DIGITAL HEARING AID DESIGN USING LOW COMPLEXITY FILTERS

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ABSTRACT

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For convenience reasons, hearing aids are getting smaller in size everyday. This requires smaller hardware and efficient software designs for digital hearing aids. The main component in hearing aid software is the signal processing part. To meet stringent requirements on size and energy, the signal processing techniques used in digital hearing aids should have low-computational complexity.

In this thesis, a comparative performance analysis for different low-complexity filter design techniques used in digital hearing aids is presented. The following low-complexity filter design method is investigated:

- Low-complexity coefficients

and utilized to design filters to be used in digital hearing aids. Simulation results suggest that filter coefficients have whose complexities are reduced using the method in provides comparable response characteristics to filters whose complexities are not reduced.

Key words: Digital hearing aid design, low complexity filters, filter coefficients

ÖZ

DÜŞÜK HESAP YÜKLÜ SÜZGEÇLER KULLANARAK SAYISAL İŞİTME CİHAZI TASARIMI

OĞUZ, Tuğba

Yüksek Lisans, Elektronik ve Haberleşme Anabilim Dalı

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Kolaylık nedenlerinden dolayı, dijital işitme cihazları her geçen gün boyut olarak küçülmektedir. Bu durum dijital işitme cihazları için küçük donanıma ve etkili bir yazılıma ihtiyaç duymaktadır. İşitme cihazında esas kısım sinyal işleme bölümüdür. Enerji ve boyuttaki bu zor gereksinimlerin karşılanması için, dijital işitme cihazlarında kullanılan sinyal işleme teknikleri düşük hesap yüklü olmalıdır.

Bu tezde, dijital işitme cihazlarında kullanılan düşük hesap yüklü filtre tasarım tekniklerinin karşılaştırmalı performans analizi gösterilecektir. Aşağıda yer alan düşük hesap yüklü filtre tasarım tekniği incelenecektir.

- Düşük hesaplanabilir katsayılar

Anahtar Kelimeler: Dijital işitme cihazı tasarımı, düşük hesap yüklü filtreler, filtre katsayıları

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LIST OF ABBREVIATIONS

LMS	Least Mean Square
FIR	Finite Impulse Response
IIR	Infinite Impulse Response
FRM	Frequency Response Masking

CHAPTER 1

INTRODUCTION

1.1 Background

Hearing aids are getting smaller in size everyday. To meet this size constraint, even smaller hardware and more efficient software are required. In this section, reasons of hearing loss, types of hearing loss and general knowledge of hearing aid types are presented. Reasons for hearing losses may be permanent or temporary. Some of them are genetics, aging, exposure to noise, illness, chemicals and physical trauma. Hearing aids are used to help people who suffer from hearing impairment. Its working principle is same as the human ear working. A microphone picks up sound and converts it into an electrical signal and then it sends to amplifier. An amplifier increases sound level and it sends it to the receiver. A receiver converts from electrical signal to sound and then it sends it into the ear.

1.2 Types of Hearing Losses

Hearing losses are classified by their type and their severity. Moreover, hearing losses may exist in only one ear (unliteral) or in both ears (bilateral). There are three main types of hearing losses.

- Conductive hearing loss
- Sensorineural hearing loss
- Mixed hearing loss

1.2.1 Conductive Hearing Loss

Conductive hearing losses happen when sound is not managed through the outer ear canal to the ear drum and tiny bones of the middle ear. Conductive hearing loss usually involves in sound level or the ability to hear faint sounds [1].

1.2.2 Sensorineural Hearing Loss

Sensorineural hearing loss occurs when there is damage to their inner ear (cochlea) ,or to the nerve pathways from the inner ear to the brain [2].

1.2.3 Mixed Hearing Loss

Sometimes a conductive hearing loss occurs in combination with a sensorineural hearing loss. In other words, there may be damage in the outer or middle ear and in the inner era (cochlea) or auditory nerve. When this occurs, the hearing loss i referred to as a mixed hearing loss [3].

1.3 Types of Hearing Aids

Hearing aids are categorized based on their functionality and placement.

1.3.1. Functional Categorization

- Analog hearing aid
- Programmable hearing aid
- Digital hearing aid

Analog hearing aids make continuous sound waves louder. These hearing aids essentially amplify all sounds (e.g.speech and noise) in the same way. Programmable hearing aids have a microchip which allows the aid to have settings programmed for different listening environments, such as a quiet place like at a library, or a noisy place like in a restaurant, or a large area like a soccer field. The analog programmable hearing aids can store multiple programs for various environments. Digital hearing aids have all the features of analog programmable aids, but they convert sound waves into digital signals and produce an exact duplication of sound. Computer chips in digital hearing aids analyze speech and other environmental sounds. The digital hearing aids allow for more complex processing of sound during the amplification process which may improve their performance in certain situations [4].

1.3.2. Placement Categorization

Hearing aids could also be categorized on the basis of how they are placed on the human body [5].

- Completely in the Canal (CIC)
- In the Canal (ITC)
- Behind the Ear(BTE)
- Body Worn Aids
- Bone Anchored Hearing Aid (BAHA)

Completely in the Canal (CIC): As the name suggests they are completely in the ear canal. They are not visible from outside and are for people who don't want other people to see that they are suffering from some kind of hearing loss. They are very cost around USD 250.

In the Canal (ITC): It is hardly visible from outside and is also a costly choice. It is good for conductive hearing losses.

Behind The Ear (BTE): These kinds of hearing aids rest behind the ear. They are not costly and they are visible to other people.

Body Worn aids: They are most simple and bigger in size. They are the least expensive. These hearing aids consists of a case, cord and an ear mold.

Bone Anchored Hearing Aid: Another type of hearing aid is the bone anchored hearing aid. It is implanted into the skull through operation.This kind of hearing aid can treate people having conductive loss. This kind of hearing aid transmits sound directly to the hearing nerve [5].

1.4 AUDIOGRAM

An audiogram is a graph that shows the audible threshold for standardized frequencies as measured by an audiometer [6]. An audiogram shows the quietest sounds you can just hear. An audiogram can be obtained using a behavioral hearing test called *Audiometry*. Audiograms are set out with frequency in hertz (Hz) on the horizontal axis,

most commonly on a logarithmic scale, and a linear dB scale on the vertical axis [6]. Figure 1 shows a typical audiogram for the normal hearing.

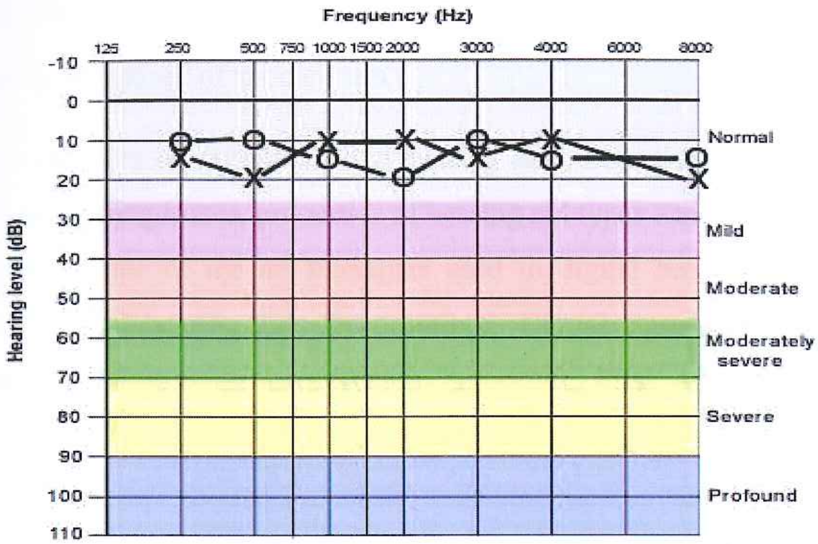

Figure 1 A typical audiogram for the normal hearing. 'O' and 'X' represent the thresholds of left and right ear, respectively

1.4.1. Hearing Loss Ranges

Hearing loss ranges are betwen 26 dB to 91 dB or greater. Normal hearing is when the softest sounds heard are between -10 and 20 dB. If the sounds are louder than 20 dB them and you still can't hear them, there is a hearing loss. Ranges of hearing loss are presented as follows:

mild -> adults -> 26-40 dB

children ->20-40 dB

Moderate -> 41 – 54 dB

Moderately Severe -> 55 – 70 dB

Severe -> 71- 90 dB

Profound -> 91 dB or greater

Totally deaf -> have no hearing at all.

1.5. Organization of the Thesis

The thesis is composed of six chapters.

Chapter 1 includes an introduction to hearing aids, objectives of this thesis, causes and types of hearing losses, properties of hearing aid types and audiogram.

In Chapter 2, state of the art tecniques used in digital hearing aids, fundamental concept of digital hearing aids are described. Digital signal processing tehcniques which are adaptive noise cancellation algorithm, least mean square algorithm, frequency shifting, controlled amplification and power compressiom methods are explained.

Chapter 3, Filter structures using digital hearing aid design are explained in Chapter 3. Infinite Impulse Response filters, Finite Impulse Response filters, frequency response masking technique, uniform and non-uniform filter banks are presented.

Chapter 4, To find a low complexity coefficients, we will use low complexity method which is proposed by Boyd et al: '*Filter Design with Low Complexity Coefficients, J. Skaf and S. Boyd,* July 2008.'Boyd 's.

Chapter 5 includes simulation results of current based techniques using in digital filters and low complexity method in MATLAB. Then, we compare simulation results according to our audiograms.

Chapter 6 includes conclusion parts.

CHAPTER 2

CURRENT TECHNIQUES FOR DIGITAL HEARING AIDS

2. 1. Fundamental Concept of the Digital Hearing Aids

Digital hearing aids contain very high computational complexity algorithms and filters. They also contain different transforms for time to frequency domain conversion. In all hearing devices one of the basic purposes is to amplify sound in specific frequency bands of interest. There are some well known techniques, the frequencies are split into different bands.These techniques are adaptive noise cancellation, controlled amplification, frequency shifting and power compression.

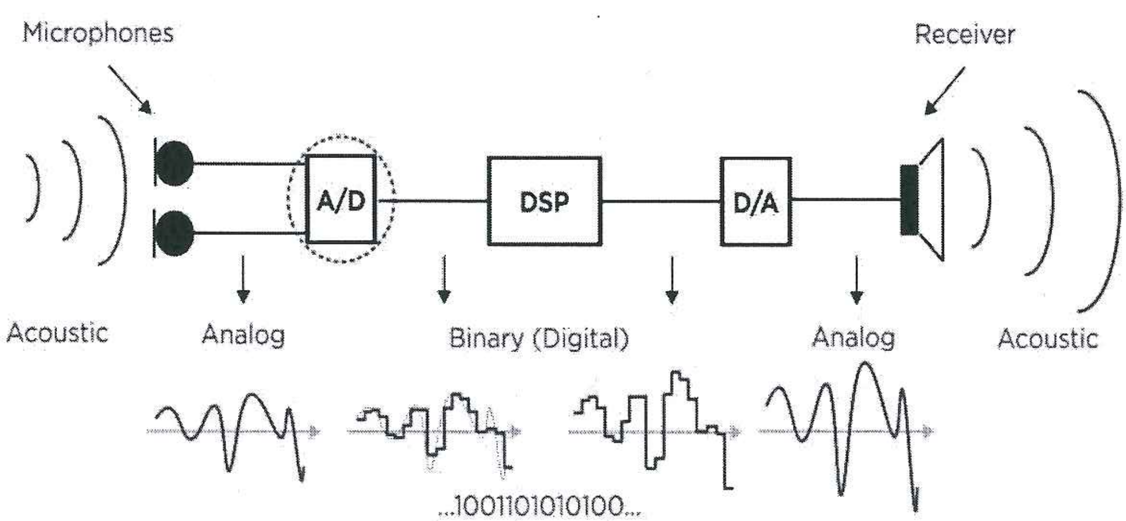

Figure 2 Block Diagram of Digital Hearing Aid

2.1. 1.Adaptive Noise Cancellation Algorithm

The basic idea is to pass the corrupted signal through a filter that tends to supress the noise while leaving the signal unchanged. This is an adaptive process which means it doesn't require a priori knowledge of signal or noise characteristics. Adaptive noise cancellation efficiently attenuates low frequency noise for which passive methods are ineffective [7].

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 provides the system to adjust to the changing statistical conditions of the environment unlike digital filters. Therefore, adaptive filters are preferred over digital filters.

2.1.2 Least Mean Square (LMS) Algorithm

Least mean squares (LMS) algorithms are a class of adaptive filter used to mimic a desired filter by finding the filter coefficients that relate to producing the least mean squares of the error signal (difference between the desired and the actual signal) [8].

