A STUDY ON WIRELESS HEARING AIDS SYSTEM CONFIGURATION AND SIMULATION

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NATIONAL UNIVERSITY OF SINGAPORE

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SUMMARY

Conventional hearing aids have their limitations in helping the hearing impaired patients when reverberation/cross-talk is present. Although various Digital Signal Processing (DSP) algorithms have been developed for noise/reverberation cancellation, the space and power limitations imposed by single-unit hearing instruments bring design difficulties when incorporating complex DSP algorithm into a digital hearing aid.

To solve these problems, several wireless hearing aid systems have been proposed by research groups. However, the drawbacks on architectural level of these designs compromise the system performance. A single-Radio Frequency (RF) linked wireless hearing aid system based on beamforming noise cancellation technique and CMOS technology has been proposed by this work.

The cost effective implementation of wireless hearing aids requires system level simulation to ensure the functionality and evaluate the system performance. System level simulation using Advanced Design System™ (ADS) in wireless hearing aid system has never been reported before. However, the fast RF simulation feature and co-simulation ability of ADS provide capabilities for simulating electro-acoustic complex systems with DSP such as wireless hearing aids.

The whole system comprises two earpieces and a body unit. The two microphones in the body unit receives incoming sound signal. A dual-input noise cancellation DSP algorithm using two-element beamforming technique is implemented in the body unit. It attenuates reverberation and cross-talks and the processed signal is sent to the earpieces. It is further passed through several stages in the earpiece, e.g. RF receiver, demodulation, D/A conversion and output buffer and converted to sound waves out of earphone.

All block models are built in ADS 2002C environment. Behavioral modeling of electro-acoustic

transducers, i.e. microphones and earphone, is realized using pre-measured data of commercial models (BK1600 and EK3024). The dual-input noise cancellation unit is developed using functional models from ADS, as well as other function blocks. A super-heterodyne receiver structure and Quadrature Phase Shift Keying (QPSK) digital modulation scheme are realized.

The output Signal-Noise-Ratio (SNR) and input SNR relation can be obtained, and improvement of SNR across the wireless system is observed which indicates the ability of the proposed system in noise suppression. The frequency response of the whole system is seen dominated by frequency response of the electro-acoustic transducers. However, the circuit plays an important role primarily in gain enhancement, control, and SNR improvement.

A programmable non-linear compression mode is simulated. Compression knee point ranges from 50 dB to 80 dB. The output SPL is clipped at 120dB. The simulated attack time is around 9 ms and release time is 150 ms, both of which are within the normal range.

Simulations to optimize the key block parameters of the subsystem of RF transmitter and receiver are also performed on the basis of system behavioral model. The optimized system performance obtained proves that our proposed system is able to suppress background noise with less consideration on power consumption and circuit area.

NOMENCLATURES

ADC: Analog to Digital Converter

ACPR: Adjacent Channel Power Rejection

ADS: Advanced Design System™

AGC: Auto Gain Control

ANSI: American National Standard Institute

AWGN: Additive White Gaussian Noise

BER: Bit Error Rate

BiCMOS: Bipolar and CMOS technology

BTE: Behind the Ear Hearing Aid

BW: Body Worn Hearing Aid

CANS: Central Auditory Nervous System

CK: Compression Knee Point

CI: Cochlear Implants

CIC: Completely in the Canal Hearing Aid

CMOS: Complimentary Metal Oxide Semiconductor

CNS: Central Nervous System

CR: Compression Ratio

DAC: Digital-to-Analog Converter

DF: Data Flow Simulator

DSP: Digital Signal Processing

FCC: Federal Communication Commission, U. S.

FDA: The U.S. Food and Drug Administration.

FIR: Finite Impulse Response

FSK: Frequency Shift Keying

HA: Hearing Aids

IC: Integrated Circuit

IME: Institute of Microelectronics, Singapore

ISM: Industrial, Scientific and Medical Bands

ITC: In the Canal Hearing Aid

ITE: In the Ear Hearing Aid

LPRS: Low Power Radio Service

NF: Noise Factor

NIDCD: National Institute on Deafness and Other Communication Disorders, U. S.

NUS: National University of Singapore

PSK: Phase Shift Keying

QPSK: Quadrature Phase Shift Keying

RF: Radio Frequency

SNR: Signal-to-Noise-Ratio

SPL: Sound Pressure Level

UCL: Uncomfortable Loudness Level

USM: Upward Spread of Masking

VCVS: Voltage-Controlled Voltage Source

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Chapter 1. Introduction

1.1. Introduction

It is reported that 28 millions of people in United States are suffering from some kind of hearing impairment now. Between 1979 and 2002, the percentage of adults with hearing difficulties in U. K. increased from 13% to 16% according to the National Statistics of U. K.[1].

Moreover, the number of hearing impaired people is climbing because of the increasing portion of elderly people in the world. According to the survey results produced by the National Institute on Deafness and Other Communication Disorders (NIDCD) of U.S [2], hearing loss affects approximately 17 in 1,000 children under age 18. The incidence increases with age: Approximately 314 in 1,000 people over age 65 have hearing loss and 40% to 50% of people older than 75 have a hearing loss.

While hearing loss is usually caused by permanent mechanical damage to the ear, there is no effective medicine against hearing impairment, and surgery helps only in certain cases. Hearing aids are the most common form of management for hearing loss currently. Thus, electronic hearing aids or prosthetics are the best solutions for the patients so far.

Conceptually, the hearing aid is just an amplifier, picks up and amplifies sound inputs to compensate for hearing impairment. However, human hearing is too complicated and no current commercial hearing aid can perfectly compensate one's hearing loss.

The hearing aid devices have been quite useful for hearing impaired people with all types of hearing loss (conductive, sensorineural or combinational). With the evolutions in technology, the digital hearing aids (Fig. 1.1) have been of higher performance compared to the earlier time bulky analog hearing aid devices [3]. The advancement in digital signal processing (DSP) technology [4],

[5], has improved much the quality of these aids, particularly allowing the audiologist to tailor to specific patient needs.

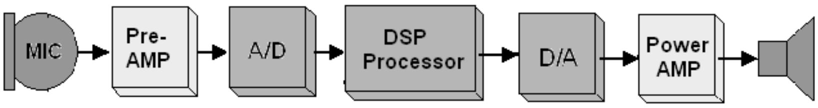

Fig. 1.1 Digital hearing aid block diagram.

Although the digital hearing aids are nowadays commercially available, and have been of several advantages [3], [6], however, they still lack to meet several requirements, particularly in size, battery life, and sound quality [7] as discussed in 2.2.2

A few attempts have been reported in order to solve the existing design problems [7]-[28]. The schemes for developing wireless hearing aid systems have been discussed in [13] and [14]. These include having multi-microphones, radio frequency (RF) circuits, and programmable DSP unit.

1.2. Challenges in Wireless Hearing Aid System Design

Several wireless hearing aid systems have been reported recently. Although as reported, these system are about to provide a better performance to the hearing impaired patients than conventional single-unit based hearing aids, demerits are still found in terms of power-consumption, RF carrier bandwidth and interference vulnerability. Thus, new conceptual architecture of wireless hearing instruments is required for a possible solution to these remaining problems.

Moreover, the cost effective implementation of the wireless hearing devices requires a thorough system level simulation before circuit design and development begin. This is to ensure the functionality of the system and freeze some key block parameter. Furthermore, the system performance can be examined through simulation. Though simulation tools like MATLAB [29] and

PSPICE [30] have been reportedly used by industries for such purpose, they can only work well at block and circuit level, thus can only be used partly in conventional hearing aid design.

The advance features of Advanced Design System™ (ADS) provide comparably more capabilities for simulating electro-acoustic complex systems with DSP such as wireless hearing aids. The ADS provides a fast RF simulation feature and co-simulation with signals of different nature (RF, digital, analog) [31], besides its features for behavioral models. However, no wireless hearing aid system simulation has been reported using ADS so far.

1.3. Objective and Scope

The research work reported in this thesis aims at two aspects concerning wireless hearing aid systems:

1) Propose a single-RF linked wireless hearing system architecture. Under this, a DSP algorithm for noise cancellation is briefly introduced. Theoretical analysis on system noise canceling performance and RF transceiver sub-system are discussed.

2) Perform system simulation on proposed wireless hearing system using ADS 2002C. The behavioral model building is described together with simulation results.

1.4. Organization of Thesis

The thesis is divided into five chapters, starting with introductions in Chapter 1. Background knowledge of both the conventional and wireless hearing aid system is introduced in Chapter 2. Chapter 3 details on the proposed system architecture. It also gives a brief introduction on a noise cancellation algorithm based on a two-element beam-forming technique. Also a theoretical analysis of noise canceling performance at system level is included, together with the analysis on RF transceivers. The ADS compatible models development and schematic presented in Chapter 4 as well as the system level simulation results. Conclusions, together with some suggestions for future work, are included in Chapter 5.

Chapter 2. Conventional Hearing Aid Devices and Wireless Hearing Aid

2.1. Human Ear and Hearing Ability

2.1.1. Overview of human auditory system

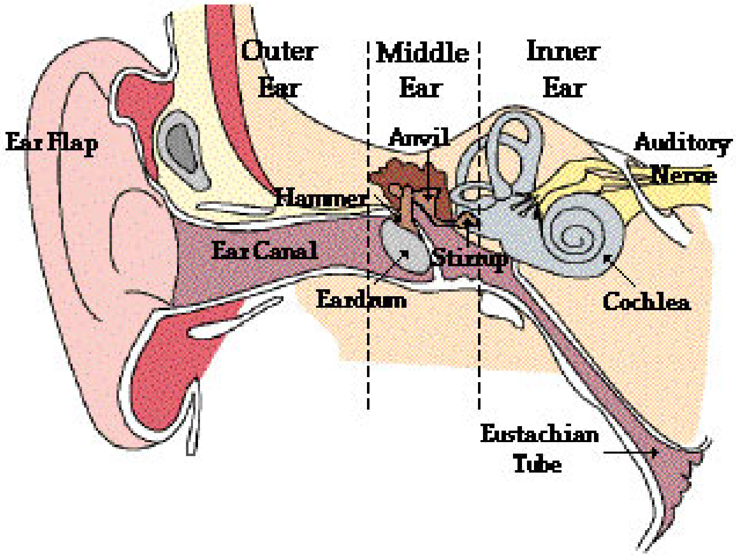

Fig. 2.1 Cross-section view of human ear

(Outer, middle and inner ear with cochlea and auditory nerve).

Hearing is one of the five senses, along with vision, taste, smell and touch. The ear serves as a receiver of incoming sound. It turns the sound from air vibration (mechanical movement) into neural stimuli (electrical signal) and then transmits to central nervous system (CNS) for further interpretation. Fig. 2.1 shows a cross-section view of the human ear. The ear can be divided into three main parts: outer, middle and inner ear. The cochlear and auditory nerve is located in the inner ear. The ear flap of the outer ear acts like a sound collector. Captured sound waves are funneled by the ear flap through the ear canal and strike the ear drum. The middle ear comprises three small bones or ossicles, the malleus (hammer), incus (anvil) and stapes (stirrup). The ear drum, together with the ossicles, transforms air vibration into mechanical movement of these small bones. The middle ear is separated from the inner ear by a bony wall. The movement of the stirrup causes waves of the fluids of the cochlea in the inner ear. As the waves travel down along the cochlea, the cochlear duct moves up and down. This movement leads to the bending of the hair cell's cilia, causing these hair cells to release neurochemicals from hair bases. Below the hair cell is the auditory nerve, which receives the neurochemicals and generates successive neural impulse. The impulses then travel along the axons to the central auditory nervous system (CANS) for sound perceiving.

Among the various parts of the ear, cochlea has its most importance as a transducer between fluid movement and electrical neural stimuli. In engineering terms, the cochlea can be regarded as a series of band-pass filters, each has a specific frequency. Thus the cochlea determines the frequency response of the ear and other important hearing characteristics [32].

The sound intensity is a term used to describe the energy delivered at a given point during a sound. Specifically, this can be expressed in terms of power, pressure, or energy. However, there is a tremendous energy difference between sounds at threshold versus those at upper levels of discomfort. If measured as sound pressure, the difference between the threshold of pain to the softest sound heard is 10 million to one. Thus, sound intensity is measured in decibels. Decibels are referenced to decibel sound pressure level (SPL) in dynes/cm^2. Zero decibels SPL refers to the minimal audible sound of 0.0002 dynes/cm^2, whereas 120 db SPL is equated to 200dynes/cm^2. The formula for dB SPL calculation is as follows:

$$dB\,SPL = 20 \times \log\left(\frac{pressure\ measured}{pressure\ reference}\right). \tag{1}$$

The softest sound intensity is 0 dB SPL, while the loudest sounds is usually set as 120 dB SPL,

The frequency range of a sound wave that human can perceive is between 20 Hz to 20 kHz. The frequency range from 100 Hz to 6 kHz contains most of the information of a human voice and is the most important frequency band.

