. For example, if it is a 2- channel hearing aid, the full bandwidth of the audiogram is divided into two parts, considering one as low frequency channel and other as high frequency channel after which the suitable gains are applied to both the channels. If a person is having severe hearing loss in the high frequency region and mild loss in the low frequency region, then more gain is applied to the high frequency channel, in order to minimize the high frequency loss and relatively smaller gain to low frequency channel. Similarly, in multiple channel hearing aids, suitable gains are applied for each channel selectively, as shown in fig. 6



**Figure.6. Spectral smearing in multiple channels**

Apart from gain, variation in knee point, compression ratio can also be varied in each channel. Available number of channels can be of 2, 4, 6, 10, 12, 16 and 20 and even more. The channels from which the processed output signals are coming are called output channels. The channels through which the inputs of different frequencies are entering are called input channels or sensors. Different terminologies may be used by the different manufacturers. The sensors used in the microphone are to capture the incoming signals of different frequencies. Nearly 40 to 50 sensors can be used in a single microphone. Increase in channels increases the clarity of the signal, thereby increases the cost of hearing instrument.

Bands are used for fine tuning within the channels. Same or more number of bands are available, for the given number of channels. There are 6 channels – 6 bands, or 2 channel – 4 bands, 20 channels – 20 bands. There are some of the hearing systems available. Bands do not have compression ratio or knee point. Octave is the band of particular frequency and it can be lower octave or higher octave. If there is any variation in 500Hz and 1 kHz or 1 kHz and 2 kHz, then the mid-frequency of 750 Hz or 1.5 kHz respectively, is tested and it is termed as mid octave.

When the input-output is varying linearly, the output gets shifted and hence the compression ratio gets changed. The point, at which the compression ratio kicks in, is called Knee point. If the amplification of input is given linearly, to increase the loudness of the input signal, then the gain value in the output is also increased, to maintain the linearity. But, when the amplified output curve crosses the maximum power output, the signal gets clipped off and become flattened. The same time, much amplification in output signal, more than the maximum level, may deteriorate the voice signal and also affects the ear drum. Hence the output signal, before reaching the saturation point, diverted curvedly, in order avoid sharp peaks in the output, which in turn changes the compression ratio. The curved point or the shift in the curve due to gain, to avoid distortion, is termed as knee point. Hence at knee point, the gain gets changed, so the compression ratio is also varied.

If the person is not showing any response for high frequency sounds, it shows that the cochlea is not responding for the same. Then there is no point of amplifying the high frequency signals. It is sufficient to amplify and transmit the low and middle frequencies alone. If suppose, that person wishes to listen to high frequency sounds also, then the range of frequencies at higher level are compressed to his audible range, which is termed as frequency compression. That is, if 8 kHz signal is not audible to the person, then that signal is compressed to 6 kHz and hence modulated sound of 6 kHz is heard.



**Figure.7.(a) Electro Acoustic Characteristics (b) Hearing aid with coupler**

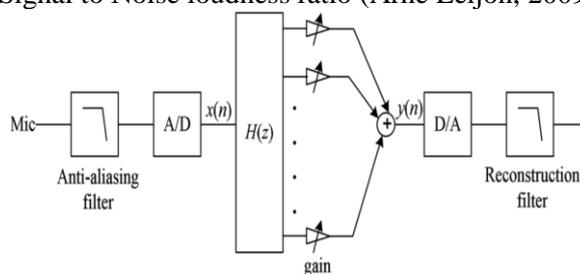
The hearing aid device has to be programmed, before the device should undergo prior checking of its Electro-Acoustic characteristics. The HA is connected one end with a coupler and other end with ear simulator. A coupler is used, as it provides the residual ear canal volume, while the ear simulator provides ear's variation of impedance with frequency. In the sound chamber, the hearing aid is checked under initial conditions, as shown in figure 7(b). Audiologist check and verify the initial conditions of hearing aid like Maximum Power Output (MPO), Full-On-Gain (FOG), Total Harmonic Distortion (THD), Equivalent Input Noise, Output Sound Pressure Level (OSPL), etc. and it is monitored through the graph as shown in figure 7(a) The device characteristics are verified based on specifications of ANSI S3.22-1987.

Once the electro-acoustic characteristics of hearing aid are measured, audiologists program the device, in which the suitable gain, compression ratio etc., can be adjusted individually for each patient. Apart from that, adaptation and the occlusion limits can also be varied.

Adaptation manager is used to provide the time limit given to the new user, to make him to adapt with the ear fittings. If a person is to be given with a particular amount of gain (target), instead of giving him the target at a stretch, he will be given with gain of target less 20. After 6 months, the gain may be raised to reach target less 10. Finally, after 3 months, the required target is reached. This is just to make him adapt with the device, as the sudden increase in volume may not be tolerated by the new user, particularly by children, persons having congenital hearing loss, etc. This time limit is been set by adaptation manager and once the time limit is reached, the device by its own can change the targeted gain automatically.

Occlusion manager is used to remove the occlusion effect. When the hearing aid is worn by the user, the speech signals are heard differently, due to the blocking of ear canal. Occlusion is the feeling of blocking of ears, by which their own voice itself heard like echoes. Each word sounds different, and could not be tolerated by the HA user. This can be reduced by the occlusion manager. Figure 8 shows the graph of frequency versus occlusion effect.

A cascaded digital filter or a multistage filter bank system can be applied in order to divide the digital signal in to number of sub bands within its bandwidth. A single channel shows no improvement in speech recognition, while improvement is shown in Signal to Noise loudness ratio (Arne Leijon, 2009).