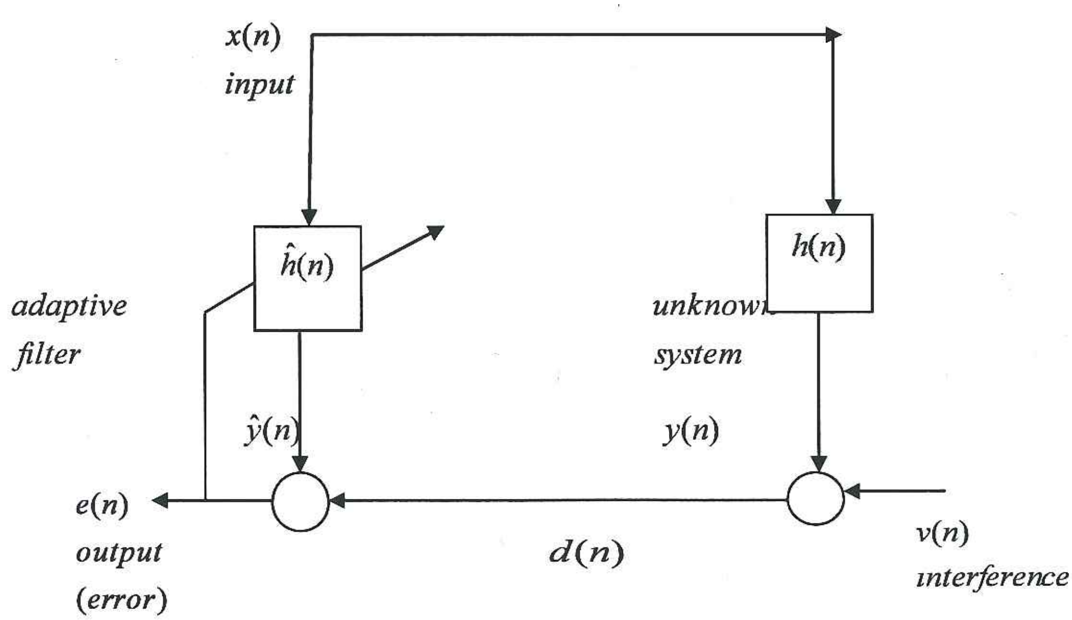

Figure 3 LMS Algorithm Block Diagram

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

$$0 < \mu < 1/\mu_{\max} \tag{2.1}$$

where μ is the convergence factor of correlation matrix and $\mu_{\max}$ is the largest eigenvalue of the correlation matrix.

The algorihm performs the following steps repeated.

The filtering operation

$$y(n) = w^{H}(n).x(n) \tag{2.2}$$

The error calculation

$$e(n) = d(n) - \hat{y}(n) \tag{2.3}$$

The coefficient correction

$$w(n+1) = w(n) + \mu.x(n).e(n) \tag{2.4}$$

If the eigenvalues are widespread, then convergence may be slow. The values are calculated by determining the ratio of the largest eigenvalue to smallest eigen value of the matrix. 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 coefficient [9].

2.1.3 Frequency Shifting

There are many people who have a hearing loss in a particular frequency range. Amplification algorithms usually are cabaple of increasing the loudness of sound to the required level. However, in certain cases where there is an extensive damage, even amplification can't provide required loudness. There are two common types of hearing loss. These 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). Meniere's disease causes sensorineural hearing loss in the low frequency range (125 Hz to 1000Hz) [9].

Frequency shifting shifts the signal from the frequency range with extensive loss to the range where the user can hear normally. For the Presbycusis, the frequencies from 4000 Hz to 8000 Hz are shifted to lower frequencies, whereas in the case of Meniere's disease, the frequencies from 125 Hz to 1000 Hz are shifted to higher frequencies.

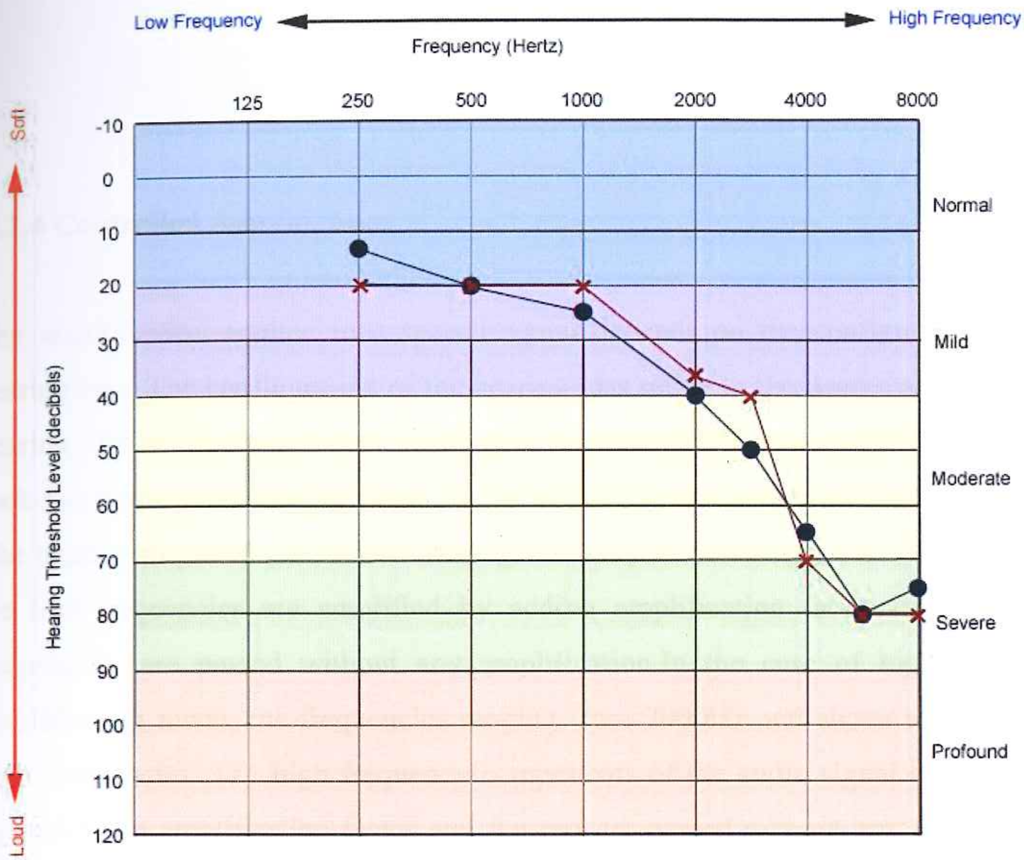

Figure 4 Audiogram example for Presbycusis hearing loss

2.1.4 Controlled Amplification

The amplification applied to a speech signal depends on the configuration of the hearing loss. The configuration of the hearing loss refers to the degree and pattern of hearing loss across frequencies. If we give an example for low amplification mode, such the frequencies ranging from 20 Hz to 500 Hz are termed as low frequencies. The hearing impaired person has hearing loss only in this range. For that user, only the low frequencies are amplified by adding amplification factor.The remaining frequencies are passed without any amplification.In the case of high frequnecy amplification mode, the frequencies ranging from 2000Hz and above are termed as high frequencies .The high frequency components of the audio signal are amplified by adding an amplification factor and the rest are passed without any amplification [9].

2.1.5 Power Compression

In digital hearing aid design, one of the basic purpose is to reduce power consumption and provide long 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 [10].

A lower level threshold (Plow) in dB is set, such that all the frequency components having a power level less than Plow are attenuated and brought down to 10% of its original value. Also, an upper power limit (Psat) in dB is set, such that any frequency having power level greater than Psat are made equal to Psat. The values of Plow and Psat can be set as per the requirement of the system [8]. In the case of low frequency mode, 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. For 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) [10].

CHAPTER 3

FILTERS IN DIGITAL HEARING AIDS

An ideal digital hearing device should include such as low power, small area and better timing. If there were no size and power constraints, filter structures would be computationally demanding. Without limitation of the power to increase the number of processors used in the filter design. These results are unwanted things in the filter design. In this section, we will mention about filter structures using in digital hearing aid design. Filter structures which are used in digital hearing aid as following;

- Infinite Impulse Response (IIR) Filter
- Finite Impulse Response (FIR) Filter
- Frequency Response Masking (FRM)
- Uniform and non-uniform Filter Banks

3.1. Infinite Impulse Response (IIR) Filter

Infinite impulse response filters are digital filters with infinite impulse response. Unlike FIR filters, they have the feedback (a recursive part of a filter) and are known as recursive digital filters. Because of this reason, IIR filters have much better frequency response than FIR filters of the same order. Their phase characteristic is not linear which can cause a problem to the systems which need phase linearity. For this reason, it is not preferable to use IIR filters in digital signal processing when the phase is of the essence. IIR filters can be designed using different methods. One of the most commonly used is via the reference analog prototype filter [11].

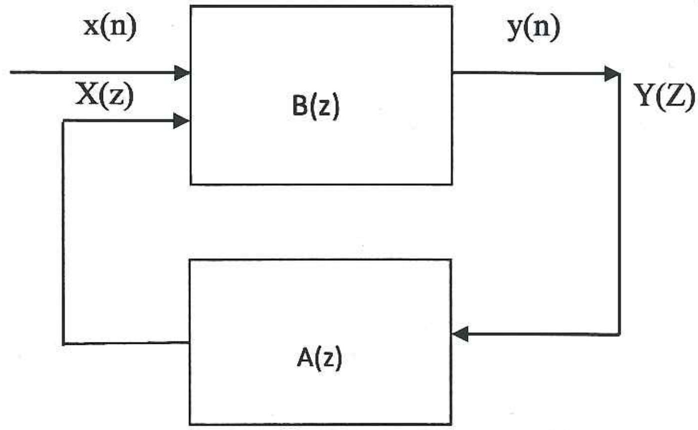

Figure 5 Block Diagram of IIR Filter

The most commonly used IIR filter method is reference analog prototype filter. It is the best method for designing standard filters such as lowpass, highpass, bandpass and bandstop filters. The filter design starts with specifications and requirements of the acceptable filter. In digital hearing aid design, we will describe frequency range of filter according to patient's hearing loss range.

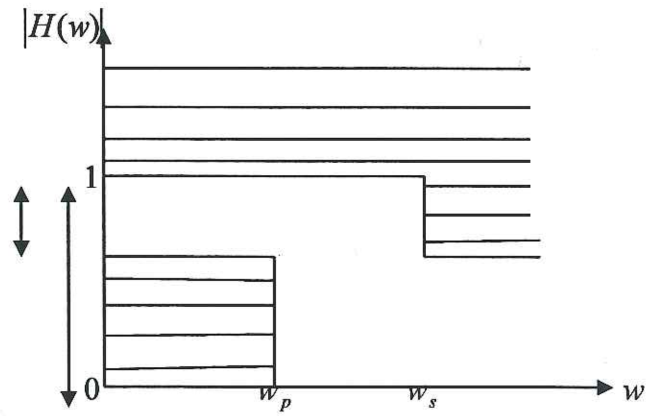

Figure 6 Low pass digital filter specification

w_p - normalized passband cut -off frequency

w_s - normalized stopband cut- off frequency

δ_1 - maximum passband ripple

δ_2 - maximum stopband ripple

ε - passband attenuation parameter

The general equation is defined as follows,

$$H(z) = \frac{b_0 + b_1.z^{-1} + ... + b_n.z^{-n}}{1 + a_1.z^{-1} + ...a_m.z^{-m}} = \frac{\sum_{k=0}^{N} b_k.z^{-k}}{1 + \sum_{k=1}^{M} a_k.z^{-k}} \tag{3.1}$$

where a_k and b_k are filter coefficients.

The transfer function of IIR filter is:

$$H(z) = k\frac{(z - z_1)(z - z_2).......(z - z_N)}{(z - p_1)(z - p_2)..(z - p_N)} \tag{3.2}$$

For the implementation of the above equation, we need to difference equation:

$$y[n] = \sum_{k=0}^{\infty} h[k].x[n-k] = \sum_{k=0}^{N} b_k.x[n-k] - \sum_{k=1}^{M} a_k.x[n-k] \tag{3.3}$$

$$y[n] = \sum_{k=0}^{N} b[k].x[n-k] + \sum_{k=1}^{M} a[k].y[n-k] \qquad \text{IIR Equation} \tag{3.4}$$

3.1.1. IIR Filter in direct form I or II:

The transfer function of IIR Filter is;

$$H(z) = \frac{Y(z)}{X(z)} = \frac{\sum_{k=0}^{N} b_k.z^{-k}}{1+\sum_{k=1}^{M} a_k.z^{-k}} = \frac{b_0 + b_1.z^{-1} + ...b_N.z^{-N}}{1 + a_1.z^{-1} + ...a_M.z^{-M}} \quad (3.5)$$

where b_k and a_k are filter coefficients.

3.1.2 IIR Filter in cascade form:

The transfer function of IIR Filter is written as a product of second-order sections;

$$H(z) = \prod_{l=1}^{L} \frac{b_{0l} + b_{1l}.z^{-1} + b_{2l}.z^{-2}}{1 + a_{1l}.z^{-1} + a_{2l}.z^{-2}} \quad (3.6)$$

3.1.3 IIR Filter in parallel form:

The transfer function is written as a sum of second-order sections

$$H(z) = \sum_{l=1}^{L} \frac{b_{0l} + b_{1l}.z^{-1}}{1 + a_{1l}.z^{-1} + a_{2l}.z^{-2}} \quad (3.7)$$

3.2 Finite Impulse Response (FIR) Filter

FIR filter is a filter whose impulse response is of finite duration. It is also known as non-recursive digital filter.

$$y(n) = \sum_{k=0}^{M-1} b_k x(n-k) = \sum_{k=0}^{M-1} h(k).x(n-k) \qquad (3.8)$$

h(n) and x(n) are finite duration sequences, their convolution is also finite in duration. The duration of sequence y (n) is L+M-1.