2.1.2. Hearing Loss Types

Measurement of hearing generally includes measurement of both air-conduction and bone-conduction thresholds. The hearing threshold at a particular frequency is the minimum sound pressure in decibels hearing level (dB HL) required to be perceived. Air conduction refers to sound traveling through air and through the auditory system. The bone conduction refers to sound traveling through the bones of the skull, thereby avoiding the outer and middle ears [6]. Hearing loss is generally indicated by raised thresholds.

Hearing loss can be categorized as four main types:

- Conductive
- Sensorineural
- Mixed
- Central auditory processing

Conductive hearing loss is due to problems in the outer and/or middle ears. In a conductive hearing loss, the air conduction threshold will be raised, yet the bone conduction threshold remains nearly unaffected. As a result, this leads to an air-bone gap (difference between the air conduction and bone conduction thresholds).

Sensorineural hearing loss results from the problem in the cochlea or inner ear. It can be further divided into sensory hearing loss, due to the problem in cochlea, and neural hearing loss, due to the auditory nerve defect. The sensorineural hearing loss can be caused by aging, prenatal or birth-related problems, viral or bacterial infections, heredity, trauma, exposure to loud noises, the use of certain drugs, fluid buildup in the middle ear, or a benign tumor in the inner ear. In the case of sensorineural loss, there will be no air-bone gap while the air conduction and bone conduction thresholds are both raised.

Mixed hearing loss occurs when there are problems both in the outer/middle ear and the inner ear. This results in raised air and bone conduction thresholds, together with an air-bone gap.

Central hearing losses are due to the lesions, dysfunction with the CANS pathway. Central hearing loss mainly results in distortions in the processing of auditory messages rather than the reduced hearing sensitivity as the first three hearing loss types.

The degree of hearing loss can be quantified in Table 2.1.

Table 2.1 Degrees of hearing loss.

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

2.1.3. The effect of hearing impairment

A hearing impaired patient may meet difficulties in his/her daily life. It is necessary to examine what effect the abnormality in human ear has on human listening ability.

1) Reduced speech understanding. A common complaint of people with hearing loss is that with the hearing aid, they can just hear but can not understand. This may due to the poorer supra-threshold processing related to cochlear dysfunction.

2) Frequency selectivity. The people with hearing impairment will have a various frequency based hearing loss. That is, their perception thresholds will be different across the frequency bands.
3) Loudness perception. Hearing impaired people will have a narrower dynamic range to the incoming sounds. The point of hearing impaired at which sounds become uncomfortably loud is about the same for normal listeners. However, the absolute threshold (the perceptible) of sound input is elevated among patients.
4) Temporal resolution. It has been assumed that impaired listeners are less able to perceive high rates of modulation than normal listeners. These patients will meet difficulties in detection of gaps in bands of noise.
5) Noise and speech perception. Individuals with hearing loss of cochlear origin have much greater difficulty in perceiving speech in a background of noise. This phenomenon is called "cocktail party effect", because it is especially difficult for patients to catch desired speech from competing speech and high-intensity background noise as in a cock tail party.

Among the pathological effects listed above, the problem of cocktail party effect plays an important role in the failed use of hearing aid. This issue will be discussed in depth in the next section.

2.1.4. Listening under noise

People with hearing loss meet difficulties in perceiving sounds and understanding speech in both quiet and noisy environment. The commercial hearing aid products have given a promising remedy and most of them perform well to help the hearing impaired listen more effectively when they are in a quiet environment [2]. However, it is clear by now that a person with hearing loss may have a substantially reduced ability to understand speech in background noise and/or reverberations

[33]. Researchers and hearing aid companies nowadays are interested in this research issue. A variety of explanations for the increased difficulty have been given in both physiological and engineering terms.

Audibility

One explanation is simply audibility-based. Much of the performance deficit when hearing impaired listens under noise can be attributed to the masking effects of the background noise in frequency spectrum. Hearing impaired listener may not be able to pick up the important frequency cues of incoming voice due to the existence of the background noise. Compared to quiet environment, it is especially more difficult to understand speech with noise. In some investigations [34], the reduced hearing sensitivity is the only reason necessary to explain performance differences in noise for hearing impaired person.

Squelch effect

Another explanation for the problem of understanding speech in noise is the loss of binaural "squelch" effect. A normal listener always listen binaurally (using two ears simultaneously) in a background of noise. Significant speech-in-noise advantages (SNR) have been reported to be around 6 dB compared to monaural listening (single ear listening). The explanation for the squelch effect is that the brain compares the inputs from each ear and utilizes the slight spectral difference to identify and separate the speech signal from background of noise.

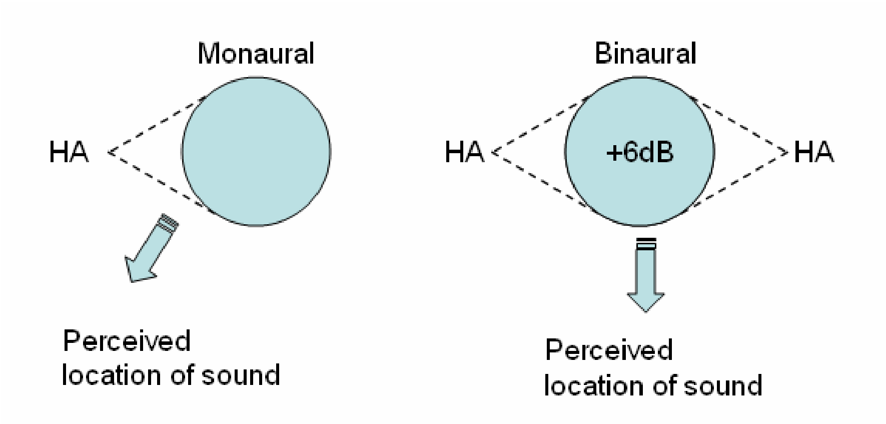

Fig. 2.2 SNR advantage for binaural listening.

In the presence of binaurally asymmetrical hearing loss, the brain does not have access to the same information from the two auditory inputs. Even in the presence of bilaterally symmetrical hearing loss, much of the squelch effect appears to be lost [34]. When the normal binaural input is disrupted, the speech target is more likely to be lost in the background of noise.

Upward spread of masking

Another important explanation to listening in noise is upward spread of masking (USM). It has been observed that in the normal ear, the ability of a low frequency masker to affect high frequency hearing is greater than the ability of a high frequency masker to affect low frequency haring [6] [20] [32]. The masking tendency is thought to be related to basilar membrane function when it is stimulated by two tones of different frequencies simultaneously. Since the traveling wave for low frequency tones is distributed along the entire basilar membrane, it will cause some depression of the membrane in the cochlea where high frequency tones are primarily located. As a result, the low frequency sound wave may "use up" some capacity of the basilar membrane to initiate a neural response for a high frequency tone."

This effect of USM is thought to partially explain the problem associated with understanding speech in noise seen in persons with sensorineural hearing loss. Moreover, it forms the basis of many current attempts to reduce the effects of noise in hearing aid design. That is, apply strategies to reduce low frequency amplification when noise presents. As some researchers argue that low frequency band contains most of the information a speech carries, trade off between reducing noise effect and maintaining speech information shall be carefully handled.

Temporal Smearing

Another explanation for poor performance in noisy situations is the temporal smearing effect. It is assumed that people with sensorineural hearing loss, because of the pathological changes in the auditory system, do not have good discrimination between the timing of auditory events.

In a situation where a listener with normal hearing is attending to a "wanted" speech signal in a background of other "unwanted" speech signals, there is a higher likelihood that the timing of the "wanted" speech signal can be discriminated from the other random events in the "unwanted" speech signal. In the case of sensorineural hearing loss, where temporal abilities have declined due to poor resolution within the auditory system, there is a greater likelihood of an effective temporal overlap between the speech signal and events in the background competition.

2.2. Historical Review on Hearing Aid System

The U.S. Food and Drug Administration (FDA) , for the purposes of labeling, has described a hearing aid as "any wearable instrument or device designed for, offered for the purpose of, or represented as aiding persons with or compensation for, impaired hearing" [35].

Hearing aid using electrical microphone/speaker appeared at the end of 19th century, following the invention of the telephone. These devices are bulky and cumbersome with their carbon granule microphones. A hearing aid design using triode vacuum tube was patented in 1921. Developments in vacuum tube technology allowed portable Body-worn aids to be developed.

The trend to miniaturization and reduction of power consumption was given a huge boost by the invention of transistor in 1947. In 1964, the first behind-the-ear hearing aid using an integrated circuit became commercially available. Driven mainly by cosmetic considerations, the miniaturization trend has continued since the 1960s to present, with current technology providing completely-in-the-canal instruments. In the 1970s, directional microphone and non-linear compression have appeared.

Digital hearing aids became commercially available at the 1990's. With the advanced DSP technology, features such as adaptive filtering, speech detection and automatic gain control have been implemented in commercial hearing instruments since the end of last decade.

2.2.1. Hearing aid types

According to the fitting position and function, hearing aids can be categorized into seven main types:

- Body worn (BW)
- Behind the ear (BTE)
- In the ear (ITE)
- In the canal (ITC)
- Completely in the canal (CIC) (Fig. 2.3)
- Middle-ear implants (Fig. 2.4)
- Cochlear implant (Fig. 2.5)

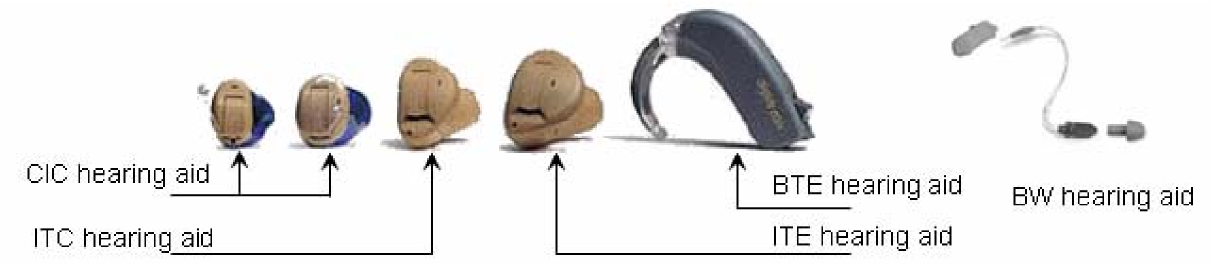

Fig. 2.3 Five types of hearing aids.

Fig. 2.4 Middle ear implants (Soundtec, Inc).

Fig. 2.5 Cochlear implant (Med-El®)

According to the different sound conduction methods, hearing devices can also be categorized into air-conduction and bone-conduction aids. While most of commercial hearing aids are air-conducted, the bone-conduction aid has been used for patients with conduction hearing loss or gross occlusion of the ear canal while surgery is deemed inappropriate. This aid differs from air-conduction aid only at the receiver that delivers mechanical vibration to the skull. As a result, the bone-conduction aid is able to bypass the middle ear and reach the cochlea effectively.

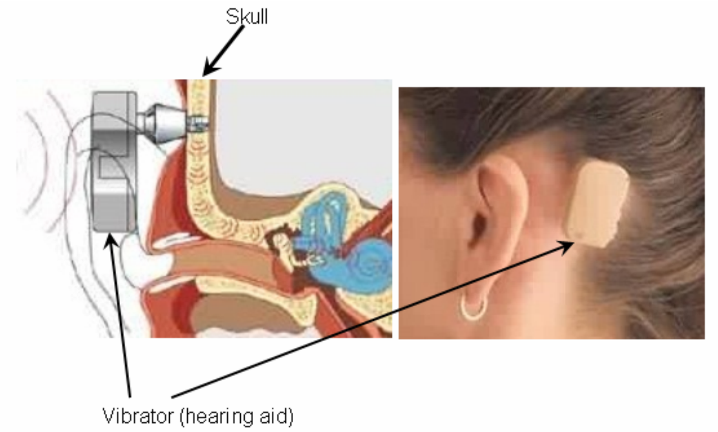

Fig. 2.6 Bone conduction hearing aid (BAHA® Bone Anchored Hearing Aids).

The third category is based on the distinction between conventional analogue and digital hearing aids. In an analog hearing aid (Fig. 2.7), the continuous time signals from the microphone are processed as a continuum, with no discretization in time or quantization of amplitude. In a fully digital hearing aid (Fig. 2.8), the continuous time signals from the microphone are filtered to reject frequencies outside of the required range. The signals are then sampled, converted and processed as a stream of binary numbers in a central DSP unit. The processed data is later returned to a continuous time signal by the combination of a digital-to-analog converter (DAC) and an anti-aliasing filter and output through a speaker.