**Figure.8.A schematic diagram for digital hearing aid**

Hence the frequency of the digital signal is split into multiple channels. The average level in each channel is calculated and the amplification coefficient is assigned and then the gain is calculated for each region, shown in fig. 9. The sum of the gains in each channel is measured and synthesized as a single digital signal. It is then passed through DAC, which converts the amplified digital signal into an audio signal and passed through the receiver.

### 3. RESULTS AND DISCUSSION

For all types of hearing aids, whether it is CIC, ITC, BTE or RIC and whether it is 2 channels, 20 channels or more, the hybrid used is similar, except the software to be used, which decides the availability of channels.

Cost of the device may be varied based on the size, number of channels available and the features it has. Soft computing techniques can be applied, when the solutions are unpredictable and uncertain and also it deals with imprecision, uncertainty, partial truth and approximation to achieve practicability, robustness and low solution cost. Soft computing is a technology to extract information from the process signal by using expert knowledge. (Parmod Kumar, 2009) It either seeks to replace a human to perform a control task or it borrows ideas from how biological systems solve problems and applies it to control processes. Soft computing has experienced an explosive growth in which Fuzzy logic is a multilevel logic system and the fuzzy logic set has a degree of membership associated with each variable. In neural networks, it deals with nonlinear mapping of objective problems, while neuro-fuzzy algorithm is a quantitative method of extracting the required information from the raw process signal (Parmod Kumar, 2009).

Hybrid soft computing systems can be provided with the combined use of genetic algorithm and fuzzy, for the irresolvable problems. GA can be hybridized with the solution searching ability of fuzzy systems, which provide better results in human subjective evaluation of hearing aids. In GA, the groups of chromosomes assigned as a single population and each group can be assigned with a task to be performed. Individuals are selected based on the fitness strategy. Fittest individuals undergo reproduction to produce offspring. By the process of crossover, the offspring exchange the genetic material for optimization. Single point crossover is mostly preferred. Even mutation can also be carried out, which will not allow the algorithm, to get stuck at the local minima. The process of crossover and / or mutation takes place, till a suitable solution is obtained. This will be carried out for 'n' number of generations, in order to get an optimum result. However most of the high-end hearing aids available in the market are of high cost targeted for the rich class. (Joseph, 2013) Binaural HA requires data transmission between both devices, which in turn increases the power consumption.

**4. CONCLUSION**

Person with conductive hearing loss may have great difficulty in hearing since they may get occlusion and feedback effect, which affect their speech intelligibility. Open fit hearing aids become very common, since hearing aid user can use their device without ear moulds. But prevention of secondary noise leakage cannot be blocked systematically. Proper refinement has to be done, as this may degrade the required speech and boost up the unwanted noise signals. More investigation in this field and efforts are to be taken for reducing the unknown noise leakage and complexity of algorithms with the help of GA.

**REFERENCES**

- Alexander Schasse, Timo Gerkmann, Rainer Martin, Wolfgang Sorgel, Thomas Pilgrim, and Henning Puder, *IEEE/ACM transactions on audio, speech, and language processing*, 23 (2), 2015, 383-393.
- Apte SD, Shridhar, *International Journal of Computer Applications*, 1(21), 2010, 44-51.
- Arne Leijon, *Workshop hearing screening and technology*, 2009
- David Ayllon, Roberto Gil-Pita and Manuel Rosa-Zurera, *EURASIP journal on advances in signal processing*, 47, 2014,
- Jens Brehm Bagger Nielsen, Jakob Nielsen, and Jan Larsen, *IEEE/ACM transactions on audio, speech, and language processing*, 23(1), 2015, 162 -172.
- Jonathan, Boley D, *Improving hearing aid dynamic gain control algorithms based on auditory nerve coding*, 2008.
- Joseph T, Oommen B C, R R Kumar R R, *IEEE Global Humanitarian Technology Conference, South Asia Satellite (GHTC-SAS)*, 2013, 44-47.
- Michael Swash, *Hutchison's Clinical Methods*, 19, ix, Bailliere Tindall & Cassell Ltd, London, 1989, 391-409.
- Nathaniel A, Whitmal, Janet C, Rutledge and Jonathan Cohen, *IEEE Engineering in Medicine and Biology*, 1996, 88-96.
- Nicolas Ellaham, Christian Giguere and Wail Gueaieb, *IEEE transactions on audio, speech and language processing*, 21(4), 2013, 879-888.
- Parmod Kumar and Alka Mahajan, *IEEE transactions on power delivery*, 24 (1), 2009, 452-461.
- Seiji Nakagawa, Chika Fujiyuki, Yuko Okubo, Takuya Hotehama and Takayuki Kagomiya, *35<sup>th</sup> Annual international conference of the IEEE EMBS*, 2013.
- Soon-Suck Jarng, Carl Swanson, Frank Lee, Joseph Zou, *CES-CUBE 2013, ASTL*, 2013, 25, 261-266.
- Umar M, Subbulakshmi N, *International Journal of VLSI and Embedded Systems-IJVES*, 2013, 4, 696.
- Uriz A.J, Agüero P.D, Moreira J.C, Tulli J.C and Gonzalez E.L, *IEEE latin America Transactions*, 11(1), 2013, 224-229.
- Ying Wei and Yinfeng Wang, *IEEE/ACM Transactions on Audio, Speech, and Language Processing*, 23 (5), 2015, 923-931.