FIR filters can be designed using different methods, but most of them are based on ideal filter approximation. The objective is to achieve sufficiently good characteristics of a filter. The transfer function of FIR filter approaches the ideal as the filter order increases, thus increasing the complexity and amount of time needed for processing input samples of a signal being filtered [12]. FIR filter design methods are based on ideal filter approximation. The resulting filter approximates the ideal characteristics as the filter order increases.

3.2.1 FIR Filter Specification

The filter design process starts with specifications and requirements of the desirable FIR filter. Which method is to be used in the filter design process depends on the filter specifications and implementation. Each method has advantages and disadvantages. Due to its simplicity and efficiency, the window method is most commonly used method for filter design. The frequency sampling method is easy to use, but filters designed this way, have small attenuation in the stopband. Therefore, choosing a right method for FIR filter design is very important [12]. For digital hearing aid design, the amplitude response of the filter should match the audiogram. Frequency range of the filter should adjust according to patient's hearing loss range. In this thesis, we will compare results of these methods for matching the audiogram.

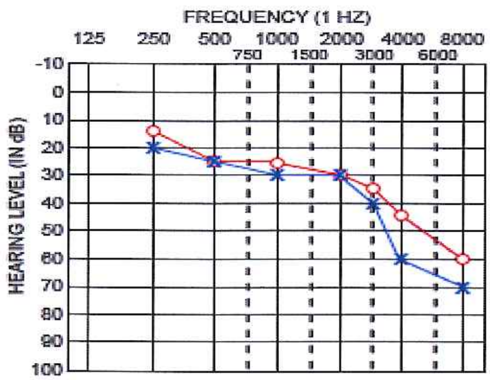

Figure 7 Mild Hearing Loss

3.2.2 FIR Filter Design Methods

3.2.2.1Windowing Method

The windowing method is most commonly used in FIR filter design. A window is a finite array consisting of coefficients selected to satisfy the desirable requirements. When designing FIR filters, we should determine which window function is to be used and the filter order according to the required specifications. In digital hearing aid design, our filter specification can adjust according to patient's hearing loss range.

$$y[n] = \sum_{k=0}^{N-1} x[k].h[n-k] \tag{3.9}$$

where

x[k] are FIR filter input samples;

h[k] are the coefficients of FIR filter frequency response; and

y[n] are FIR filter output samples.

The z transform of a N point FIR filter is given by

$$H(z)=\sum_{n=0}^{N-1}h(n).z^{-n} \tag{3.10}$$

In windowing method, the desired frequency response [13] $H_d(w)$ corresponding unit sample response $h_d(n)$ is determined by following equation :

$$h_d(n)=\frac{1}{2\pi}\int_{-\pi}^{\pi}H_d(w).e^{jwn}d_w \tag{3.11}$$

where

$$H_d(w)=\sum_{n-=\infty}^{\infty}h_d(n).e^{-jwn} \tag{3.12}$$

In general, unit sample response $h_d(n)$ must be truncated at some point say n= M-1 to yield an FIR filter of length M (i.e. 0 to M-1). This truncation of $h_d(n)$ to length M-1 is same as multiplying $h_d(n)$ by the rectangular window defined as

$$\begin{aligned} w(n) &= 1, 0 \le n \le M-1 \\ w(n) &= 0, otherwise \end{aligned} \tag{3.13}$$

Thus the unit sample response of the FIR filter becomes

$$h(n)=h_d(n).w(n) \tag{3.14}$$

$W(w)$ is the frequency domain representation of the window function

$$W(w)=\sum_{n=0}^{M-1}w(n).e^{-jwn} \tag{3.15}$$

3.2.2.1 a) Hamming Window Function

The hamming window is a raised cosine window. The hamming window exhibits similar characteristics to the Hanning window but further suppress the first side lobe. The equation for Hamming window sequence can be defined by

$$w(n) = \alpha - \beta.\cos(\frac{2\pi n}{N-1}) \tag{3.16}$$

for $-\frac{N-1}{2} \leq n \leq \frac{N-1}{2}$ with $\alpha = 0.54$, $\beta = 1 - \alpha = 0.46$

The non-causal hamming window function is related to the rectangular window function;

$$w_H(n) = w_R(n).[0.54 + 0.46.\cos(\frac{2\pi n}{N-1}) \tag{3.17}$$

3.2.2.1 b) Hanning Window Function

The Hanning window is a raised cosine window and can be used to reduce the side lobes while preserving a good frequency resolution compared to the rectangular window.

The window function of a causal Hanning window is given by

$$w_{hann}(n) = 0.5 - 0.5.\cos(\frac{2\pi n}{N-1}) \quad 0 \le n \le N-1 \tag{3.18}$$

$$0, otherwise$$

The window function of a non-causal Hanning window is given by

$$w_{hann}(n) = 0.5 + 0.5.\cos(\frac{2\pi n}{N-1}) \quad 0 \le n \le \frac{N-1}{2} \tag{3.19}$$

$$0, otherwise$$

3.2.2.1 c) Kaiser Window Function

This windowing method is designed to generate a sharp central peak. Although it is computationally involved, its coefficients are easy to generate largely as a result of helpful design rules that eliminate iterations typically needed to achieve desired ripple and transition specifications [14].

The Kaiser windowing method is defined by the following equation using the windowing parameter, α :

$$w[n] = \frac{bessel[\alpha\sqrt{\frac{2n}{N-1}\left(2-\frac{2n}{N-1}\right)}]}{bessel[\alpha]} \tag{3.20}$$

3.2.2.2 Frequency Sampling Method

In the frequency sampling method, the frequency response of the FIR filter is specified in terms of samples of the desired frequency response. The frequency sampling method allows us to design recursive and non-recursive FIR filters for both standard frequency selective and filters with arbitrary frequency response [15].

Sampling the continuous frequency response $H_d(w)$ at N points essentially gives us the N-point DFT of $H_d(\frac{2\pi kn}{N})$ [16].

Thus by using the IDFT formula, the filter coefficients can be calculated using the following formula

$$h(n) = \frac{1}{n}\sum_{n=0}^{N-1}(H(k).e^{\left(\frac{j2\pi n}{N}\right)k}) \tag{3.21}$$

where

H(k),k=0,1,.......,N-1.

3.3 Frequency Response Masking Technique

Frequency response masking (FRM) technique reduces number of multipliers by using several subfilters of lower order and combining them to get the final filter. The conventional FRM approach concerns the design of low pass FIR digital filters in terms of a pair of interpolated band edge shaping and a corresponding pair of low pass masking digital sub-filters [17].

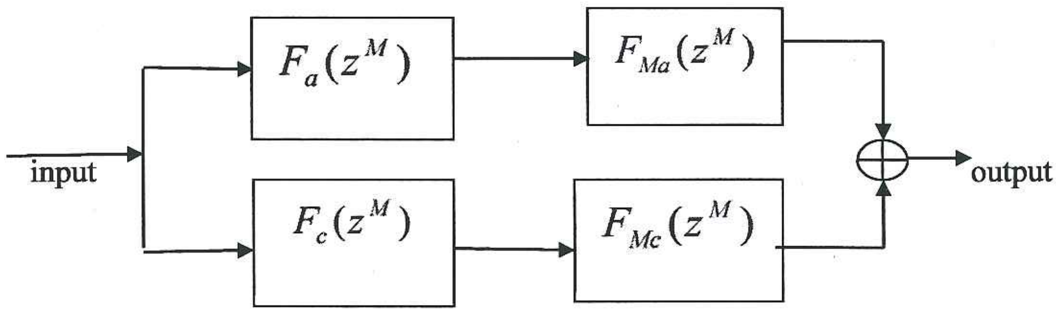

Figure 8 Basic block diagram of the overall FRM filter

Block diagram of FRM approach starts with a prototype filter $F_a(z)$ of passband edge frequency Θ, stopband edge frequency Φ and with transition width Δ .The narrow transition width of FRM results from the interpolated version of the prototype filter $F_a(z^M)$ derived by replacing each delay element of $F_a(z)$ by M delay elements and $F_c(z^M)$ is its complementary version obtained by subtracting the output of $F_a(z^M)$ from a suitably delayed version of the input [19].

$$F_c(z^M) = z^{(\frac{-M(Na-1)}{2})} - F_a(z^M) \tag{3.22}$$

Where Na is the the filter length of $F_a(z)$.There are two parallel branches each of which is composed of an interpolated model filter in cascade with masking filters $F_{Ma}(z)$ and $F_{Mc}(z)$respectively. Masking filters are used to select the useful part of $F_a(z^M)$and $F_c(z^M)$. Addition of two masked responses gives the response of a sharp wideband FIR filter [18].

3.4 Uniform and Non-Uniform Filter Bank

Filter banks are group of low pass, high pass, bandpass filters. Filter banks are used in many signal processing applications such as audio and image processing. Filter banks are part of a group of signal processing techniques that decompose signals into frequency subbands [19].Filter banks supply lower latency and better frequency compensation in digital hearing aids. They also provide reliable automatic speech recognition.

All the filters of uniform filter banks have equal bandwidth. All filters use a common input signal sampled at the same frequency [20]. Their architecture is very simple. It is also very easy to implement in hardware. The bands in uniform filter banks are uniformly distributed since it doesn't provide sufficient resolution at low frequencies for the logarithmic behaviour of hearing [20].

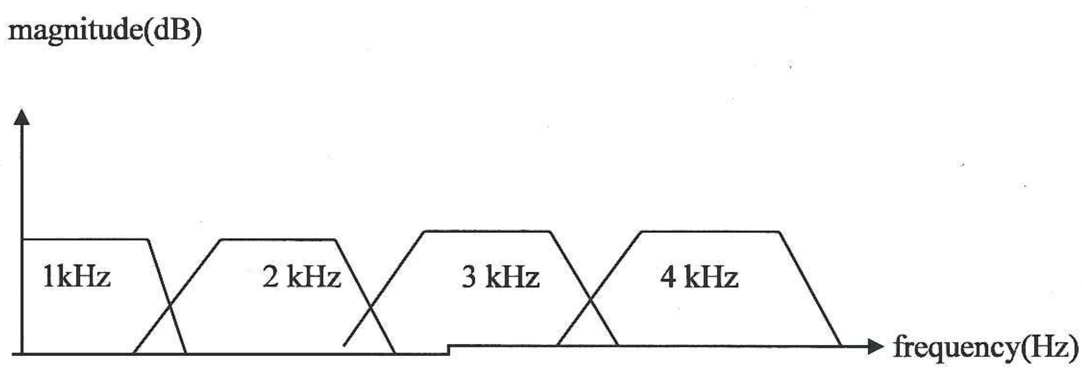

Figure 9 Uniform Filter Bank

Figure 10 shows the basic idea of band division. Sampling frequency is 8 kHz and there are 4 filter banks each with a bandwidth of 1 kHz. Increasing number of the bands will fix the resolution problem at low frequencies but will give unnecessary high resolution at high frequencies [20].

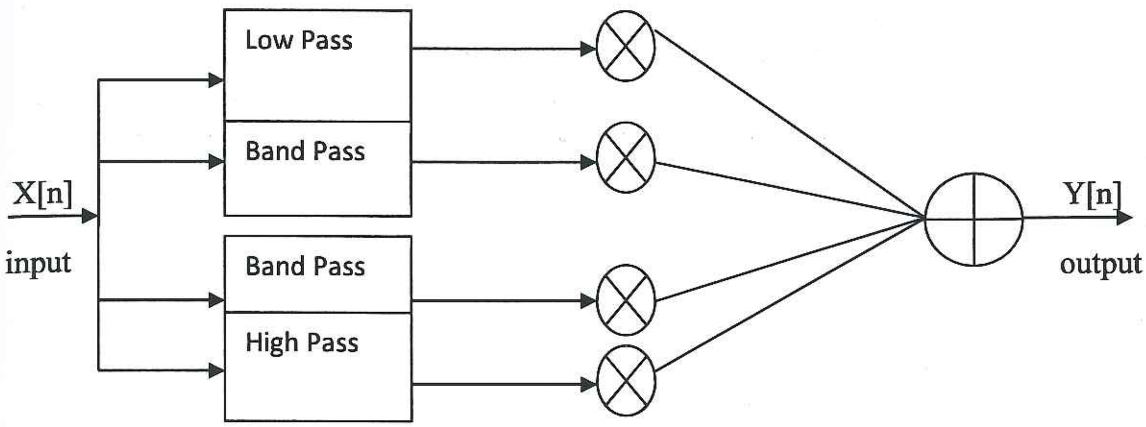

Figure 10 Filter Bank Architecture

When the number of coefficients is gretaer in number, frequency response wil be better. However, this situation causes increasing the latency such as delay from input to output. Acceptable latency of human ear is 12ms.