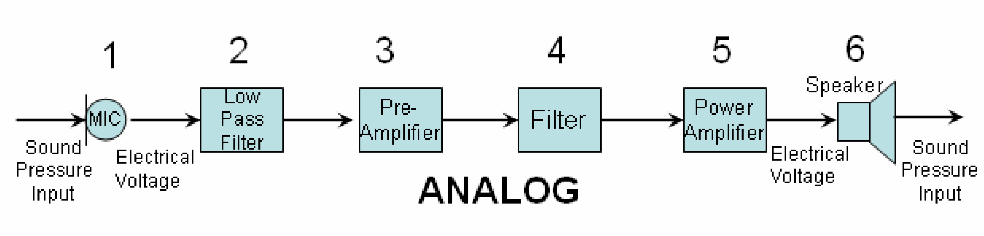

Fig. 2.7 Analog hearing aid block diagram.

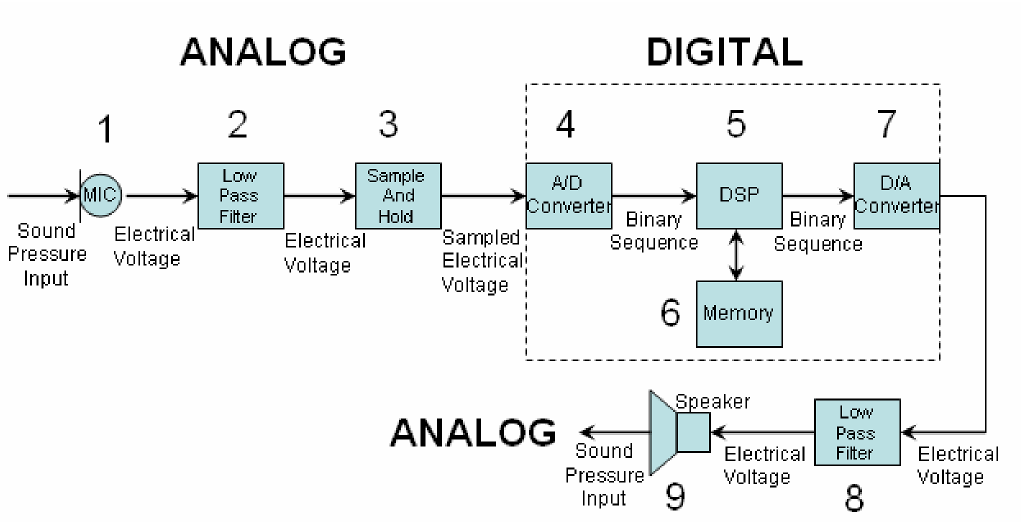

Fig. 2.8 Digital hearing aid block diagram.

2.2.2. Current research issues in hearing aid design

<u>Battery Life and Power consumption</u>

Battery life is a crucial characteristic of hearing aid devices. Since hearing devices are switched on all the day by patients, they are expected to have a long working life. However, since the processing speed of DSP chips is increasing dramatically, so does its power consumption. As a result, the implementation of complex DSP algorithm in hearing instruments is limited. Currently, most of the researches are focused on either high-performance hearing aid batteries or reducing power consumption of the hearing systems. With the advanced semiconductor technology, power cost of a hearing device is expected to be minimized [6], [8].

Size and Portability

Based on cosmetic consideration, hearing aid systems invisible to others like CIC or ITC hearing aids are more acceptable to the hearing impaired people nowadays. Since it is placed deep in the canal and is much closer to ear drum, the requirement on system gain of a CIC aid is less stringent. Thus less power will be consumed. As the other side of a coin, reduced size leads to less power supply due to battery constraints and it brings problem to maintain complex DSP algorithms. While highly integrated circuit is under development to realize complex functions as many research works, separating redundant components from ear-piece to a body unit can be another choice [6].

Noise and echo cancellation

Currently, one of the major issues in the design and development of hearing aids is reverberation/noise cancellation using DSP [3] [5]. In a confined environment, sound perceived is often the mixture of the original signal, a number of echoes/reverberations reflected from different directions and the sound noise in the proximity. Hearing difficulty is further worsened when reverberation from environment interference is present. The situation gets more complicated when cross-talk speech and background noise exist [3], [6], and [14]. Conventional hearing aids which amplify all inputs and/or do simple filtering do not perform well especially for the cases of significant hearing loss. As a consequence, a few noise cancellation algorithms have been implemented in digital hearing aid devices [11], [14]. However, the requirement on low size and low power consumption in earpiece prohibits the noise canceling DSP algorithm insertion in the earpiece [14], [23], [26], and [36], Also, because many noise cancellation algorithms have adopted multi-microphone array which is space-consuming [11], [14], difficulties are met in introducing them into conventional hearing devices. Advancements in technologies such as IC design, wireless digital technology and DSP technique allow wireless ring aids having multi-microphone, RF circuit and DSP unit to a promising solution [13], [14] by linking the earpiece wirelessly to a body unit.

Auto Gain Control (AGC)

Conversational speech sounds vary from 65-70 dB SPL for the low frequency vowels and diphthongs, while the consonants may be as much as 30 dB lower in intensity. Speech may be embedded in a background of noise as much as 20 to 30 dB higher. The result is that many impaired ears do not have enough residual hearing ability to discriminate a variety of speech cues [3].

The reduced dynamic range of the impaired ear may be matched to that of the “normal” ear by a non-linear compression scheme. More over, AGC scheme also attempts to perform fast reduction of gain in response to sudden large increases in sound level and to restore the gain quickly when the loud sound has ceased.

The compression functions in the most recent digital aids may be combined and distributed through the signal-processing chain and many multi-band digital hearing aids now apply compression schemes either independently to groups of frequency bands or dynamically link the compression functions across neighboring bands.

Frequency shaping

Hearing instruments must be able to separate incoming signals into different frequency regions to compensate for the difference in the frequency configurations of hearing impairment. The wideband input signal is separated into frequency bands, typically done with a bank of filters. The filter bank is characterized by the number of output frequency channels, the crossover frequencies between adjacent channels and the steepness of the filter slopes [12] [20]. The more frequency channels, and the steeper the filter slopes are, the finer the control of the signal manipulation in later processing can be. The frequency response of hearing instruments can be adjusted by the gain in each individual channels.

Binaural Configuration

Although traditional single audio output setup is quite common, binaural listening has been strongly recommended by the clinicians [3] [34]. Keep two ear hearing is crucial to improve the patient's ability of locating the sound source easily and prevent deterioration of the unaided ear.

The low frequency component (below 1 kHz) of audio signals arriving at both ears is important to speech reception. The interaural delay that arises from spatial separation of the ears is largely sufficient for providing signal detection and speech reception advantages. The received audio signal can also be separated to two parts. One is for determination of location the signals, the other can be used for noise cancellation [11]. For hearing aid design, not only the target-to-jammer ratio should be increased, but also the location information of the voice should be extracted, which is beneficial for the binaural configuration.

2.3. Noise Cancellation Methods

Among all the issues related to modern hearing aid, the noise cancellation method is considered to be the most helpful to the people with hearing impairment. Several methods, different from their nature, have been developed to cope with this issue.

2.3.1. Single omni-directional microphone

Many single-unit based hearing aids have an omni-directional microphone as input. The structure diagram of an omni-directional microphone is shown in Fig. 2.9. It has one sound inlet and signals are processed equally regardless of azimuth. Thus, the conventional hearing aids are not able to distinguish background noises according to incoming azimuth.

However, improvements have been made on such systems. Researchers and manufacturers have implemented DSP algorithms in digital hearing aids for noise cancellation. While no additional information about signal and noise source can be obtained using omni-directional microphone, the differences between signal characteristics have been utilized to extract speech out of noise. Some of the algorithms include frequency spectrum analysis [37] and wavelet transforms [38].

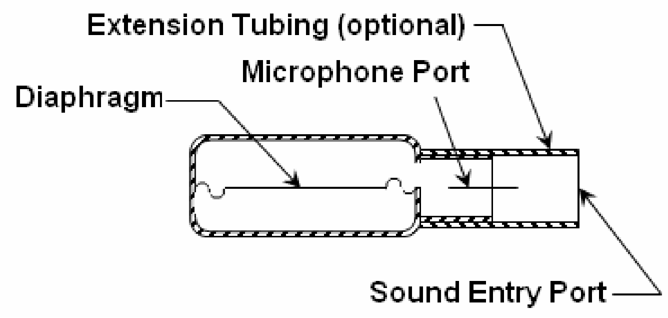

Fig. 2.9 Schematic drawing of an omni-directional microphone (side view)

2.3.2. Directional microphone

Another method for noise reduction in many current commercial hearing devices is using a single directional microphone for voice pick-up. By inhibiting background noise, SNR can be increased (Siemens, Unitron Hearing).

This microphone has two sound inlets (front and back), divided by a diaphragm. The diaphragm senses differences in air pressure between the two inlets. An acoustical time delay network is placed in the rear inlet. Equivalent sound pressure on opposite sides of the diaphragm simultaneously results in sound cancellation because the diaphragm can not move. Therefore, engineers can design different directional microphone patterns by adjusting the delay.

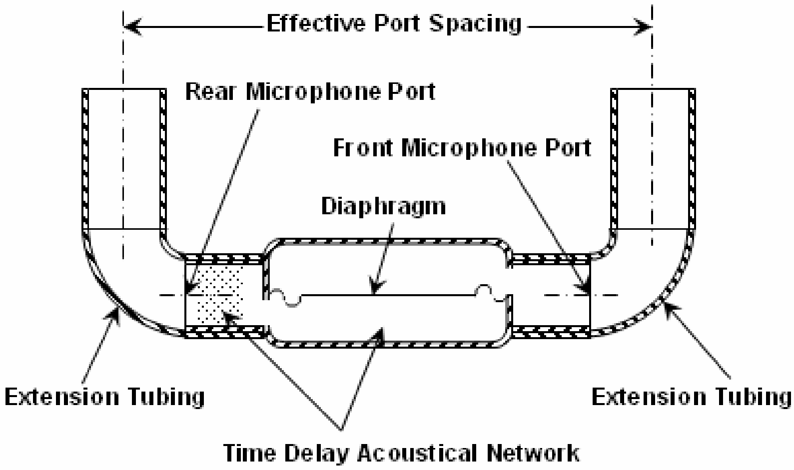

Fig. 2.10 Schematic drawing of a directional microphone structure (side view).

2.3.3. Microphone array

Due to the fact that interference often overlaps in the frequency domain with the desired speech, the single microphone setup is not sufficient. Current researches are focused on using more than one microphone, especially on dual-microphone setups.

The principle is that by using more than one microphone, the system obtains more information on both the desired speech and undesired noise [39] [40]. Although the internal noise source of each microphone may add up, it is still possible to extract the desired signal from the inputs by signal processing algorithms. Adaptive filtering is used as fundamental method in these studies. Some researchers also use an estimator to estimate the noise then cancel the noise from the original signal [13]. However, currently there are few commercial products implementing a mature multi-inputs signal processing technology [33] [41].

2.4. Noise Cancellation Performance and Space/Power limitation

The potential future power of DSP hearing aids is very great. However, in reality, severe technical limitations have so far prevented practical implementation of generalized DSP functions in ear-level hearing aids.

According to the National Statistics of U. K., The use of a hearing aid does not necessarily solve patients' hearing problems. Table 2.2 shows that 62% of the people wearing an aid reported continuing problems with their hearings in U. K. Other data suggests that 25% of people who own hearing aids do not wear them due to the problem of background noise still occurred.

Table 2.2 Whether hearing problems continue when wearing hearing aid by age.

Age	Percentage who continue to have hearing problems with an aid in U.K. [1]
16-44	70
45-64	65
65-74	60
75 and over	62
Total	62

Further improvements of digital hearing aid in sound quality and noise suppression are limited by power and size of conventional single-unit hearing aid. General purpose DSP circuits are currently available in digital hearing instruments; however, even so-called "low-power" off-the-shelf circuits operate at a minimum of three volts and may require a supply current of up to 150 milliamps [3]. This is several orders of magnitude above the 1.2 volt supply and 1.0 milliamp current drain available from an A13 zinc-air battery. Table 2.3 shows capacity of hearing aid battery in the market. The limited power supply of hearing aid systems makes the integration of DSP unit difficult.

The problem of size is strict when placing all the necessary power supply and support circuitry into the small space available inside a single-unit based hearing aids. Off-the-shelf DSP circuits are not available that will fit the sub-miniature requirements of current advanced hearing aid packaging.

Table 2.3 Hearing aid battery capacity in the market.

Battery model number	Hearing aid type	Capacity / mAH
A675	BTE	600
A13	BTE/ITE	260
A312	ITE/ITC	150
A10	ITC/CIC	80
A675P	Cochlear	520

2.5. Wireless hearing aid instruments (Prior Art)

With the requirement of clear voice, multi-function, etc, digital signal processing begins to play a key role in a hearing aid. The need of the hearing aids with noise/echo cancellation feature is also highly appreciated by the hearing impaired patients. All these impose a higher demand on capacity, power cost of the DSP unit in the system. However, using DSP chip in earpieces is intuitively not the best choice due to the limiting size and battery power.

To solve the problem, there are two alternatives: (1) by simplifying the algorithm, the DSP chip can be replaced by a few numbers of basic digital components [16], (2) specially designed DSP chip for hearing aid with desired low supply voltage and power consumption [5] [18]. Even with these methods, it is also a challenge to build a hearing aid with small size and low power consumption.