Non-uniform filter bank is same as the uniform filterbank but only filter cut off frequency changes. Due to working logarithmically of non uniform filter banks, their behavior like the human ear. This approach provides better resolution at lower frequencies.

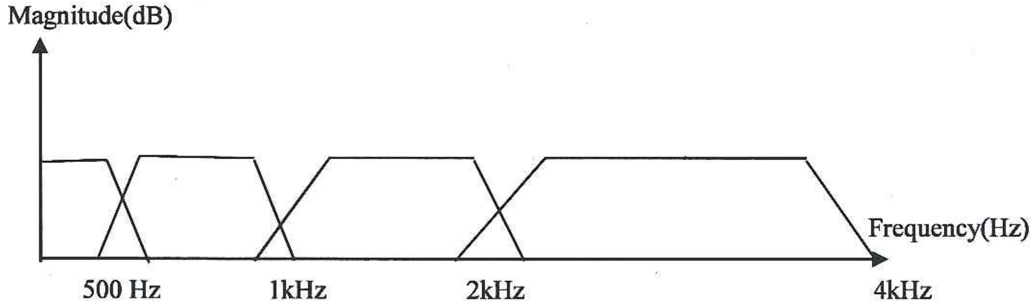

Figure 11 Non- Uniform Filter Bank

CHAPTER 4

FILTER DESIGN WITH LOW COMPLEXITY COEFFICIENTS

In this thesis, we try to design filter for digital hearing aid with low power, small area and low computational complexity. To do this, our filter coefficients should have low complexity. In this chapter, we aim to design filter whose coefficients have low complexity. To find a low complexity coefficients, we will use low complexity method which is proposed by Boyd et al: '*Filter Design with Low Complexity Coefficients, J. Skaf and S. Boyd,* July 2008.'Boyd 's.

In this method, we design a discrete time filter whose transfer function is H_{β} : $C \to C$ where **C** is set of complex number set. We also define $\beta \in R^{p}$ is the vector of real coefficients. β is the coefficients of the filter. For Infinite Impulse Response (IIR) filter in direct form, H_{β} is our filter's tranfer function with β_i is the coefficients in its numerator and denominator.

Firstly,we define a nominal filter design without any filter specifications. Its coefficient vector is β_{nom}. H is set of acceptable filter designs which is set of the transfer functions.It shows that our filter's performance requirements. C is the set of coefficinet vectors according to acceptable filter designs.

$$C = \left\{\beta \in .R^{p} \middle| H_{\beta} \in \mathrm{H}\right\} \tag{4.1}$$

In this filter design nominal filter match performance specification. Our aims to find β has lowest complexity.Φ: $R^{P} \in R$ is the complexity vector of filter coefficients.

$$\Phi(\beta) = \sum_{i=1}^{p} \phi(\beta_i) \tag{4.2}$$

where $\phi(\beta_i)$ is the complexity of i^{th} coefficient of β.

In this method, we deal with a complexity. This is $\phi_{bin}(\beta_i)$ which is the number of the 1s in the binary expansion of the coefficient β_i.

In this method, we define a randomized heuristic algorithm for our filter design. We start with nominal filter design and algorithm works randomly. We run this algorithm randomly a few times to find best filter coefficients. Our aim is in order to find a filter whose coefficients have a low complexity. It means that filter coefficients have a few number of 1s digits in the case of binary expansion.

4.1 Binary Digit Representation

We assume that numbers $z \in R$ such that

$$z = s\sum_{i=-L}^{R} b_i.2^{-i} \tag{4.3}$$

where L and R nonnegative integers, s is the sign of z. $s \in \{-1,0,1\}$ for the binary case $b_i \in \{0,1\}$

L+1 and R are the number of the bits in the integer and fractional part of z respectively for the binary expansion. In this work, we will focus on a measure.This is the number of 1s in the binary representation of the coefficients $\phi_{bin}(z)$. We define also $\Phi_{bin}(\beta)$ as the binary complexity of the β.

$$\phi_{bin}(z) = \sum_{i=-L}^{R} b_i \tag{4.4}$$

4.2 Filter Specifications

In this thesis we try to design finite impulse response filter with low complexity coefficients. While we are making this, we should describe a filter specifications. For our algorithm, we should check filter coefficient vector β is acceptable. To do this, we must control the H_β is the stable or not. We describe a filter specifications according to upper and lower bounds of magnitude, phase and group delay values. Our filter specifications can be written such that;

$$L_{mag}(w) \leq \left|H_\beta(e^{-jw})\right| \leq U_{mag}(w) \tag{4.5}$$

$$L_{ph}(w) \leq \angle H_\theta(e^{-jw}) \leq U_{ph}(w) \tag{4.6}$$

$$L_{gd}(w) \leq G_\beta(w) \leq U_{gd}(w) \tag{4.7}$$

where

G_β is the group delay of the filter.

$$G(w) = -\frac{d}{dw}\angle H_\beta(e^{-iw}) \tag{4.8}$$

$$= \Re\left\{e^{-iw}\frac{H'(e^{-iw})}{H(e^{-iw})}\right\}$$

where

$$z = e^{-iw} \text{ with } w \in [0,2\pi] \tag{4.9}$$

We describe our filter specification according to patients' hearing loss range. We will take frequency range of filter between 125 Hz and 1000 Hz. It is a hearing loss range which person has a meniere hearing loss type.

4.3 Filter Parametrization

We accept our filter transfer function is $H_\beta(z)$ and its derivative with respect to z, $H_\beta^{'}(z)$.

Different filter coefficients give us different transfer functions. Transfer function of FIR filter is such as

$$H_\beta(z) = \sum_{n=0}^{N} b_n.z^{-n} \tag{4.10}$$

Filter coefficient vector $\beta = (b_0, b_1, ... b_N) \in R^p$ with p= N+1.

$$\begin{aligned} H_\beta(z) &= b_0.z^0 + b_1.z^{-1} + b_2.z^{-2} + ... b_N.z^{-N} \\ &= b_0 + b_1.z^{-1} + b_2.z^{-2} + ... + b_N.z^{-N} \end{aligned} \tag{4.11}$$

4.4. Algorithm

Firstly we design nominal filter design without any filter specification. We assume that filter has finite complexity and its filter order is 40 bits. Each step has an index i. $i \in \{1 .... p\}$.Index of filter is randomly chosen and for each step it is updated. All filter coefficients must be fixed except β_i. We will use adjust procedure to find β_i with lowest complexity. When β_i decreases, filter's complexity will decreases. Our algorithm runs randomly and it runs each time there will be different values for different points. Adjust procedure runs until β_i is fixed. In this algorithm, input is β ($\beta \in C$) and index is i ($i \in \{1 .... p\}$). Output is $\hat{\beta}_i$ and its index i is

$$\hat{\beta}_i = (\beta_1, ..., \beta_{i-1}, \hat{\beta}_i, \beta_{i+1}, ... \beta_p) \tag{4.12}$$

4.4.1 Main algorithm

Our main algorithm has following form [21]:

$\beta = \beta_{nom}$

repeat

$\beta^{prev} = \beta$

We can choose permutation of π from 1 to p randomly such as

$\pi = (2,4,1,5,6,8,7,3.....39,40)$ where p=15

for i= 1: p

j=π(i)

$\beta_j = adjust(\beta, j)$

until

$\Phi(\beta) = \Phi(\beta^{prev})$

Main algorithm may be executed at most $\Phi(\beta^{nom})$ times. Each step the main loop calls adjust procedure p times.

4.4.2 Adjust procedure

In this procedure, our aim is in order to find β_i value with lowest complexity. Procedure runs randomly p times and main algorithm calls adjust procedure at most p times. Procedure is following form:

For example, we take β_i value from main algorithm randomly and then we look β_i value 0 or not. If β_i value is 0, $\hat{\beta}_i$ will be equal to β_i, it means that filter has lowest complexity.If β_i value is different than 0, trunc procedure will be called automatically. Trunc procedure is run for each coefficient to find complexity of coefficients. Adjust procedure show that complexity of $\hat{\beta}_i$ is lower than complexity of β_i. We try to find lowest complexity of β_i.

4.4.3 Trunc procedure

In this procedure, we aim to find complexity of each of coefficient, respectively. Firstly, coefficients should have converted from decimal to binary representation and then we should define upper and lower values for all coefficient.Trunc calls adjust procedure for each coefficient.It runs each time there will be different values for different points. Also, upper and lower values should have converted from decimal values to binary value represantation. This process must have applied binary and csd values of filter coefficients.Trunc procedure start with comparing of sign of upper and lower values of coefficient. If sign of these values are opposite, it shows lowest complexity. If sign of these values are same, trunc procedure start working and it calculates lower and upper values of coefficient and it averages these values and then we find $\hat{\beta}_i$ value whose complexity is lower than β_i.We design a new filter with $\hat{\beta}_i$ values for each coefficient and then we will compare complexity and transfer functions of these filters. Complexity of filter should be lower than complexity of nominal filter.

CHAPTER 5

SIMULATIONS

In this section, we will compare simulation results of current based techniques using in digital filters and low complexity method in MATLAB. Then, we compare simulation results according to our audiograms.

5.1 Case I. Hamming Window Function Method

If we apply windowing method with hamming function for fir filter design;

Filter order , N =10;
Sampling frequency – fs=22050Hz;
Passband cut-off frequency – fc1=3KHz;
Stopband cut-off frequency – fc2=6KHz; and
Minimum stopband attenuation – 40dB.

A total number of filter coefficients is larger by one, i.e. N+1=11; and coefficients have indices between 0 and 10.

$$w_{c1} = \frac{2\pi f_{c1}}{fs} = \frac{2.\pi.3000}{22050} = 0.2721\,\pi \tag{5.1}$$

$$w_{c2} = \frac{2\pi f_{c2}}{fs} = \frac{2.\pi.6000}{22050} = 0.5442\,\pi \tag{5.2}$$

The required transition region of the filter is:

$$\Delta w = w_{c2-} - w_{c1} = 0.2721\pi \tag{5.3}$$

The Hamming window function coefficients are found via expression:

$$w[n] = 0.54 - 0.46(1 - \cos(\frac{2\pi n}{N-1}));0 \le n \le N-1 \tag{5.4}$$

$$w[n] = \begin{Bmatrix} 0.08,0.167852,0.397852,0.682148,0.912148, \\ 1,0.912148,0.682148,0.397852,0.167852,0.08 \end{Bmatrix}$$

The ideal low-pass filter coefficients are expressed as;

$$h_d[n] = \begin{Bmatrix} \dfrac{\sin[w_c(n-M)]}{\pi[n-M]};n \ne M \\ \dfrac{w_c}{\pi};n = M \end{Bmatrix} \tag{5.5}$$

where M is the index of middle coefficient.

$$M = \frac{N}{2} = 6,\ w = w_{c1} = 0.2721\pi$$

$$h_d[n] = \begin{Bmatrix} -0.0048511,-0.057649,-0.021826,0.057883,0.0157622, \\ 0.240157,0.272109,0.240157,0.157622,0.57883,0.-0.021826, \\ -0.057649,-0.048511\ \} \end{Bmatrix} \tag{5.6}$$

The designed FIR filter coefficients are found via expression:

$$h[n] = w[n] * h_d[n]\ 0 \le n \le 12 \tag{5.7}$$

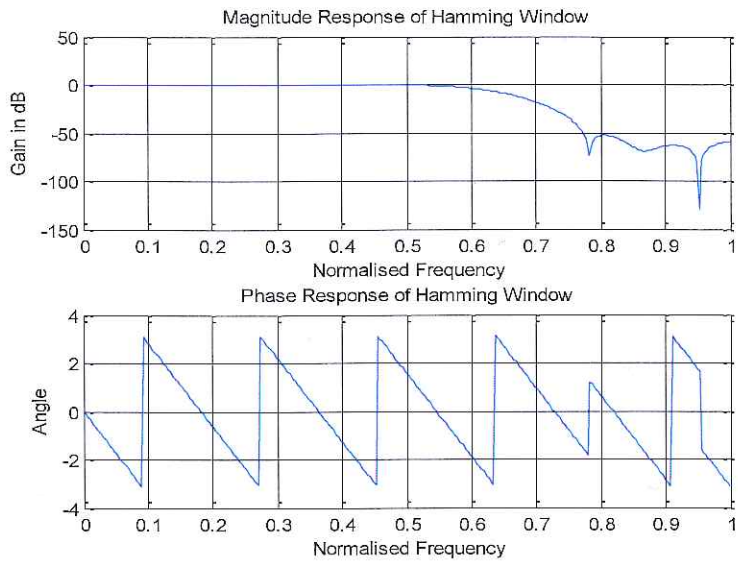

Figure 12 Magnitude response of hamming window function

5.2 Case II. Hanning Window Function Method

Filter order, N =10;

Sampling frequency – fs=8000Hz;

Passband cut-off frequency – fc1=2KHz;

Stopband cut-off frequency – fc2=3KHz;

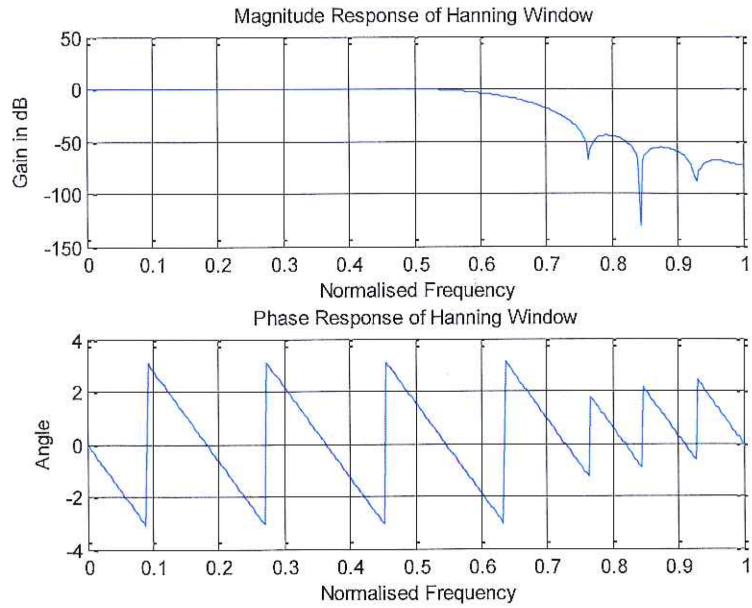

Figure 13 Magnitude and phase response of Hanning Window

5.3. Case III. Kaiser Window Function

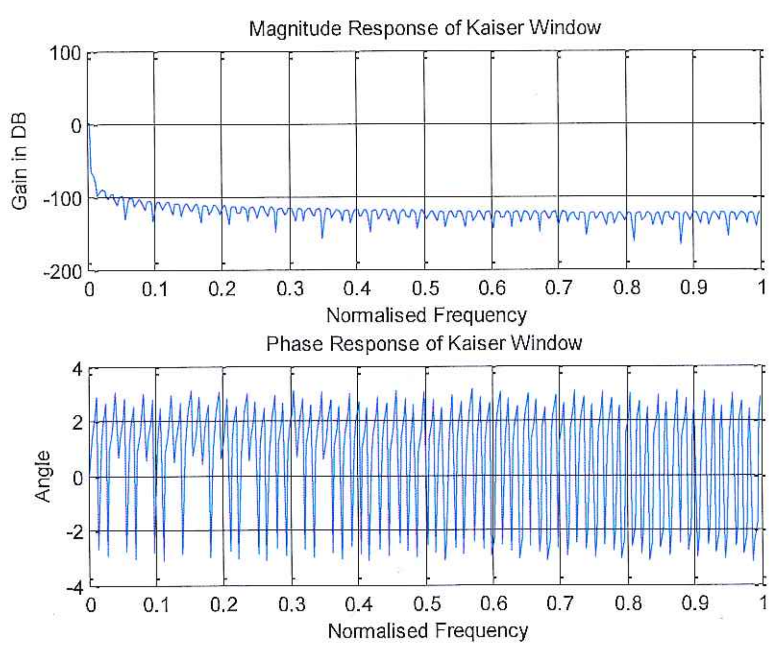

Figure 14 Magnitude and phase response of kaiser window function

5.4 Case IV. Frequency Sampling Method

If we use frequency sampling method for FIR filter design with frequency range from 3kHz to 6 kHz. Generally, hearing loss is around 4 kHz. It shows noise induced hearing loss example. Our filter specification should be adjusted according to this frequency range.

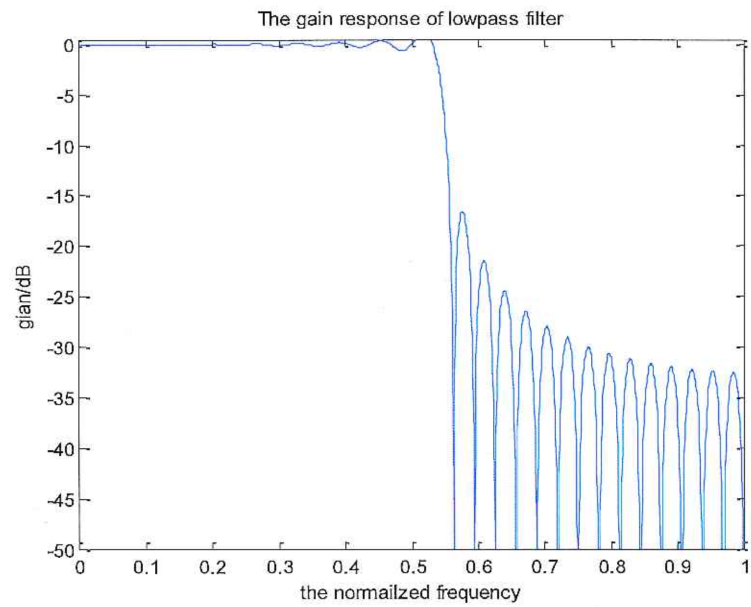

Figure 15 Magnitude response of Frequency Sampling method

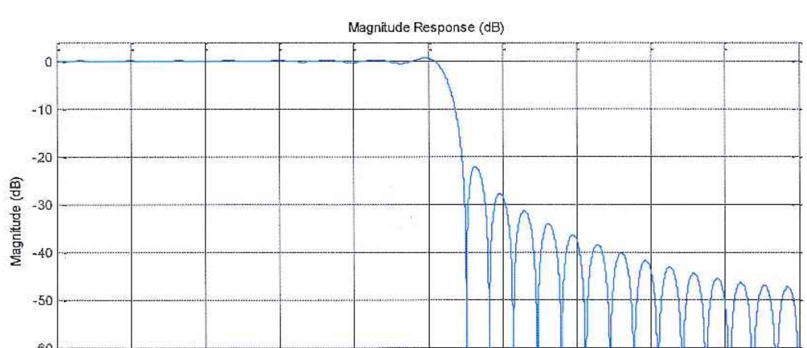

Figure 16 Kaiser Window for low pass filter

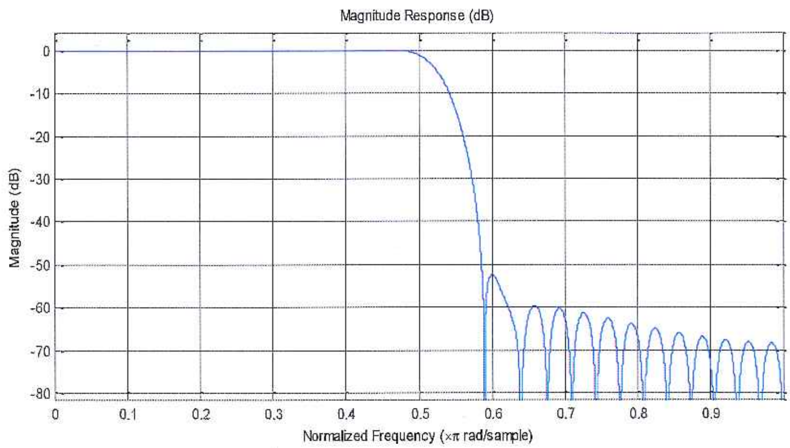

Figure 17 Hamming Window for low pass filter

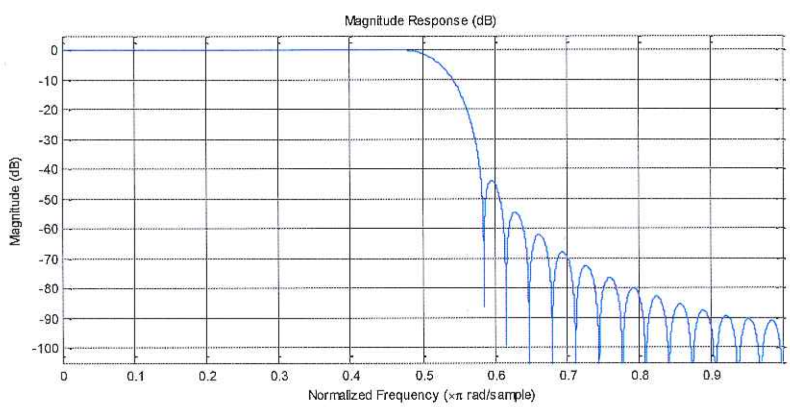

Figure 18 Hanning Window for low pass filter

Table 2 Filter Specification

Parameter	Values
Sampling Frequency	1000 Hz
Cut off Frequency	[0.125 .275]
Order	73

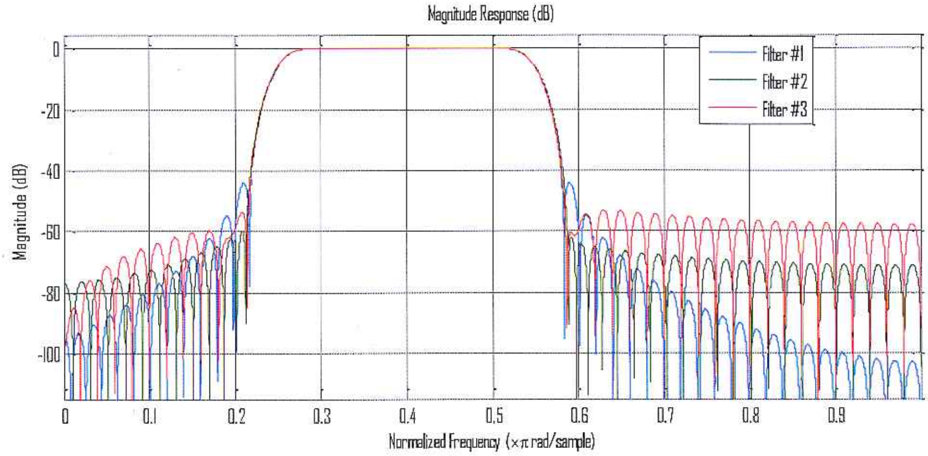

Figure 19 Bandpass FIR Filter comparison of matching result for Hamming(Filter1), Hanning (Filter2) and Kaiser Window(Filter3)