2.5.1. Basic Concept of wireless hearing aids

Separating the hearing aid into a body unit and an earpiece has become a better choice for the problem above mentioned. The limitations of size and power usage can be bypassed at the cost of enhanced design complexity. Wireless hearing aids are usually comprised of at least 2 basic parts: one body unit and one/two earpiece/s. Radio frequency links are used to achieve communication between the body unit and earpiece. The DSP unit is built in the body unit, responsible for most of digital signal processing algorithms. Data exchange is performed between body unit and earpieces.

The advantages of wireless hearing aids are discussed below:

1) Much less power limitations exist in the body unit. The released space requirement makes the choice of more powerful battery feasible other than existing hearing-aid batteries. Subsequently, it frees the power limitation imposed on DSP unit. Designers can perform more complex algorithm on noise/reverberation cancellation in the system.
2) More circuits can be built in body unit with less consideration on circuit area than in traditional single-unit based instruments. For integrated circuit (IC) designers, the circuit area in a hearing aid is very small and thus tradeoff between circuit area and system performance is often a key issue in R&D work. With the separate body unit, more function and high performance circuit components can be added and help boost the system performance.
3) Multi-microphone array which proves to be noise-canceling effective but bulky now can be integrated into hearing aid system. Because many noise cancellation algorithms have adopted multi-microphone array which is space-consuming [11], [14], difficulties are met in introducing them into single-unit based devices.
4) Integration with mobile electronic devices (etc. pager, hand phone) is easily realized. The body unit can also be easily integrated with any audio devices or portable physiological telemetric instruments [9], [10].

The wireless hearing aids also have some demerits which are listed below:

1) Enhanced RF transceivers design complexity. In order to establish wireless link between units, RF transceivers are built in earpiece which has a very stringent power and size requirement. Thus, ultra low power and low noise RF circuits are required for optimal system performance. This enhanced circuit design complexity is expected to be solved by

current fast-developing IC technology.

2) No comprehensive system level simulation across the entire system has been reported to support design works. A system level simulation is not only economic but also mandatory for designers to examine the proposed system performance. However, no comprehensive simulation methods have so far been reported and the feasible simulation tool remains unclear.

2.5.2. Prior Art

Until now, there are two typical wireless hearing aid systems which have been reported. They will be introduced and comparison between conventional single-unit hearing instruments and these wireless aids will be made as follow.

In 2001, B. Widrow [14] described a wireless hearing aid system, utilizing a T-coil inside the earpiece to pick up audio frequency magnetic field generated by a neck loop as shown in Fig. 2.11. A six-microphone array placed on the user's chest picks up and filters input signal using adaptive filter technique. Processed speech then drives the neck-loop to generate electro-magnetic field. This audio-frequency electro-magnetic field is then picked up by the T-coils in the earpiece wirelessly. Since this magnetic field is within the audio-frequency range, it is considered a non-RF communication. Thus, the bandwidth efficiency and invulnerability to communication noise are low due to its simple "neck-loop to T-coil" structure. Also, the bulky body unit holding the 6 microphone array is another disadvantage of this hearing aid system.

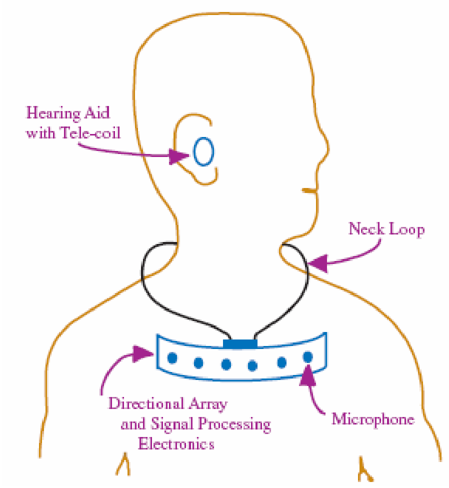

Fig. 2.11 B. Widrow's neck-lace wireless hearing aid.

Another concept of using two RF communications is introduced in 2003 [13] as shown in Fig. 2.12. The whole system comprises two earpieces and a body unit. A bi-directional 8-ary RF modulation scheme is suggested. A BiCMOS implementation of two-receiver and two transmitters based RF communication between each earpiece and the body unit has been reported. The microphones in both earpieces pick up input sounds and transmit them down to the body unit. Processed signal is then sent back to the earpieces, and later, to the patient. However, the whole system architecture based on above has not been implemented as reported, especially for the noise cancellation block that is very crucial to the overall performance. A possible system structure based on the concept given in [13] is configured as shown in Fig. 2.13. For this system with duplex RF link, building a RF transmitter inside earpiece along with a RF receiver will tremendously increase the earpiece power consumption, shortening the battery life. Besides that, the system employs four RF links, increasing design complexity while keeping the bandwidth efficiency low.

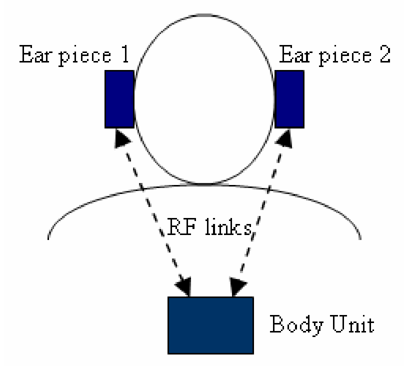

Fig. 2.12 Duplex RF hearing aid system configuration.

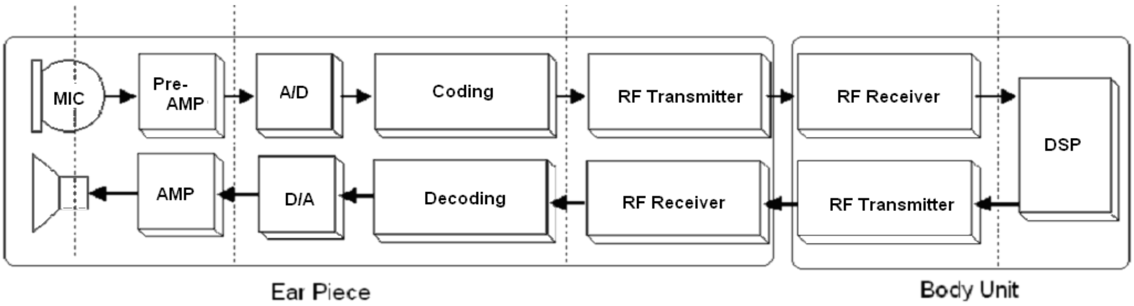

Fig. 2.13 Block diagram of the duplex RF hearing aid (summarized from [13]).

Table 2.4 compares, at the conceptual level, a single-unit based hearing aid, the two wireless hearing aids mentioned above, and our proposed new architecture. It can be seen clearly from the table that conventional digital hearing aid keeps the lowest power consumption in earpiece with the simplest signal process scheme, while the proposed hearing aid system keeps a comparable battery life time, a low RF bandwidth with a complex beam-forming noise cancellation scheme. The battery life (288 h) of proposed system is shorter than the conventional hearing aid due to the power dissipation of noise cancellation unit and RF ends. It can be improved by rechargeable battery option. A lower RF bandwidth of 200 kHz compared to other wireless hearing devices reduces inter-device interference and saves bandwidth resource as per regulated.

Table 2.4 Architectural level comparison of hearing aids (HA)

Feature	Conventional Digital HA	Neck-Loop Wireless HA	HA with Duplex RF Link	Proposed Wireless HA
Ref/Year	[16]/2002	[14]/2001	[13]/2003	This work (Fig. 3.1)
IC technology	0.6μm CMOS	NA (not available)	0.8μm BiCMOS	0.18μm CMOS
Supply voltage	1.1V	NA	NA	1.2V
Working Current in RF receiver	Not applicable	NA	667μA	2.2mW
Power Cost in earpiece	270μW	NA	NA	2.5mW
Battery Type	NA	NA	NA	A675
Battery Life	700h	NA	NA	288h
No. of units	One	Two	Three	Two/Three
No. of microphones	One	Six	Two	Two
Noise Cancellation	Sub-band filtering	Adaptive filter	Nonlinear estimator	Beamforming
Wireless link	Not applicable	Unidirectional	Bi-directional	Unidirectional
Total RF bandwidth	Not applicable	NA	800kHz	200kHz

2.5.3. System Simulation on wireless hearing instruments

Another critical factor of wireless hearing aid design, besides the above, is about the simulation of wireless hearing systems. The complexity exists as it comprises various internal blocks of different functions, which process the signals of different types, e.g. analog, RF, digital signals as depicted in Fig. 2.13. Using electro-acoustic transducers and their interfaces with electronic circuits increase the design challenges. Inherent problems associated with inter blocks interfacings makes it a more uphill task. Thus, a computer aided simulation based on behavioral model before physical realization is cost effective and necessary to ensure the functionality of the system, apart from specification freezing for sub-block design.

As an option for behavioral modeling and system simulation, computer aided design (CAD) tools like Pspice and MATLAB have been widely used [29], [30]. However, the available features of these tools limit their suitability for simulating a complex system of above type. Although SIMULINK in MATLAB is powerful in DSP and behavioral modeling, it does not provide adequate help in IC circuit design and RF communication. Pspice has been chosen [30] for circuit system simulation, which is good at analog circuit analysis but helps little in incorporating DSP algorithms. A comprehensive system simulation based on behavioral model using ADS [31] is introduced in the following chapters as a solution to above problem of simulating wireless hearing aids.

Chapter 3. Proposed Concept and Theoretical Analysis

In this chapter, methods detailing the proposed architecture, two-element beam forming algorithm for noise cancellation, overall noise analysis with SNR improvements and basic analysis on RF transceivers are to be elaborated.

3.1. Proposed Wireless Hearing Aid Architecture

The proposed system concept is shown in Fig. 3.1. It applies only one way RF communication, from body unit to the earpiece. The whole system requires a body unit equipped with two microphones, preprocessing circuits, DSP unit and a RF transmitter. Output through the RF transmitter transfer processed signal from the body unit to the earpiece. It leaves an option for the audiologist to link it with one or two earpieces according to patient's hearing loss degree and budget.

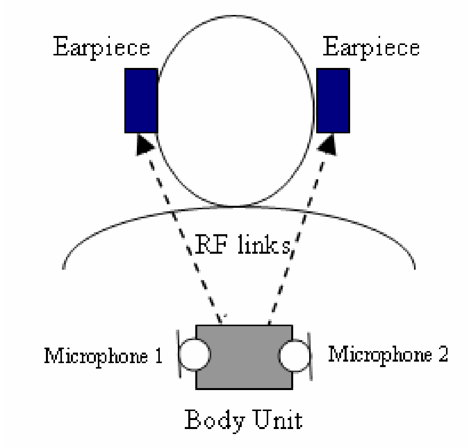

Fig. 3.1 Proposed RF hearing aid system configuration.

The proposed system structure diagram based on above is shown in Fig. 3.2. Input sound is first received by two microphones placed at a distance in the body unit. After pre-amplification and A/D conversion, input signal is sampled as bit stream, processed in the DSP block to attenuate unwanted speech and noise using beam-forming technique, and then transmitted to the earpiece via RF link. It is further passed through several stages in the earpiece which includes down conversion, filtering, amplifications, and output buffer and converted to the sound waves through earphone in the case of hearing aids or to the stimuli on electrodes in case of Cochlear Implants (CI) [8], [42].

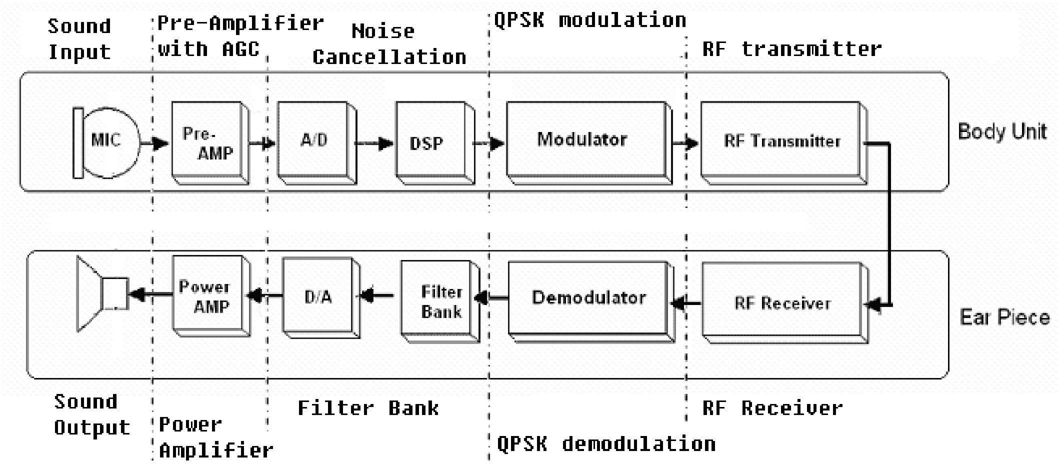

Fig. 3.2 Proposed wireless hearing aid system structure.