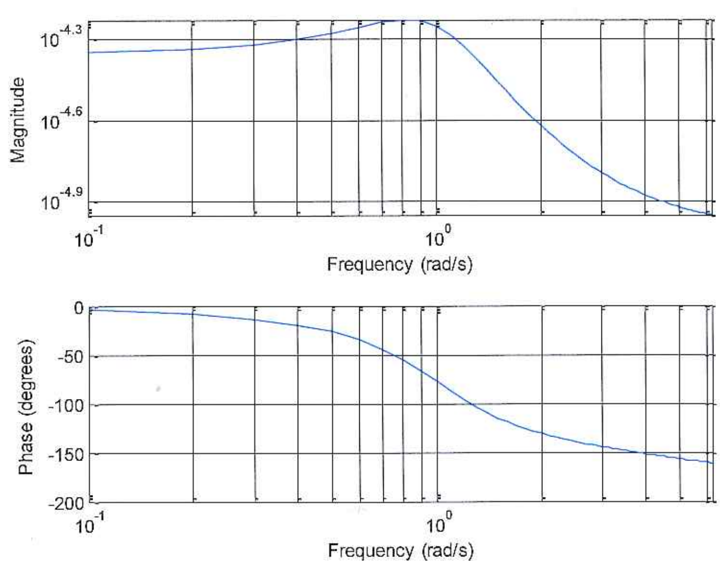

Figure 20 Frequency and Phase Response of Kaiser Window for low pass filter

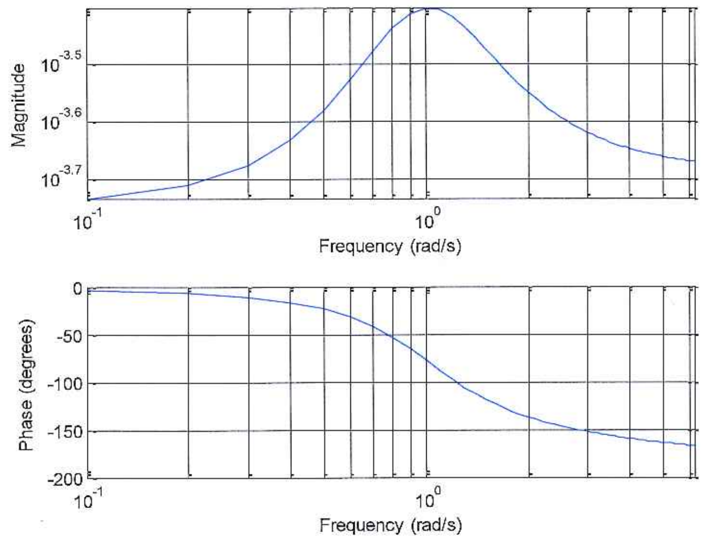

Figure 21 Frequency and Phase Response of Hanning Window for low pass filter

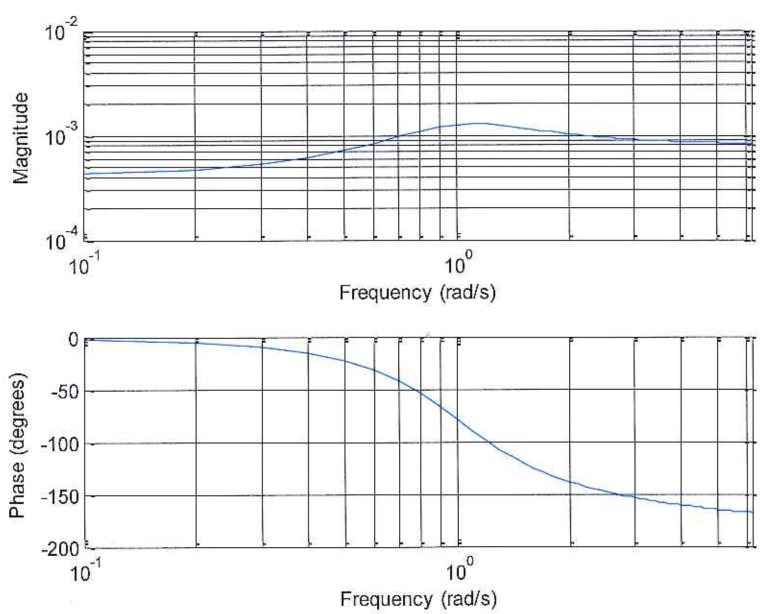

Figure 22 Frequency and Phase Response of Hamming Window for low pass filter

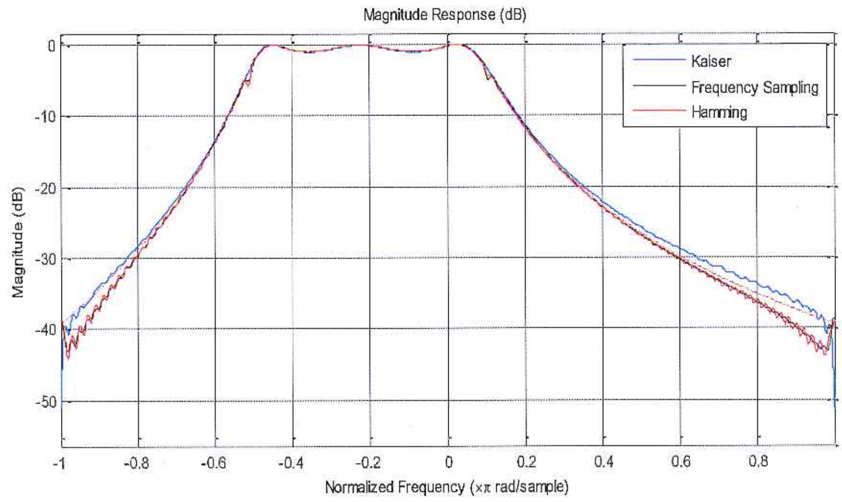

Figure 23 Matching Result for Kaiser, Frequency Sampling and Hamming Functions

Table.3 Values of bandwidth and side lobe for different functions

Function	Side Lobe Level	3 dB BW	6 dB BW	Noise BW
Hannning	-32,17	1,472	2,048	1,52
Hamming	-43,5	1,344	1,856	1,37
Kaiser	-69,5	1,728	2,432	1,82
Frequency Sampling	-19,2	1,253	2,045	1,62

Ideal filter for hearing aid must have small main lobe width, high side lobe (for good noise suppression) and side lobes should fall off rapidly. Figure 19 shows that magnitude response of windowing functions. Kaiser window function has higher side lobe than other window functions. Therefore, Kaiser window function can be good selection to match the audiogram for hearing aid.

5.5 Low Complexity Method

If we apply low complexity algorithm our filter, magnitude response is,

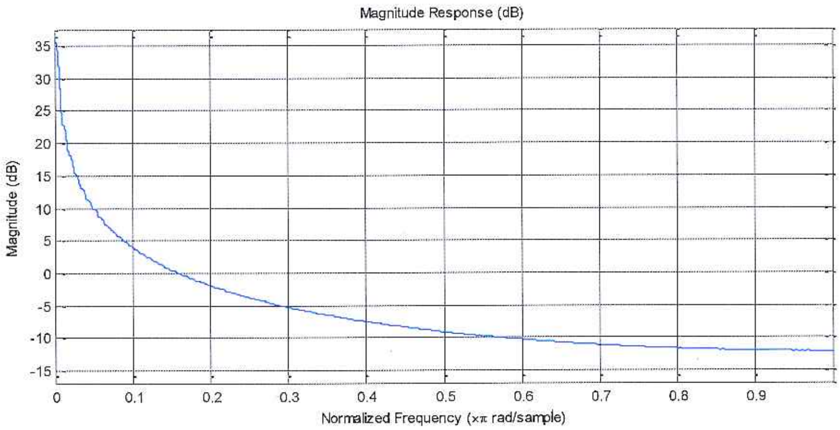

Figure 24 Frequency response magnitude of the nominal filter for Menire's hearing loss

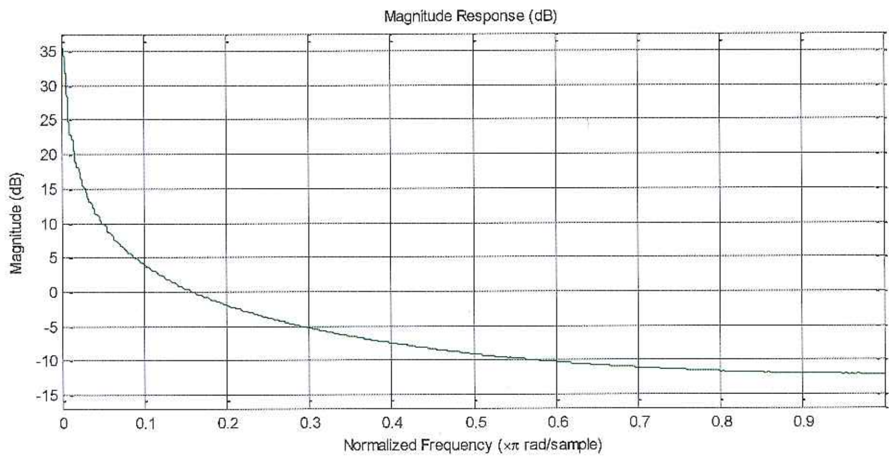

Figure 25 Frequency response magnitude of low complexity filter

If we compare magnitude response of nominal filter and low complexity filter, frequency responses are approximately same. Complexity of nominal filter is 476, after we run algorithm ,complexity of filter will become 421.We run algorithm again, complexity becomes 300, at last it will be 185. It means that, $\Phi(\beta_{nom}) = 476$ at first time running algorithm, $\Phi(\beta) = 421$ and if we run a few times algorithm, binary complexity of filter will become $\Phi(\beta) = 185$around 3 1 s per coefficient.

If we apply low complexity algorithm a filter whose frequency range is between 4 kHz and 8 Khz. This frequency range shows that hearing loss range of presbycusis.

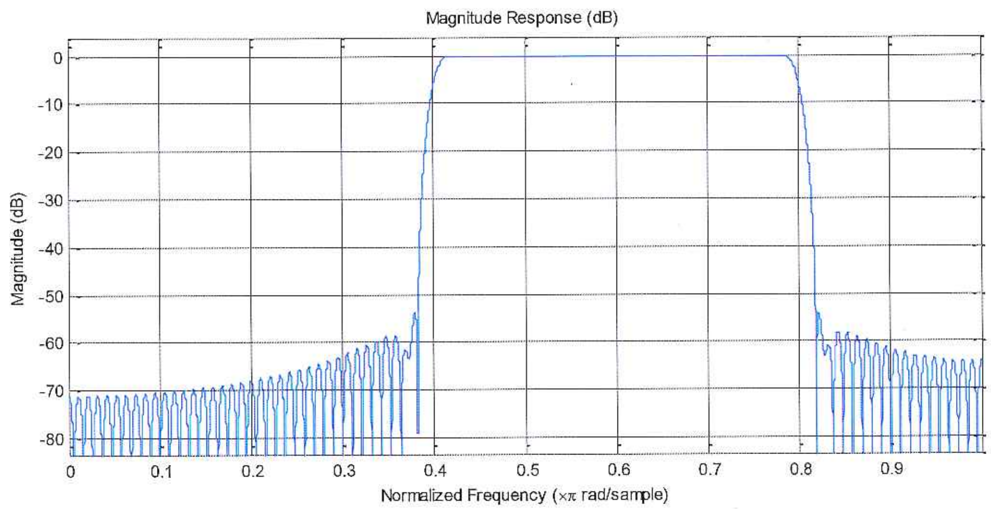

Figure 26 Frequency magnitude response of the nominal filter for Presbycusis hearing loss

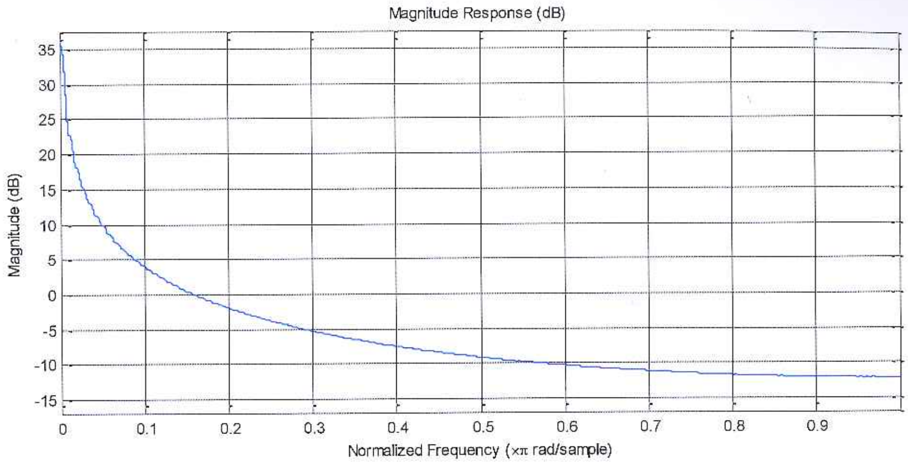

Figure 27 Magnitude response of the nominal filter for Presbycusis hearing loss

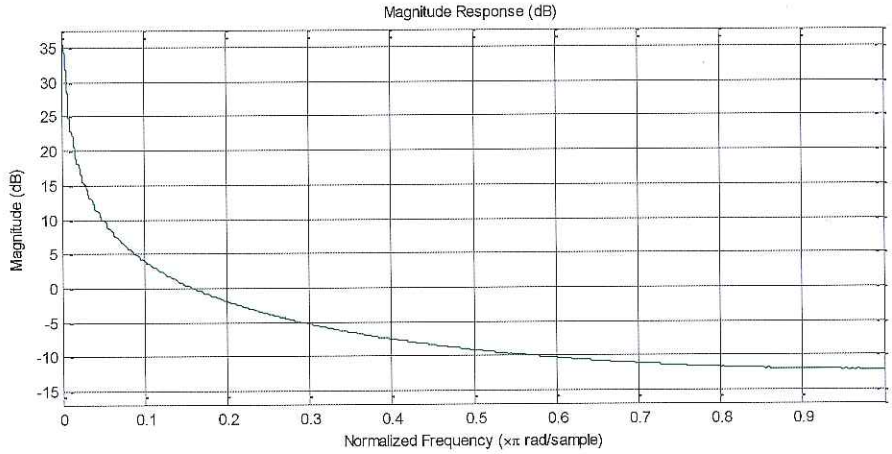

Figure 28 Magnitude response of the low complexity filter for Presbycusis hearing loss

Figure 26 shows magnitude response of FIR filter for the Presbycusis hearing loss. If we apply low complexity algorithm once, magnitude response of filter will be approximately same. If we apply algorithm to filter several times, magnitude response of filter will be same. Figure 30 shows this situation.

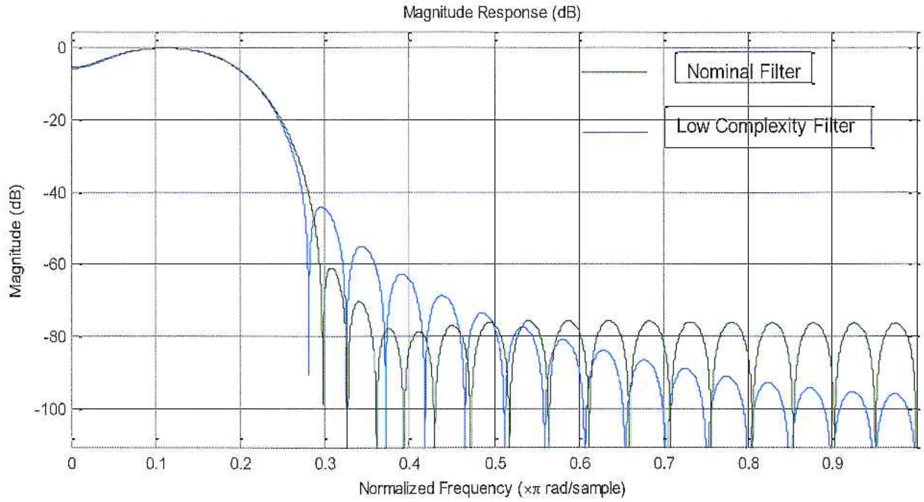

Figure 29 Comparison of magnitude responses of nominal and low complexity filter Meniere hearing loss

In Figure 29, we compare magnitude response of nominal and low complexity filter for the Meniere haering loss.

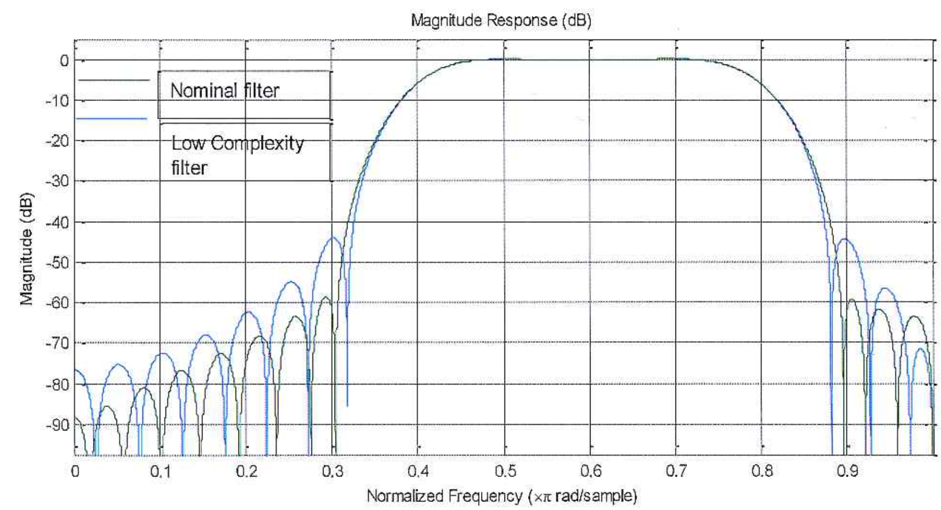

Figure 30 Comparison of magnitude responses of nominal and low complexity filter Presbycusis hearing loss

If we want to calculate error rate of magnitude response of standart filter and low complexity filter, we can define as following;

MR_SF= Magnitude response of standart filter

MR_LF= Magnitude response of low complexity filter

$$\text{Error}=|MR\_SF-MR\_LF|/\ |MR\_SF|*100 \tag{5.8}$$

If we calculate error rate of magnitude response of standart filter and low complexity filter for Meniere hearing loss, result will be following;

$$\text{Error} = |31 - 30|/|31|*100$$
$$= 3{,}22\ \%$$

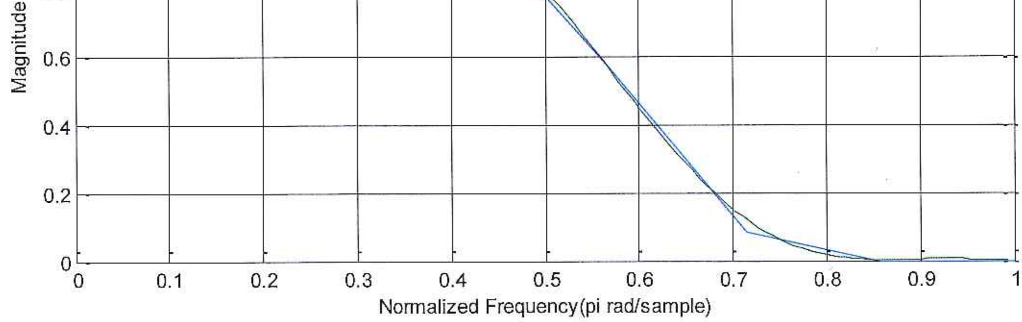

Figure 31 Comparison of error rate magnitude responses of standart and low complexity filters

If we calculate error rate of magnitude response of standart filter and low complexity filter for Presbycusis hearing loss, result will be following;

$$\begin{aligned} \text{Error} &= |MR\_SF - MR\_LF| / |MR\_SF| * 100 \\ &= |20 - 19| / |20| * 100 \\ &= 5\ \% \end{aligned}$$

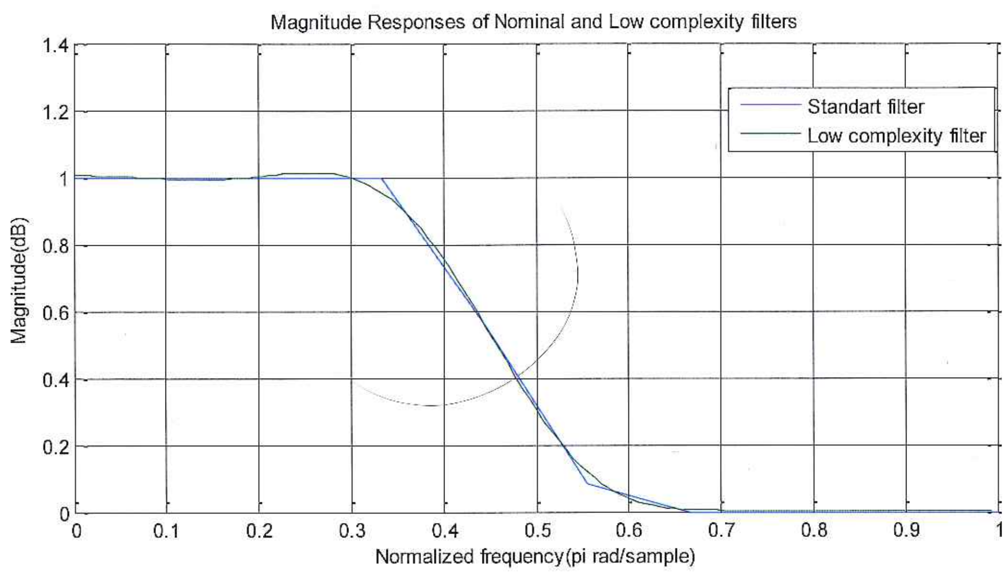

Figure 32 Comparison of error rate of magnitude responses of standart and low complexity filters

Figure 31 shows error rate of magnitude responses of standart and low complexity filters for Meniere hearing loss. When we calculate error rate of magnitude response, result will be 3,22 %. It means that magnitude response of audiogram and low complexity filter match. There is only an error rate of 3,22 percent difference. Nonetheless, Figure 32 shows error rate of magnitude responses of standart and low complexity filters for Presbycusis hearing loss. When we calculate error rate of magnitude response, result will be 5 %. It means that magnitude response of audiogram and low complexity filter match. There is only an error rate of 5 percent difference.

Table 3 Nominal and truncated coefficient value for Meniere's

Nominal value	truncated value
0.0019	2^(-10)+2^(-11)+2^(-12)+2^(-13)
0.0048	2^(-8)+2^(-10)
0.0094	2^(-7)+2^(-10)+2^(-11)
0.0152	2^(-6)
0.0206	2^(-6)+2^(-7)+2^(-8)
0.0238	2^(-6)+2^(-7)
0.0224	2^(-3)+2^(-5)+2^(-6)
0.0143	2^(-6)+2^(-8)+2^(-9)
-0.0012	-(2^(-7)+2^(-9)+2^(-10))
-0.0230	-(2^(-7)+2^(-8))
-0.0479	-(2^(-5))
-0.0708	-(2^(-4))
-0.0857	-(2^(-4)+2^(-5))
-0.0875	-(2^(-5)+2^(-6))
-0.0731	-(2^(-4)+2^(-5))
-0.0427	-(2^(-5)+2^(-8))
-0.0042	-(2^(-1)+2^(-4)+2^(-5)+2^(-7)+2^(-9))
0.0466	2^(-4)+2^(-6)
0.0893	2^(-4)+2^(-5)+2^(-6)
0.1192	2^(-4)+2^(-5)+2^(-7)+2^(-8)
0.1298	2^(-3)
0.1192	2^(-4)+2^(-5)+2^(-7)+2^(-8)
0.0893	2^(-4)+2^(-5)+2^(-6)
0.0466	2^(-4)+2^(-6)
-0.0042	-(2^(-1)+2^(-4)+2^(-5)+2^(-7)+2^(-9))
-0.0427	-(2^(-5)+2^(-8))
-0.0731	-(2^(-4)+2^(-5))
-0.0875	-(2^(-5)+2^(-6))
-0.0857	-(2^(-4)+2^(-5))

Table 4 Nominal and truncated coefficient value for Presbycusis

Nominal value	truncated value
-0.00314	-(2^(-2)+2^(-3)+2^(-8))
0.0098	2^(-1)+2^(-4)+2^(-7)+2^(-8)
0.0042	2^(-1)+2^(-3)+2^(-4)+2^(-7)
-0.0043	-(2^(-1)+2^(-2)+2^(-3)+2^(-7))
-0.0010	-(2^(-1)+2^(-2)+2^(-8))
-0.00013	-(2^(-1)+2^(-3)+2^(-4)
0.00115	2^(-2)+2^(-8)+2^(-10)
0.0051	2^(-1)+2^(-2)+2^(-6)+2^(-7)+2^(-8)
-0.0054	-(2^(-1)+2^(-3)+2^(-6)+2^(-7))
-0.0013	-(2^(-1)+2^(-2)+2^(-8))
0.00115	2^(-1)+2^(-3)+2^(-8)
0.00151	2^(-1)+2^(-6)+2^(-8)
0.00682	2^(-1)+2^(-2)+2^(-4)+2^(-8)
-0.00727	-(2^(-1)+2^(-2)+2^(-4)+2^(-8))
-0.00183	-(2^(-1)+2^(-3)+2^(-4)+2^(-5)+2^(-7))
0.000211	2^(-1)+2^(-2)+2^(-3)+2^(-5)
0.000209	2^(-1)+2^(-4)+2^(-5)
0.000946	2^(-2)+2^(-4)+2^(-5)+2^(-6)
-0.00101	-(2^(-2)+2^(-8)+2^(-10))
-0.000255	-(2^(-1)+2^(-3)+2^(-4))
-0.000809	-(2^(-1)+2^(-5)+2^(-6))
0.00291	2^(-2)+2^(-5)^+2^(-8)
0.00131	2^(-1)+2^(-3)+2^(-8)
-0.00140	-(2^(-1)+2^(-6)+2^(-8))
-0.0035	-(2^(-2)+2^(-5))
-0.00355	-(2^(-1)+2^(-2)+2^(-4)+2^(-6))
0.00401	2^(-1)+2^(-2)+2^(-4)+2^(-6)
0.0018	2^(-1)+2^(-3)+2^(-4)+2^(-5)+2^(-8)
-0.0019	-(2^(-1)+2^(-5)+2^(-6))
-0.0004	-(2^(-1)+2^(-4)+2^(-5))
0.00553	2^(-2)+2^(-7)+2^(-9)
0.00543	2^(-2)+2^(-7)+2^(-8)
0.00244	2^(-2)+2^(-5)+2^(-8)
-0.00258	-(2^(-1)+2^(-3)+2^(-4)+2^(-6)+2^(-7))

CHAPTER 6

CONCLUSIONS

In this thesis, we compared performance analysis of FIR filter in different dsp algorithms for digital hearing aids. Firstly, we analyzed windowing method types and frequency sampling method for fir filters. We try to match frequency response of these methods with auidogram.

Moreover, a defined coefficient groups should be used in hearing aid processing. We designed a filter without any specifications. With low complexity algorithm we aim to provide a low-computational complexity filters and we try to this algorithm different hearing loss types such as Presbycusis and Menire's hearing loss. Low complexity means that number of 1 s in each coefficient is minimum. For digital hearing aid design, the magnitude response of the filter should match the audiogram.

In this thesis, we deal with comparing of results of these methods for matching the audiogram. In the filter design, one of the important thing is calculation of filter coefficients. If the values of the filter coefficient are complex numbers, filter structures would be computationally demanding. These results are unwanted things in the filter design. Because of these reasons, with low complexity algorithm, we provide to implement low complexity filter design which has lowest complexity coefficient.

If we compare magnitude response of nominal filter and low complexity filter, magnitude responses are approximately same. As well as that, complexity of nominal filter is 476, after we run algorithm, complexity of filter will become 421 and then we run algorithm again, complexity becomes 300, at last it will be 185. It means that, $\Phi(\beta_{nom}) = 476$ at first time running algorithm, $\Phi(\beta) = 421$ and if we run a few times algorithm, binary complexity of filter will become $\Phi(\beta) = 185$ around 3 1 s per coefficient.Figure 26 shows magnitude response of FIR filter for the Presbycusis.

If we apply low complexity algorithm once, magnitude response of filter will be approximately same. In Figure 31 shows error rate of magnitude responses of standart and low complexity filters for Meniere hearing loss. When we calculate error rate of magnitude response, result will be 3,22 %. It means that magnitude response of audiogram and low complexity filter match. There is only an error rate of 3,22 percent difference. Nonetheless, Figure 32 shows error rate of magnitude responses of standart and low complexity filters for Presbycusis hearing loss. When we calculate error rate of magnitude response, result will be 5 %. It means that magnitude response of audiogram and low complexity filter match. There is only an error rate of 5 percent difference.