Wireless RF link is established from body unit to earpiece and there is no need of a RF transmitter in earpiece unlike the case of [13]. The proposed architecture is amenable to CMOS implementation. A CMOS RF receiver at 900MHz can generally have power consumption as low as 2.2mW [43]. An estimated total power consumption of the earpiece based on 0.18μm CMOS is given in Table 2.4. It shows the user will need to change the battery approximately once in a month based on typically 10hrs usages per day [16], which is comparable with most of the hearing aids in the market.

In the proposed hearing aids Fig. 3.2, a dual-microphone scheme is used unlike the usage of a complex multi-microphone array as in [14]. This allows the body unit to be designed in a compact manner in order to be incorporated into the body-worn electronic devices, e.g. the mobile phone and pager [10]. Trade-off between portability and device size can thus be reached. The DSP block is built inside the body unit to eliminate unwanted noise and echoes. Either general purpose DSP chips or specially designed DSP circuit can be used. A quadrature phase shift keying (QPSK) digital modulation is selected in this design for its balance between power consumption and bandwidth efficiency [44].

Since there is only one-way data transfer from the body unit to the earpiece, no RF transmitter is needed in the earpiece. This system is also DSP compatible and is expected to be realized using cheap CMOS technology. The most power-hungry parts (DSP block and RF transmitter) are implemented in the body unit, so as to increase battery life of the earpiece while enabling a better noise-cancellation performance. The noise cancellation algorithm in the body unit plays a key role in attenuating incoming noise/reverberations and improving the speech intelligibility. Apart from these benefits, possible integration of wireless hearing aid into portable electronic devices becomes possible.

3.2. Beamforming DSP Algorithm for Noise Cancellation

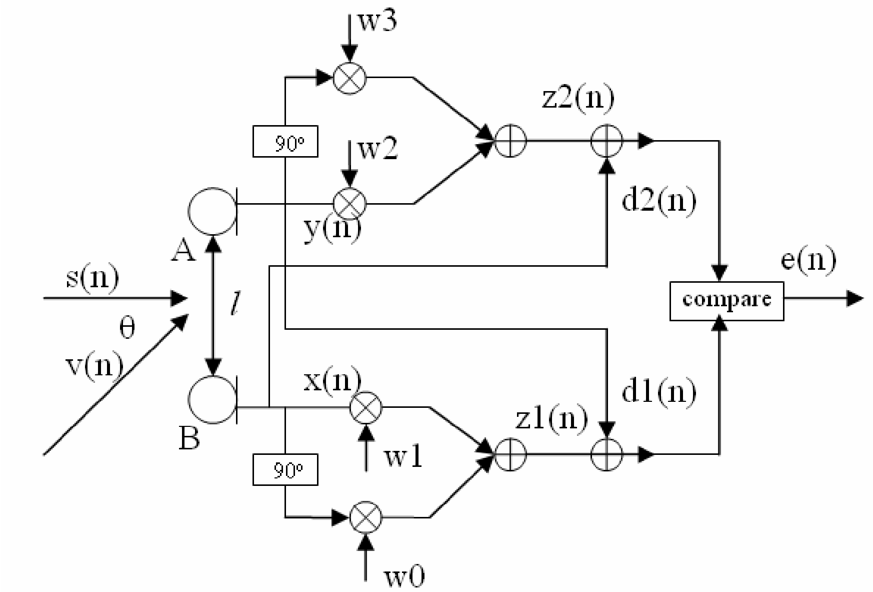

Fig. 3.3 Block diagram of two-element beam-former [36].

Noise cancellation unit is the key block in our proposed system for signal-to-noise-ratio (SNR) improvement. As it is built in the body unit, the size limitation and power constrain have been remarkably alleviated. Thus, general purpose DSP processor can be used for its programmability and flexibility although it may consume more power than a specially design DSP chip [5], [18]. It is worth to state here that the term noise cancellation refers to the cancellation of the unwanted sound signals due to reverberations, echoes, cross-talks, interferences.

The beam-forming method used here as shown in Fig. 3.3 is reported in [36] by our group. It is based on the constrained adaptive beamformer of Griffiths and Jim [45] and found amenable for DSP implementation in hearing aid applications. The signal $s(n)$ represents the desired voice, and $v(n)$, the competing speech signal. Two omni-directional microphones A and B pick up input signal and noise from different directions. It is assumed that the desired speaker is always in front of the hearing aid user. Thus, desired speech always comes from the direction straight ahead. Sounds from any other directions are deemed "noise" and attenuated for better sound quality. The angle between signal and noise is represented by θ. These signals can be expressed as below:

$$v(n) = \beta(n)\cos n\omega_o, \tag{2}$$

$$s(n) = \alpha(n)\cos n\omega_o, \tag{3}$$

where $\alpha(n)$ and $\beta(n)$ are narrow banded base band signals, ω_0 is the center frequency of $s(n)$ and $v(n)$.

The time delay when $v(n)$ reaches A compared to B is expressed as δ_o. It is described as:

$$\delta_o = \frac{l\sin\theta}{cT}, \tag{4}$$

In Fig. 3.3, the distance between A and B is l, and it is set to be half length of the center frequency signal input. It can be calculated as follow where center frequency f_{center} is set to be 3 KHz [3], [34].

$$l = \frac{c}{2f_{center}} = \frac{334m/\sec}{2 \cdot 3KHz} = 5.57cm \ . \tag{5}$$

This theoretical length shows possibility to integrate the body unit into most of the portable electronic devices (e.g. Mobile phone), because the distance between two microphones determines the minimal size of the body unit.

The output $e(n)$ can be expressed by [45]:

$$e(n) = \frac{\cos n\omega_o - \cos[(n-\delta_o)\omega_o]}{\sigma_\alpha^2 + \sigma_\beta^2}\left[\sigma_\beta^2\alpha(n) - \sigma_\alpha^2\beta(n)\right], \tag{6}$$

where σ_α^2 and σ_β^2 are the variance of $\alpha(n)$ and $\beta(n)$.

Further analysis shows that system output signal-to-noise spectrum density ratio is reverse proportional to the input signal-to-noise spectrum density ratio, which have been used for SNR improvement in wireless hearing aids. The upper half of the beam-former is symmetric to the lower half and helps to build a symmetrical directivity of the beam-former, the detail of which is given in shown in [36].

3.3. System Noise Analysis/SNR Improvement of Proposed System

As the performance of the beam-former mentioned above is influenced by other blocks when integrated into the proposed wireless system, it is worth to look into the SNR performance of the entire system rather than that of a single DSP block.

Traditionally, evaluation on hearing aid system noise is done by measuring the signal-to-noise-ratio at system output [46]. This is based on the assumptions that all sound input is considered signal, and the sound quality is degraded mainly due to the analog circuit noise [47].

However, with the introduction of DSP unit for noise cancellation, the incoming sound can no longer be considered pure signal. The input speech is always accompanied by unwanted competing speech/reverberation. And only the desired speech is valuable to the patients. For better improvement, competing speech and noise/reverberation need to be attenuated. Besides that, the system internal noise originates not only from the analog circuits, but also from the A/D (Analog to Digital) quantification and RF data transfer errors, which differs in nature and signal property. By far, there is no research analysis on the SNR performance across such a complex system with DSP noise cancellation unit, RF communication systems and analog/digital circuits.

In order to resolve the problem of SNR measurement, noise factor (NF) is used to examine the SNR improvement at system level. The noise factor is defined as $NF = SNR_{in} / SNR_{out}$. SNR_{in} and SNR_{out} are the signal-to-noise ratios measured at the input and output, respectively. NF can also be expressed by internal noise and input noise power as analyzed in [48].

$$NF = \frac{N_{\text{int}ernal}^2}{N_{sys,in}^2} + 1 \tag{7}$$

In (7), $N_{\text{int}ernal}^2$ is the total input referred circuit noise power of a specific stage, $N_{sys,in}^2$ is the total noise power at system input due to the unwanted speech signals. By using the definition of noise factor, noise factor of a cascaded system can be calculated from noise factors of separate blocks [48].

$$NF_{system} = \frac{SNR_{input}}{SNR_{output}} = NF_1 + \frac{NF_2 - 1}{G_1} + \frac{NF_3 - 1}{G_1 G_2} + \ldots + \frac{NF_n - 1}{G_1 G_2 \ldots G_{n-1}} \tag{8}$$

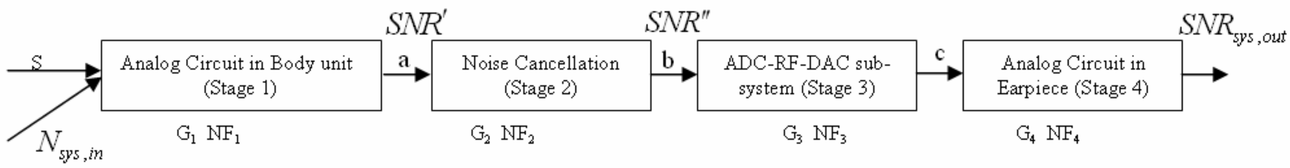

Fig. 3.4 Simplified block diagram of proposed system for SNR analysis.

The noise factors NF_1, NF_2 and so on are of the corresponding stages with respect to system input noise and can be calculated using (7). The power gains of separate blocks are denoted by G_1, $G_2,\ldots,G_{n-1}$.

For the convenience of our analysis, the system is divided into four main stages as shown in Fig. 3.4 according to the different nature of noise sources in each stage. Stage 1 is the analog circuit in the body unit, generating analog circuit noise. Stage 2 is the noise cancellation unit. It is assumed that the effect of quantizing noise caused by the A/D converter can be ignored in stage 2 as it is not a dominating source in this stage. Stage 3 is composed of A/D converter, RF transmitter, RF receiver and D/A converter. The noise in Stage 3 is mainly due to the quantization and the data transfer error during RF data transmission [49]. Stage 4 represents the analog circuit inside the earpiece, including the output buffer and the earphone. NF_1 to NF_4 are the noise factors and G_1 to G_4 are the unloaded power gains of each stage separately. The matching between each stage is assumed to be perfect.

The noise factors of stage 1 and stage 4 can be expressed using (7) respectively.

$$NF_1 = \frac{N_1'^2}{N_{sys,in}^{\;2}} + 1, \tag{9}$$

$$NF_4 = \frac{N_4'^2}{N_{sys,in}^{\;2}} + 1, \tag{10}$$

where N_1' and N_4' are the input referred circuit noise of stage 1 and stage 4.

Stage 2 represents the noise cancellation unit. The noise attenuated/cancelled by this stage is represented by N_2'. Thus, NF_2 for stage 2 can be defined as:

$$NF_2 = \frac{N_2'^2}{N_{sys,in}^{\;2}} + 1. \tag{11}$$

Since N_2' varies with input SNR as analyzed in section 3.2, NF_2 reflects the variations in input

SNR. Thereby, the expression for NF_2 is derived below in term of SNR. The input-output SNR relation for stage 2 can be given as:

$$\Delta SNR_2 = \frac{SNR'}{SNR''} = \frac{N_2'^2}{N_{2,in}{}^2} + 1, \tag{12}$$

where SNR' and SNR'' are SNR at the input and output of stage 2 as shown at node a and node b in Fig. 3.4. $N_{2,in}$ is the noise at the output of stage 1 , which is given by:

$$N_{2,in}{}^2 = (N_{sys,in}{}^2 + N_1'^2) \cdot G_1 . \tag{13}$$

Substitute (13) for $N_{2,in}$ in (13), N_2' can be expressed as:

$$N_2'^2 = (\Delta SNR_2 - 1) \cdot (N_{sys,in}{}^2 + N_1'^2) \cdot G_1 . \tag{14}$$

Thus NF_2 can be expressed by substituting $N_2'^2$ in (11) with (14) as follow:

$$NF_2 = (\Delta SNR_2 - 1) \cdot NF_1 \cdot G_1 , \tag{15}$$

where ΔSNR_2 varies with the system input SNR and can be simulated using ADS as shown in Fig. 4.19 (curve b).

The relation for the noise factor of stage 3, NF_3 is derived as follow: the internal noise power due to the quantizing error and data transfer error can be derived as follow [49]:

$$N_3'^2 = \frac{V^2}{SNR_{pk,out3}} , \tag{16}$$

where V is the designed peak value of A/D converter and $SNR_{pk,out3}$,the peak signal to average noise ratio at output, is given by [49]:

$$SNR_{pk,out3} = \frac{3M^2}{1 + 4(M^2 - 1)P_e} . \tag{17}$$

In (17), M is the quantizing level of the AD converter. P_e is the error probability due to RF transmission. P_e is affected by transceiver specification and transmission channel noise.