In digital hearing aid design, an important thing is to match to the magnitude response of filter and audiogram for the different hearing loss types. With low complexity algorithm, we try to match our filter magnitude response to audiogram with low comlexity coefficients. Our results show that magnitude response of filters approximately same.

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APPENDICES

Appendix A

Main Algorithm

```
function [tetai]=nomfilter(tetanom,teta)
%format long
N=40;% FIR filter order
d = design.bandpass('N,Fst1,Fp1,Fp2,Fst2,C',N,100,125,1e3,1.2e3,1e4);
d = fdesign.bandpass('N,Fst1,Fp1,Fp2,Fst2,C',N,100,4e3,7e3,7.2e3,1e4);
d.Stopband1Constrained = true; d.Astop1 = 60;
d.Stopband2Constrained = true; d.Astop2 = 60;
Hd = design(d,'equiripple');
fvtool(Hd);
measure(Hd)
tetanom=[0.001950 0.0048383 0.009473 0.015204 0.020696 0.023891 0.022422
0.014309 -0.001223 -0.023065 -0.047955 -0.070825 -0.085792 -0.087599 -
0.073166 -0.042761 -0.00424 0.046604 0.089358 0.119197 0.129898 0.119197
0.089358 0.046604 -0.0042 -0.042761 -0.073166 -0.087599 -0.085792 -0.070825 -
0.047955 -0.023065 -0.001223 0.014309 0.022422 0.023891 0.020696 0.015204
0.009473 0.004838 0.001950];
 teta=tetanom;
 tetaprev=teta;
while tetaprev==teta
 p=15;
 for i=1:p
   j=tetanom(i);
   nomfilter=@adjust
   teta(i)=tetanom(randi(numel(tetanom)));
   tetaj=adjust(teta(i),j);
   phiteta=sum(tetaj,'double');
  phitetaprev=sum(tetanom,'double');
      if phiteta==phitetaprev;
  break
       end
      disp(tetaj)
  phiteta=phitetaprev;
  break
end
end
```

Appendix B

Adjust procedure

```
function [tetahdi] = adjust(li,ui)

tetanom=[0.001950 0.0048383  0.009473  0.015204   0.020696  0.023891  0.022422
0.014309 -0.001223 -0.023065 -0.047955  -0.070825 -0.085792  -0.087599 -
0.073166  -0.042761 -0.000424  0.046604  0.089358  0.119197  0.129898  0.119197
0.089358  0.046604 -0.000424 -0.042761 -0.073166 -0.087599 -0.085792 -0.070825
-0.047955 -0.023065 -0.001223  0.014309  0.022422  0.023891   0.020696  0.015204
0.009473  0.004838  0.001950];
tetai =tetanom(randi(numel(tetanom)));
%disp(prm);
% tetai=prm;
disp(tetai);
 tetahdi=@trunc;
tetahdi=(li+ui)/2;

-2*tetai<tetai ,tetai<2*tetai;
for k=0:15
   li=sum(2.^(-k));
   ui=sum(2.^(-k));
   end
 l=-2*abs(tetai);
u=2*abs(tetai);
tetahdi=(li+ui)/2;
N=40;
Fst1=100;
Fp1=125;
Fp2=1000;
Fst2=1200;
C=10000;
if tetai==0
   tetahdi=tetai;
end
   l=-2*abs(tetai)
   u=2*abs(tetai)
   return
   if N==40
      Fst1=100;
      Fp1=125;
      Fp2=1000;
      Fst2=1200;
      C=10000;
```

```
    tetahdi=trunc(tetai,li,ui);
    tetahdi=(li+ui)/2;
    disp(tetahdi);
    else
    if tetahdi>tetai
       u=tetahdi;
    else
       l=tetahdi;
    end
    end
    trunc
end
```

Appendix C

Trunc procedure

```
function [tetahdi]= trunc(li,ui)

su=[-1,0,1];
sl=[-1,0,1];
i=1:15;
if su.*sl==-1
    tetahd(i)=0;
else

tetai=0.0018;
l(i)=tetai-0.09*tetai
u(i)=tetai+0.09*tetai
tetaii=Fr_dec2bin(tetai)
li=Fr_dec2bin(l(i))
ui=Fr_dec2bin(u(i))

  for i=1:15;

   if i==1;
     l(1)= input('enter the l(1) value')
     u(1)= input('enter the u(1) value')
    teta1=(l(1)+u(1))/2
   end
   if i==2;
     l(2)= input('enter the l(2) value')
     u(2)= input('enter the u(2) value')
    teta2=(l(2)+u(2))/2
   end
   if i==3;
     l(3)= input('enter the l(3) value')
     u(3)= input('enter the u(3) value')
    teta3=(l(3)+u(3))/2
   end
   if i==4;
     l(4)= input('enter the l(4) value')
     u(4)= input('enter the u(4) value')
    teta4=(l(4)+u(4))/2
   end
   if i==5;
     l(5)= input('enter the l(5) value')
     u(5)= input('enter the u(5) value')
```

```
 teta5=(l(5)+u(5))/2
end
if i==6;
  l(6)= input('enter the l(6) value')
  u(6)= input('enter the u(6) value')
 teta6=(l(6)+u(6))/2
end
if i==7;
  l(7)= input('enter the l(7) value')
  u(7)= input('enter the u(7) value')
 teta7=(l(7)+u(7))/2
end
if i==8;
  l(8)= input('enter the l(8) value')
  u(8)= input('enter the u(8) value')
 teta8=(l(8)+u(8))/2
end
if i==9;
  l(9)= input('enter the l(9) value')
  u(9)= input('enter the u(9) value')
 teta9=(l(9)+u(9))/2
end
if i==10;
  l(10)= input('enter the l(10) value')
  u(10)= input('enter the u(10) value')
 teta10=(l(10)+u(10))/2
end
if i==11;
  l(11)= input('enter the l(11) value')
  u(11)= input('enter the u(11) value')
 teta11=(l(11)+u(11))/2
end
if i==12;
  l(12)= input('enter the l(12) value')
  u(12)= input('enter the u(12) value')
 teta12=(l(12)+u(12))/2
end
if i==13;
  l(13)= input('enter the l(13) value')
  u(13)= input('enter the u(13) value')
 teta13=(l(13)+u(13))/2
end
if i==14;
  l(14)= input('enter the l(14) value')
  u(14)= input('enter the u(14) value')
 teta14=(l(14)+u(14))/2
end
if i==15;
```

```
    l(15)= input('enter the l(15) value')
    u(15)= input('enter the u(15) value')
   teta15=(l(15)+u(15))/2
  end
  if i>15
     break
  end
end


        disp(tetahd(i))
       end
        end


if ((l<tetahdi)&&(tetahdi<u)&&(phitetahdi<=phitetai))
  return
end
```

Appendix D

Comparison of frequency responses of filters

```
b= [ 0.0019 0.0095 0.0152 0.0206 0.238 0.0143 -0.00122 -0.0230 -0.0482 -0.0708 -
0.0857 - 0.0875 -0.0731 0.0466 0.0893 0.1192 0.1299 0.1192 0.0893 0.04666];
a=1;
[Hf,f]=freqz(b,a,256,1);
HfA=abs(Hf);
Hfphi=angle(Hf);
plot(f,HfA)
%fvtool(f,1)
d=1;
c=[ 0.0018 0.085 0.0151 0.0195 0.0195 0.0127 -0.0093 -0.0176 -0.0313 -0.0625 -
0.0781 -0.0781 -0.0703 0.0391 0.0742 0.1055 0.1250 0.1050 0.0742 0.0399]

[Hx,x]=freqz(d,c,256,1);
HfX=abs(Hx);
plot(x,HfX)
%fvtool(x,1)
fvtool(f,1,x,1)
```

Appendix E

Windowing Method

```
clear all;
close all;
clc ;
% Kaiser Window
 %pass band ripple = 0.087dB
%stop band attenuation = 60dB
%pass band edge frequency = 0.4? rad/sec
%stop band edge frequency = 0.6? rad/sec
%sampling frequency = 100 rad/sec
fs = 100;              %rad/sec, sampling frequency
pf = 0.4*pi;           %rad/sec, pass band frequency
sf = 0.6*pi;           %rad/sec, stop band frequency
fsamp = fs/(2*pi);
 % fs into hertz
pf1 = pf/(2*pi);
% pf into hetz
sf1 = sf/(2*pi);
% sf into hertz
d1 = 10^(-0.05*60);    %stop band ripple
d2 = (10^(0.05*0.087)-1)/(10^(0.05*0.087)+1);  %pass band ripple
fcuts = [pf1 sf1];     %pass and stop band cut off frequency
mags = [1 0];          %magnitude assigning
%d3 = min(d1,d2);
devs = [d2 d1];        %setting deviation
w = 0:0.01:pi;
[n,Wn,beta,ftype] = kaiserord(fcuts,mags,devs,fs);
%filter design using windows
 h = fir1(n,Wn,ftype,kaiser(n+1,beta),'noscale');
[H,f] = freqz(h,1,w);  % frequency response and frequency vector
gain = 20*log10(abs(H));
a = angle(H);
%PLOTTING THE GRAPHS
subplot(2,1,1);
plot(f/pi, gain); grid on;
title(' Magnitude Response of Kaiser Window');
xlabel(' Normalised Frequency');
ylabel (' Gain in DB');
subplot(2,1,2);
plot(f/pi, a); grid on;
title('Phase Response of Kaiser Window');
xlabel(' Normalised Frequency');
ylabel(' Angle');
```

```
fsamp = 8000;
%sampling frequency
fcuts = [2000 3000];
%pass band and stopband edge frequency
mags = [0 1];
%magnitude
d1 = 10^(-0.05*44);
%stop band ripple
d2 = (10^(0.05*0.1)-1)/(10^(0.05*0.1)+1);
%pass band ripple
devs = [d2 d1];    %setting deviation
W = 0:0.01:pi;
[n,Wn,beta,ftype] = kaiserord(fcuts,mags,devs,fsamp);  %to get the order
% hanning window
 h = hann(n+1);
b = fir1(n,Wn,h);
w = 0:0.01:pi;
[H,f] = freqz(b,1,w);
gain = 20*log10(abs(H));
an = angle(H);
%plotting magnitude and phase response of hanning window
figure;
subplot(2,1,1);
plot(f/pi, gain); grid on;
title('Magnitude Response Hanning Window');
xlabel('Normalised Frequency');
ylabel('Gain in dB');
subplot(2,1,2);
plot(f/pi, an); grid on;
title('Phase Response of Hanning Window');
xlabel(' Normalised Frequency');
ylabel(' Angle');
% HAMMING WINDOW
h = hamming(n+1);
b = fir1(n,Wn,h);
w = 0:0.01:pi;
[H,f] = freqz(b,1,w);
gain = 20*log10(abs(H));
an = angle(H);
%plotting magnitude and phase response of hanning window
figure;
subplot(2,1,1);
plot(f/pi, gain); grid on;
title (' Magnitude Response of Hamming Window ');
xlabel(' Normalised Frequency');
ylabel (' Gain in dB');
subplot(2,1,2);
plot(f/pi, an); grid on;
```

```
title(' Phase Response of Hamming Window ');
xlabel (' Normalised Frequency');
ylabel(' Angle');

%RECTANGULAR WINDOW
h = rectwin(n+1);
b = fir1(n,Wn,h);
w = 0:0.01:pi;
[H,f] = freqz(b,1,w);
gain = 20*log10(abs(H));
an = angle(H);
%plotting magnitude and phase response of hanning window
figure;
subplot(2,1,1);
plot(f/pi,gain); grid on;
title (' Magnitude Response of Rectangular Window ');
xlabel (' Normalised Frequency');
ylabel (' Gain in dB');
subplot(2,1,2);
plot (f/pi, an); grid on;
title ('Phase Response of Rectangular Window');
xlabel(' Normalised Frequency');
ylabel(' Angle');
```

Appendix F

Frequency Sampling Method

```
sf1 = 0.1;
pf1 = 0.35;
pf2 = 0.8;
sf2 = 0.9;
pb = linspace(pf1,pf2,1e3)*pi;
bp = fdesign.bandpass('bandpassfir', 'StopbandAttenuation1',40, 'StopbandFrequency1',sf1,...
   'PassbandFrequency1',pf1,'PassbandRipple',3,'PassbandFrequency2',pf2, ...
   'StopbandFrequency2',sf2,'StopbandAttenuation2',30);
[h,w] = freqz(bp,1024);
hpb = freqz(bp,pb);
subplot(2,1,1)
plot(w/pi,abs(h),pb/pi,abs(hpb),'.-')
axis([0 1 -1 2])
legend('Response','Passband','Location','South')
ylabel('Magnitude')
subplot(2,1,2)
plot(w/pi,db(h),pb/pi,db(hpb),'.-')
axis([0 1 -60 10])
xlabel('Normalized Frequency (\times\pi rad/sample)')
ylabel('Magnitude (dB)')
```

APPENDIX G

CIRRICULUM VITAE

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