Thus, from (16) and (17),

$$NF_3 = \frac{N_3'^2}{N_{sys,in}^2} + 1 = \frac{V^2}{N_{sys,in}^2 \cdot SNR_{pk\,out3}} + 1. \tag{18}$$

Using (8) , (15) and (18), the noise factor for the whole system comprising stage 1 to stage 4 as shown in Fig. 6, can be expressed as:

$$NF_{system} = \frac{SNR_{in}}{SNR_{out}} = NF_1 \cdot \Delta SNR_2 + \frac{V^2}{N_{sys,in}^2 \cdot SNR_{pk,out3} \cdot G_1 G_2} + \frac{NF_4 - 1}{G_1 G_2 G_3} \tag{19}$$

Since $NF = SNR_{in} / SNR_{out}$, in order to increase the output SNR as much as possible for better noise cancellation, the NF_{system} needs to be higher. The analysis in this section helps designers have an in-depth understanding on the system noise and its relationship with gain and NF of several intermediate stages. By investigating equation (19), a few guidelines based on above are provided below in order to optimize the system performance:

1) The first term $NF_1 \cdot \Delta SNR_2$ is most dominating as the gain of this stage does not help minimize its effect.
2) Increasing ΔSNR_2 help maximize the NF_{system}. It is worth to say that ΔSNR_2 directly depends on input SNR and the method of noise cancellation adopted.
3) The effect of the second term can be minimized by optimizing M and P_e as it is inversely proportional to $SNR_{pk,out3}$.
4) The last term is ignorable as it is inversely proportional to the gain. So the performance of blocks in the earpiece is believed to be less important in noise performance of the whole system.

It may, however, be noted that the (19) provides a NF analysis for such a complex wireless hearing aid system comprising not only the gain stages but also the noise canceling units, modulation/demodulation and RF transmission. The NF for the systems simply comprising of cascaded gain stages is given in (8). Analysis makes it clear that (19) provides deeper insight in designing the complex system of the nature of as shown in Fig. 3.2.

3.4. RF Transceiver Analysis

Analysis on RF transceivers and propagation channel before building behavioral model is essential to guarantee that the simulation is carried out correctly. In this section, brief analysis and some parameter calculation are performed to deepen the understanding of the RF transceiver design before going into actual system simulation.

3.4.1. Propagation channel simulation

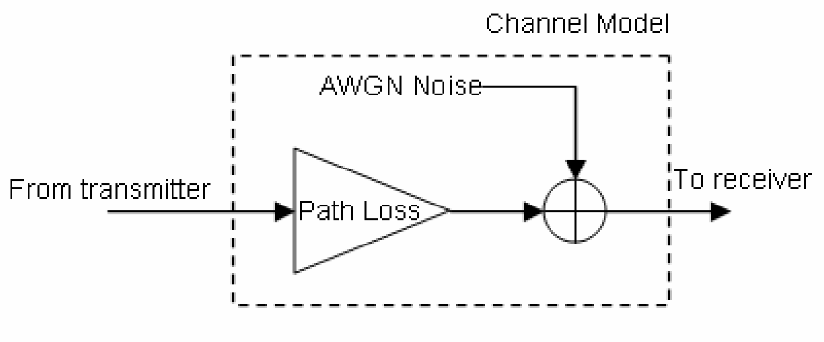

Fig. 3.5 Communication channel model.

It is assumed that the distance between body unit and earpiece is kept within 1 meter. Since the location of these two units do not change when patient is moving, and there is no obstructions between the transmitter and the receiver, a single-path, free space propagation model is expected to accurately describe the channel behavior.

The RF signal power at receiver input can be expressed as follow,

$$S_r = S_t G_t G_r \left(\frac{\lambda}{4\pi d} \right)^2 , \tag{20}$$

where S_r is the Received Power in watts,

S_t is the Transmitted Power in watts,

G_t is the Transmit Antenna Gain (isotropic),

G_r is the Receive Antenna Gain (isotropic),

λ is the wavelength of RF carrier,

d is distance between transmitter and receiver.

If expressed in dB units, the equation above can be rewritten as

$$S_r(dBw) = S_t(dBw) + G_t(dBi) + G_r(dBi) + 20\log_{10}\left(\frac{\lambda}{4\pi} \right) - 20\log_{10}(d) . \tag{21}$$

Assume

$d = 1m$,

$G_t(dBi) = G_r(dBi) = 6\, dBi$,

Carrier frequency is 900 MHz,

The received power consumption will be:

$$S_r(dBm) = S_t(dBm) - 19.53dB . \tag{22}$$

As shown in (22), the channel loss will be 19.53 dB based on the conditions listed above. The propagation noise can be modeled as additive white Gaussian noise (AWGN) [44] [49] and will be modeled using ADS transmission channel models depicted in Chapter 4.

3.4.2. RF frequency, Bandwidth, Power

Due to the different locations and functions of body unit and earpiece, the design principles on transmitter and receiver are different as well. The receiver in earpiece requires a system structure with low power consumption and a circuit area as small as possible. While in the body unit, power cost and circuit area are no longer a problem. The design on RF transmitter in body unit requires high Adjacent Channel Power Rejection (ACPR) and proper transmission power under regulation.

<u>Carrier Frequency and IF frequency</u>

The ISM (industrial, scientific and medical) bands limit choices of RF frequency are 30, 200, 400 and 900 MHz. Although low frequency is beneficial for power dissipation, the circuit area can be much larger [13], in the case that the bulky passive components like filters can only be built off-chip. However, RF blocks built in 900 MHz now can reach satisfactory low power dissipation [15] [43]. In terms of antenna efficiency and compact size, choosing 900 MHz as RF frequency can reach a good compromise between power and size. Another concern of choosing 900 MHz as our carrier frequency is for the compatibility to GSM mobile phones, the frequency band of which lies close to our selection. So it will be easier for future integration in mobile electronic devices.

<u>Data Transmission rate and Bandwidth:</u>

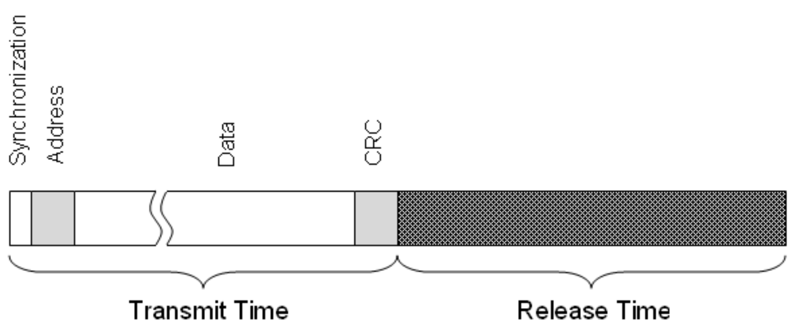

Fig. 3.6 Segmentation of a time slot.

Sampling an audio signal at 16 kHz at a resolution of 12bit/sample with data compression (12 bit to 8 bit) can yield a data rate of 128kbit/s. The time slot is divided into two parts, transmit time and release time as shown in Fig. 3.6. Thus, the data transfer rate is doubled to 256kbit/s. Since the implementing communication protocol needs redundant data bits for synchronization, addressing and CRC check as listed in Fig. 3.6 and Table 3.1, the final gross bit rate from the body unit to earpiece is estimated to reach 288kbit/s. A quadrature phase shift keying (QPSK) modulation scheme is currently employed in our system which reduces symbol rate to approximately 144 baud/sec. A RF bandwidth of 200 kHz is planned with plenty of design margin.

Table 3.1 Summary of data for time slot.

Sampled bit rate	128kbit/s
Transmit time	2msec
Release time	2msec
Synchronization	16 bit
Address	32 bit
Data	512 bit
CRC	16 bit
Bit per packet	576
Gross bit rate	288kbit/s
Symbol rate (m=4)	144kbaud

However, the time slot and data packet are complicated and difficulties are met to be realized in system level simulation without compromising simulation setup complexity and simulation time. To solve this problem, in the system simulation using ADS, the release time is minimized to zero, and the sampling frequency is doubled to 32 kHz. This ideal situation (ignoring the release time) can greatly simplify system models, and the gross bit rate remains the same. As a result, the RF behaviors of system can still be accurately simulated

3.4.3. Transmit power and APCR

As there is no existing regulation on similar wireless hearing aid system to determine transmitter power, Federal Communication Commission (FCC)'s regulation on Low Power Radio Service (LPRS) [50] is used as a reference standard. The LPRS is a private, one-way short-distance communication service providing auditory assistance to persons with disabilities, persons who require language translation, and other purposes. While this LPRS is especially made for one-way communication, it is feasible to choose LPRS as a reference regulation in the system design. Moreover, there is no need for a FCC license for a LPRS transmitter in a wireless hearing aid system according to FCC regulation.

The authorized maximum transmission power in this regulation is 100mW (20dBm) and the minimum Adjacent Channel Power Ratio (ACPR) is set to be 30dBc. However, taking into consideration of the short distance within 1 meter, portability, and power cost, especially the RF safety to human, the minimal transmission power is chosen to be 0.1mW instead. This value is comparable to other wireless hearing aid [13] [14]. The expected ACPR remains 30dBc. The proposed transmitter parameters are listed in Table 3.2.

3.4.4. RF receiver specification analysis

One of the design criteria mentioned above is to minimize receiver power consumption. However, ADS Ptolemy can not simulate the block power consumption at system level. The only way to evaluate the digital communication receiver performance in ADS Ptolemy is by measuring the Bit Error Rate (BER), which is more meaningful to digital communication system compared to SNR measurement.

The feasible solution is to choose a receiver architecture which has more advantage on lowering power cost and try to realize desired power in circuit level design. In our system simulation, the parameters are to be frozen for optimized BER performance within such structure.

Both frequency shift keying (FSK) and phase shift keying (PSK) seem to be the choice. During current stage, the QPSK modulation scheme and super-heterodyne receiver structure are chosen due do their well establishment, maturity and available models in ADS2002C.

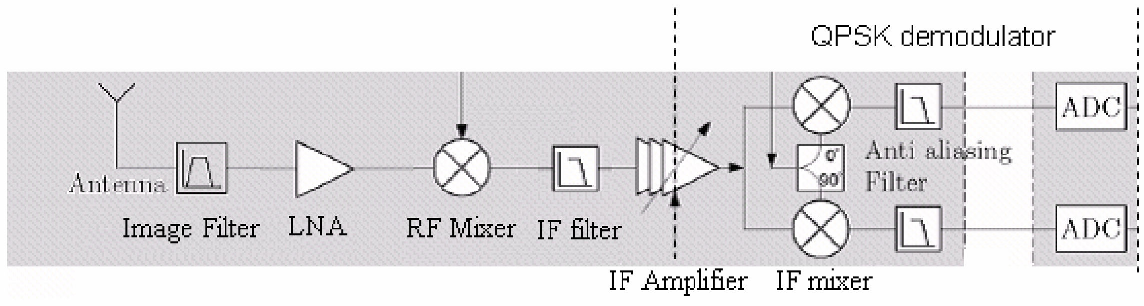

Fig. 3.7 QPSK receiver structure.

The RF receiver structure with QPSK demodulator is shown in Fig. 3.7. The signal strength at the input of receiver antenna can be calculated using

$$V_s = \sqrt{30\Omega G_T P_T} * \frac{l_{eff}}{d} \quad .$$

If take:

Transmitter output Power P_T = -10dBm (0.1mW),

Effective Antenna Length l_{eff} =2 cm,

Distance between transmitter and receiver d<=1 m,

Antenna Gain G_T <=3,

The minimal input signal at the receiver input is calculated to be 35.5dBμV.

Thus, the nominal signal strength V_{ADC} at the ADC input is set to be 103dBμV.

The maximal receiver gain is set to

$$A_{V\max} = V_{ADC} - V_s = 67.5dB \ .$$

For QPSK modulation scheme, in order to achieve a low Bit Error Probability (P_e <0.1%), the SNR before ADC input should meet $\frac{S_b}{N_o} >= 5dB$ [49].

$$V_{n\,out} = V_{ADC} - SNR \le 103dB\mu V - 5dB\mu V = 98dB\mu V$$

$$V_{n\,in} = V_{n\,out} - A_{V\max} = 98dB\mu V - 67.5dB\mu V = 30.5dB\mu V$$

The maximal equivalent input noise density will be

$$\sqrt{\phi_{V_{in}}} = \frac{V_{n_{in}}}{\sqrt{B}} \le 74.9 \frac{nV}{\sqrt{Hz}}$$

System simulation and parameter optimization are performed in Chapter 4 with the input noise density and maximal receiver gain obtained in this section.

Table 3.2 Expected parameters of RF transceivers.

Parameters	Value
bit rate	288kbit/s
Baud rate (m=4)	144kbaud/sec
Minimal receiver input	35.5 dBμV
Receiver BER	<=0.1% (input SNR=13dB)
Maximal receiver gain	67.5 dB
Maximal input noise density	74.9 $nV/\sqrt{Hz}$

Chapter 4. System Model Building and Simulation Results

Chapter 3 has proposed the new architectural concept of our wireless hearing aid system, also explained how the various parameters of sub-blocks affecting system noise factor and analyzed the basic settings of the RF transceivers. This chapter will discuss about the system level simulation. Its primary focus has been on

1) Functional check for the proposed system behavior,
2) Developing and embedding the behavior models for critical blocks in order to define the whole system in ADS environment,
3) The verification measure for the system SNR analysis discussed in section 3.3.

It is mandatory to highlight that system level simulation has been proven well in order to free the block level simulation. It has been a secondary concern in this simulation.

There is hardly available any CAD environment suitable for the simulation of the system with complexity as high as that of Fig. 3. However, an attempt is made using the Agilent Advanced Design System™ 2002C [31], as (i) It allows to build the behavioral model (ii) It has feature for extracting external data file using Data Access Component. (iii) It has closure platform for IC design which also allows transistor level simulation in frequency and time domain as well and (iv) Its Ptolemy signal processing simulation feature enables fast RF simulation, integration with signal processing, and co-simulation with integrated circuit simulators. As the most conventional hearing aid chips can be designed using the industry standard CAD environment like cadence, or ADS but these are limited to handle the simulation of the wireless hearing aids mainly because (i) lack of equivalent electrical models for transducers (ii) poor capability in measuring the random signal, and

(iii) equivalent behavior models needed to define the beamformer method.

4.1. Behavioral Model Building

As an attempt to enable the wireless hearing aid system simulation, the inherent features of the ADS are exploited to generate an equivalent ADS compatible system. Fig. 4.1 depicts the setup developed which is compatible in ADS environment. The details of the development of this set up are given below:

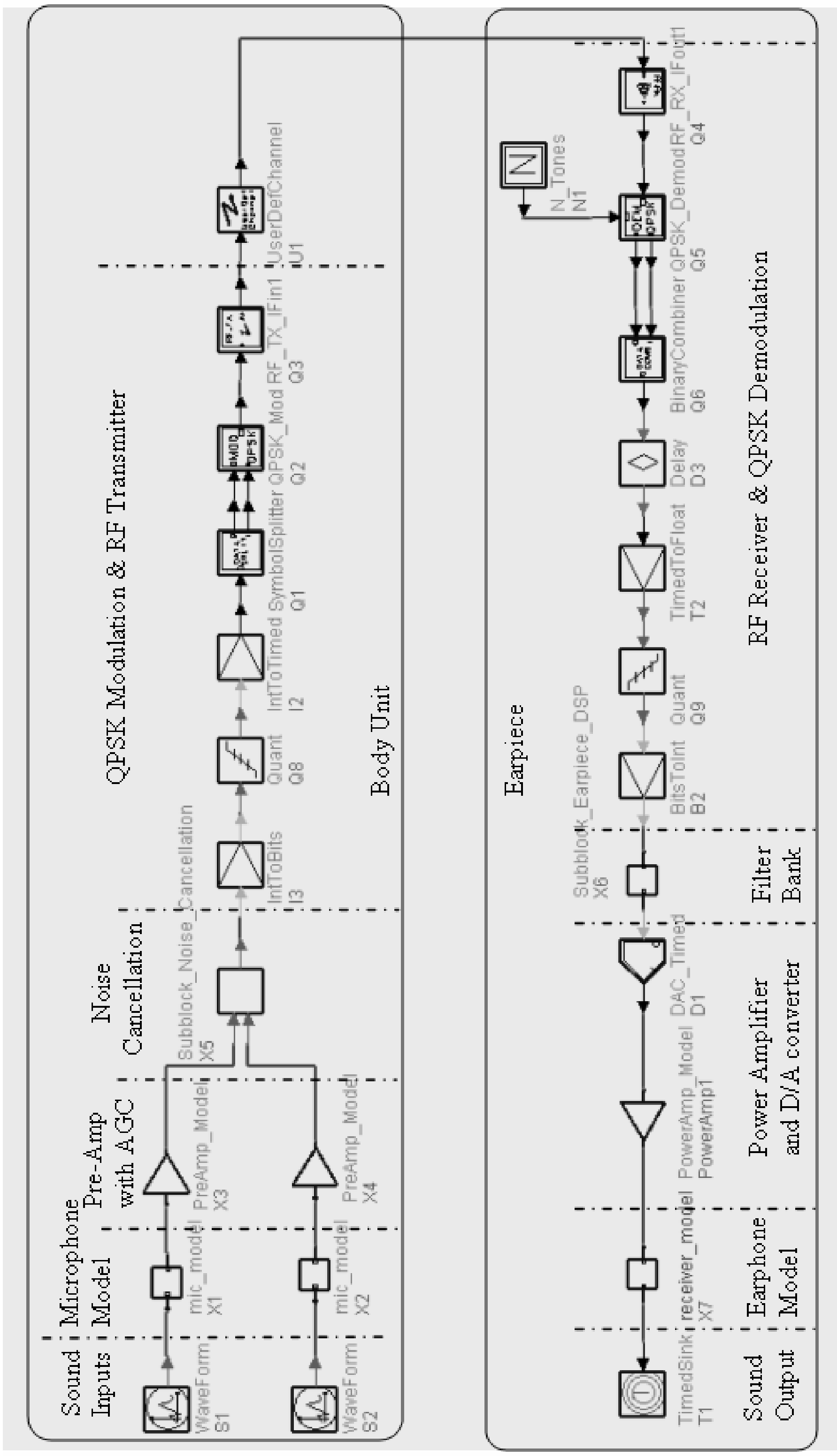

Fig. 4.1 Proposed system simulation setup in ADS environment.

4.1.1. Electro-Acoustic Transducers

In Fig. 4.1, signal sources S1 and S2 are used as input in sound pressure level (SPL) to the two microphones X1 and X2 respectively.

Behavioral modeling of electro-acoustic transducers, e.g. microphones (X1, X2) and earphone (X7) in Fig. 4.1, is realized using pre-measured data from frequency response of earphone model BK1600 and microphone model EK3024 [30]. These data are saved in external data files. A *voltage-controlled voltage source (VCVS)* is used within the microphone model. The *VCVS* defines its gain according to the data file by means of a look up table to define the microphone's electro-acoustic behavior, shown in Fig. 4.2. The interface between ADS simulation and data file is implemented using the *DataAccessComponent* (*DAC*) model. A transient simulation controller is placed in this block for circuit behavioral simulation. As the simulation in system level is Data Flow (DF) simulator, a co-simulation will perform during simulation.

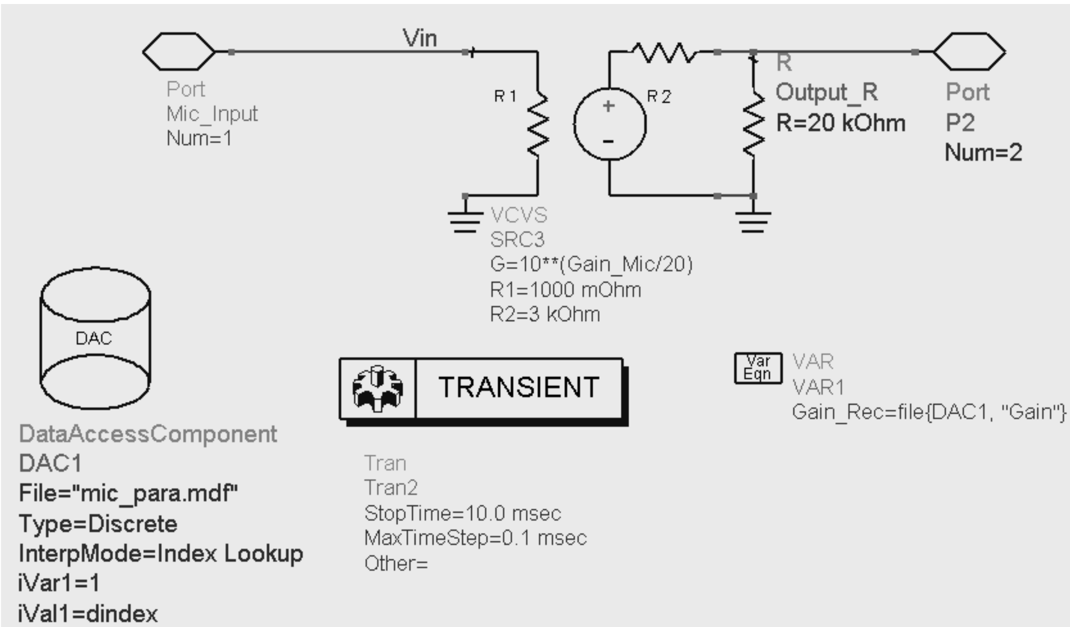

Fig. 4.2 Microphone model setup.

The earphone model X7 is built using *DAC* in a similar manner as in X1 and X2 except that an additional RLC circuit is added as shown in Fig. 4.3 to represent the AC and DC loading on the amplifier output stage [30]. The pre-measured data of BK1600 earphone model from external file is used to set model parameters.

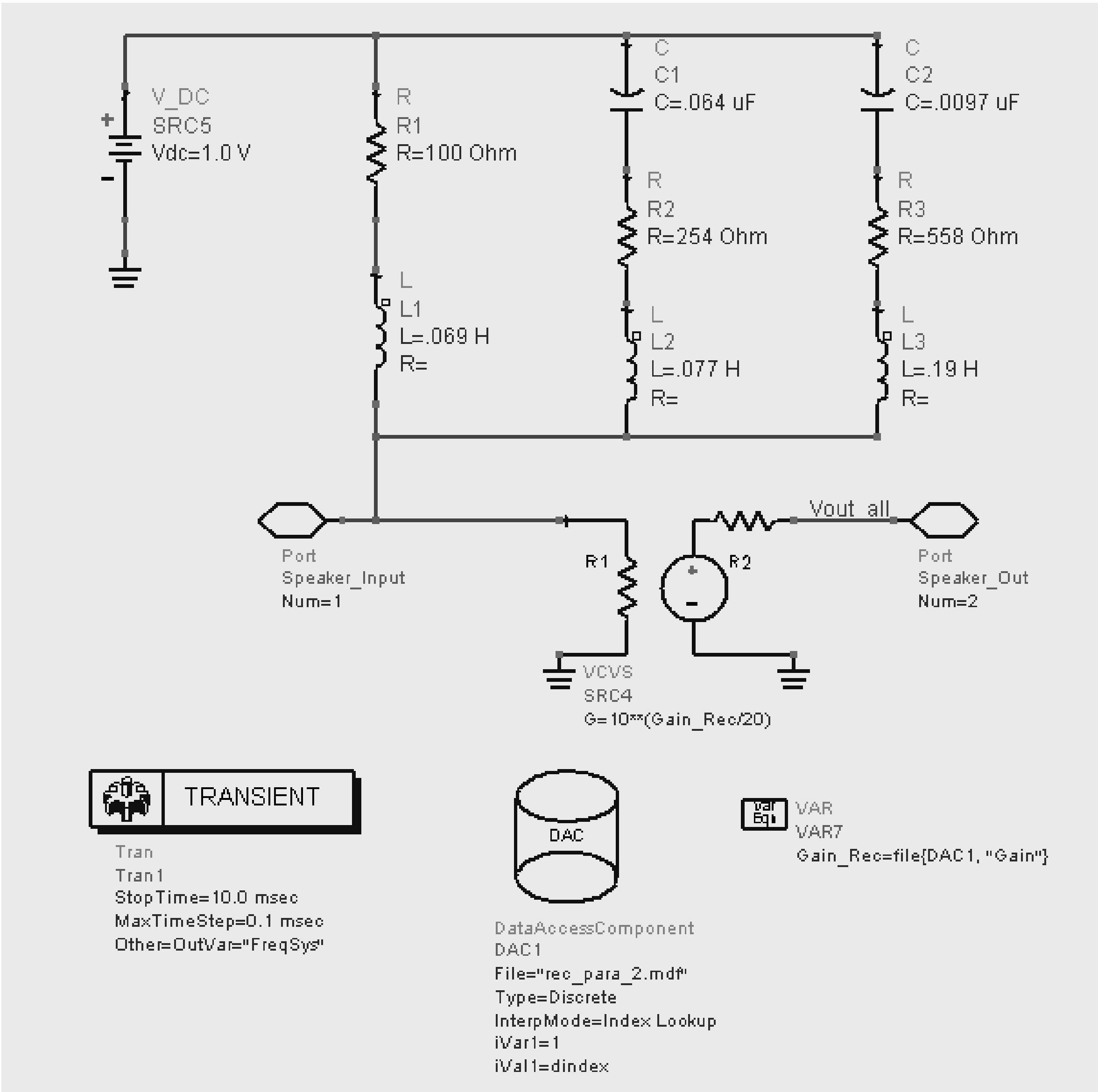

Fig. 4.3 Earphone model setup

4.1.2. Pre-amplifier

Preamplifier blocks in the body unit (X3 and X4 in Fig. 4.1) are built using generic amplifier model from ADS library. The model also enables circuit noise definition of X3 and X4 by setting the value of *Vnoise* in *AMP3* shown in Fig. 4.4. The noise value is set according to [17]. Also, a model for non-linear auto gain control (AGC) is built in this stage to provide effective loudness compression. The compression knee point (CK) is programmable and can be set for patients with different dynamic ranges. Simulation on AGC is carried out according to American national standard institute (ANSI) S3.22 [34] for both static and dynamic properties.

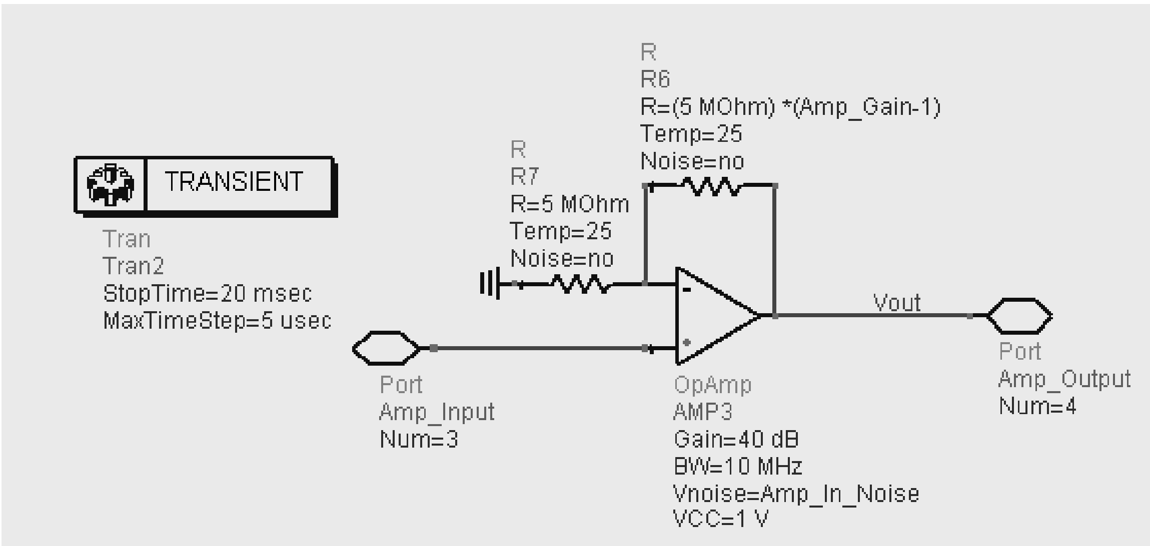

Fig. 4.4 Pre-amplifier model setup.

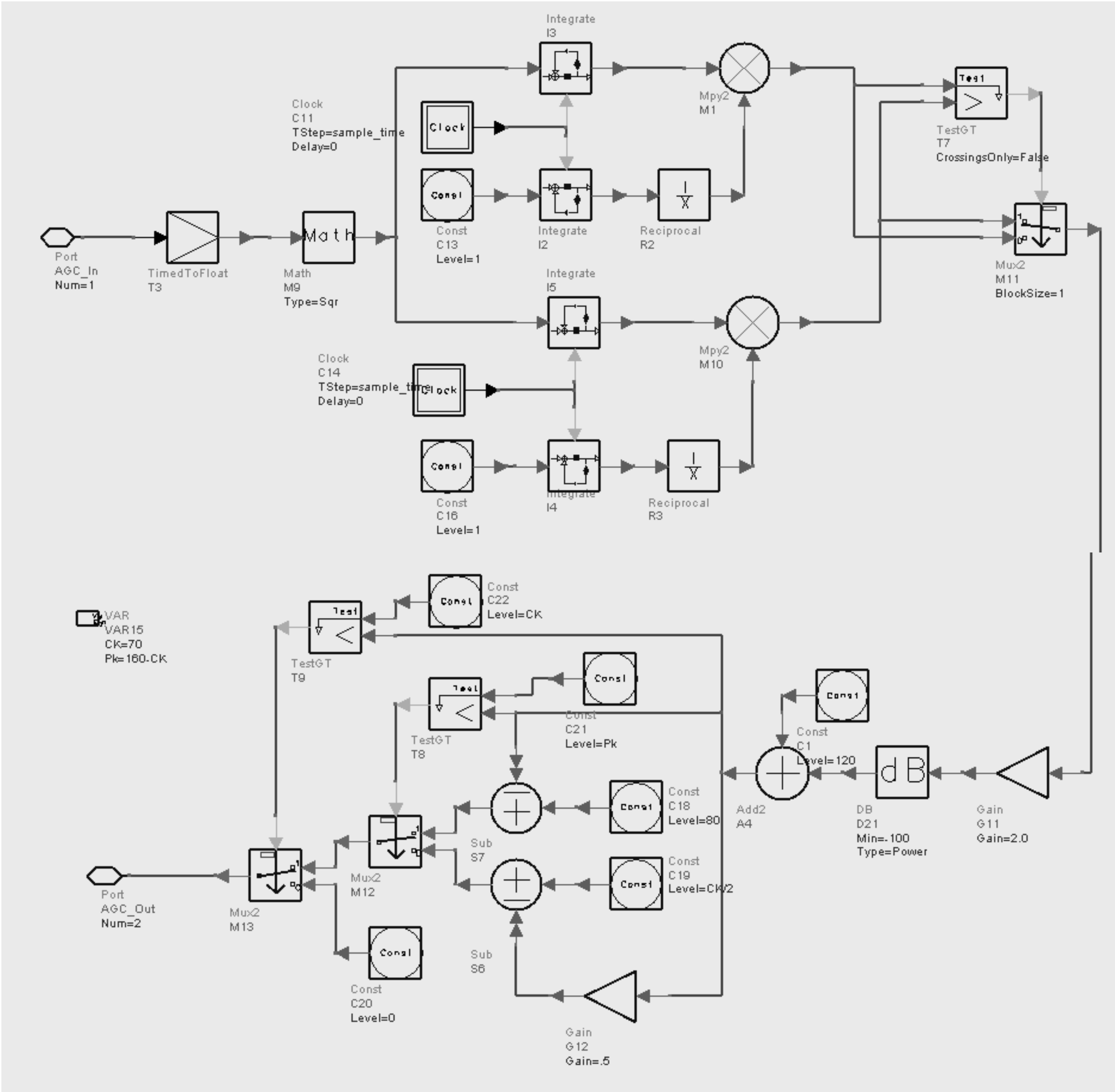

Fig. 4.5 AGC simulation setup.

4.1.3. Noise Cancellation Unit

The noise cancellation unit is built in X5. It contains two A/D converters and a circuit implementing the beamforming algorithm for speech enhancement. The equivalent model for A/D converters and two-element beamforming algorithm [36] is developed by configuring the generic blocks from ADS library. Fig. 4.6 shows the corresponding ADS setup developed for a single path of the beamformer. Based on this a complete setup for whole noise cancellation unit was developed and used in the whole system simulation.

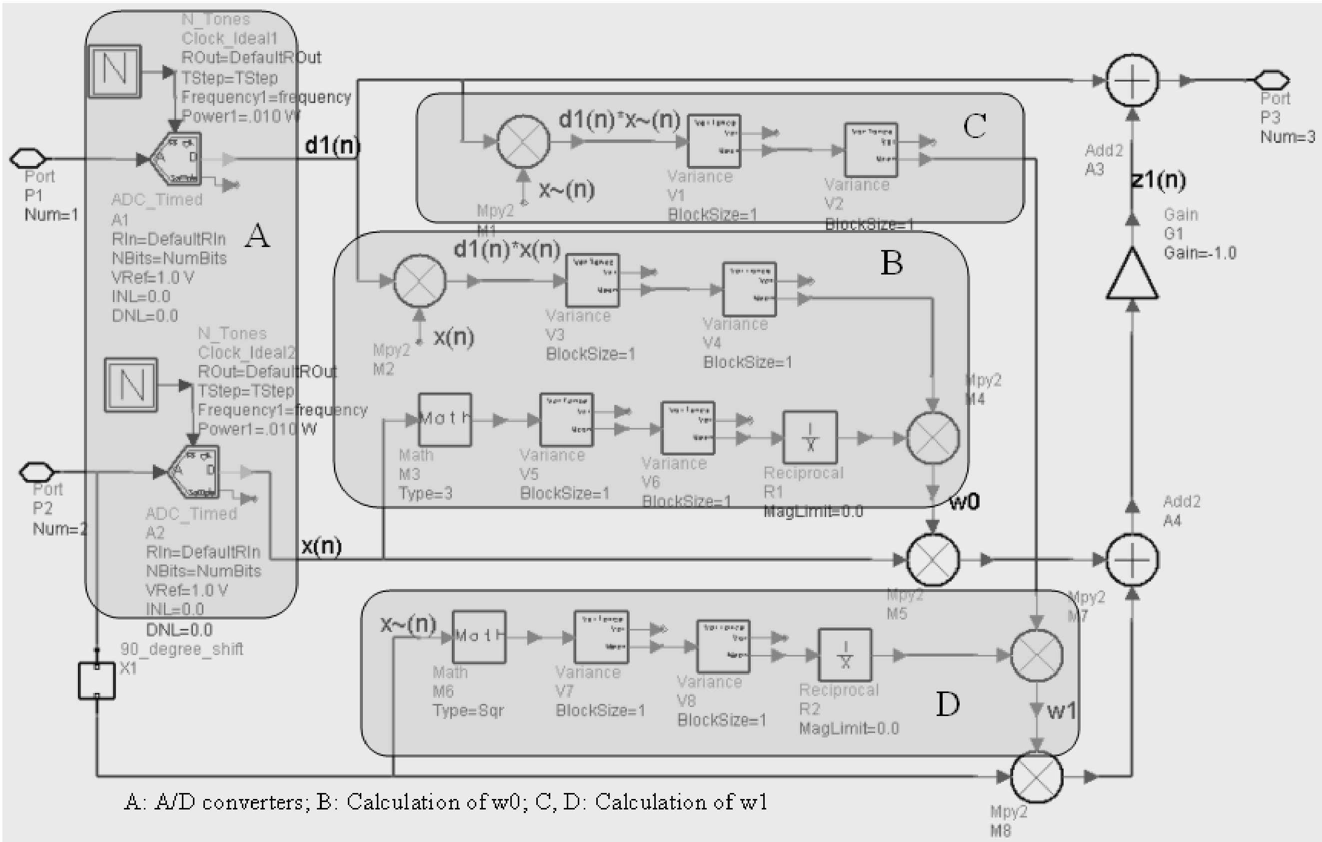

Fig. 4.6 Beam-former model set up in ADS environment.

4.1.4. RF Transmitter

An ADS compatible functional setup is developed for the two stage RF transmitter as shown in Fig. 4.7. The first 4 blocks have no physical meaning and functions to separate in-phase (I) and quadrature (Q) signals from incoming data streams and make these signals compatible to *QPSK_Mod*. The *QPSK_Mod* block is a generic function model from ADS library to modulate input signals up to QPSK RF signals at IF frequency. After modulation, it is up-converted into carrier frequency by *Rx_model* and *EnvOutShort*.

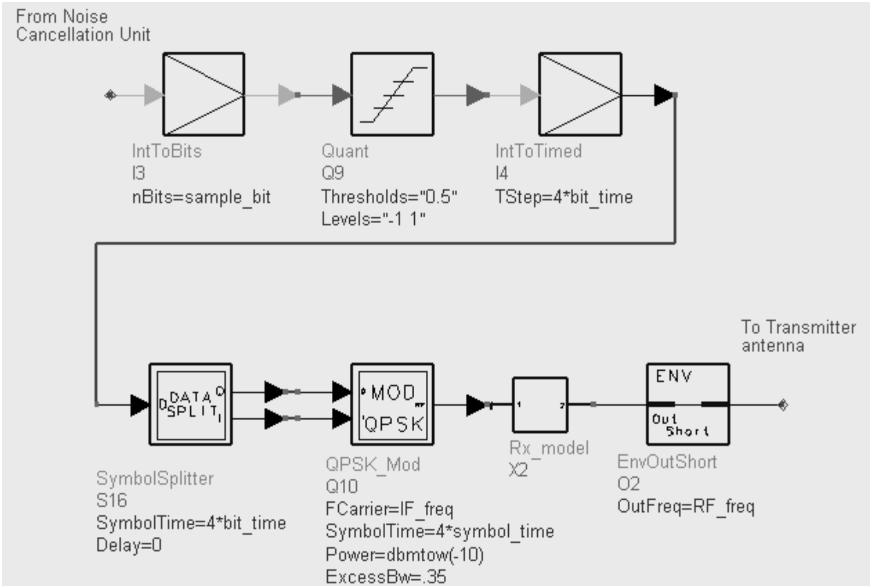

Fig. 4.7 RF Transmitter model (system level).

The *Rx_model* block comprises several RF behavioral sub-blocks as shown in Fig. 4.8. It is driven by a RF envelope simulator. Comparing to the generic transmitter models provided by ADS library, our behavioral model setup is more flexible to hold simulation as well as parameter optimization. To ensure a successful co-simulation between DF simulator (system level) and ENVELOPE simulator (this block), an *EnvOutShort* component is added after this block in Fig. 4.7 [31].

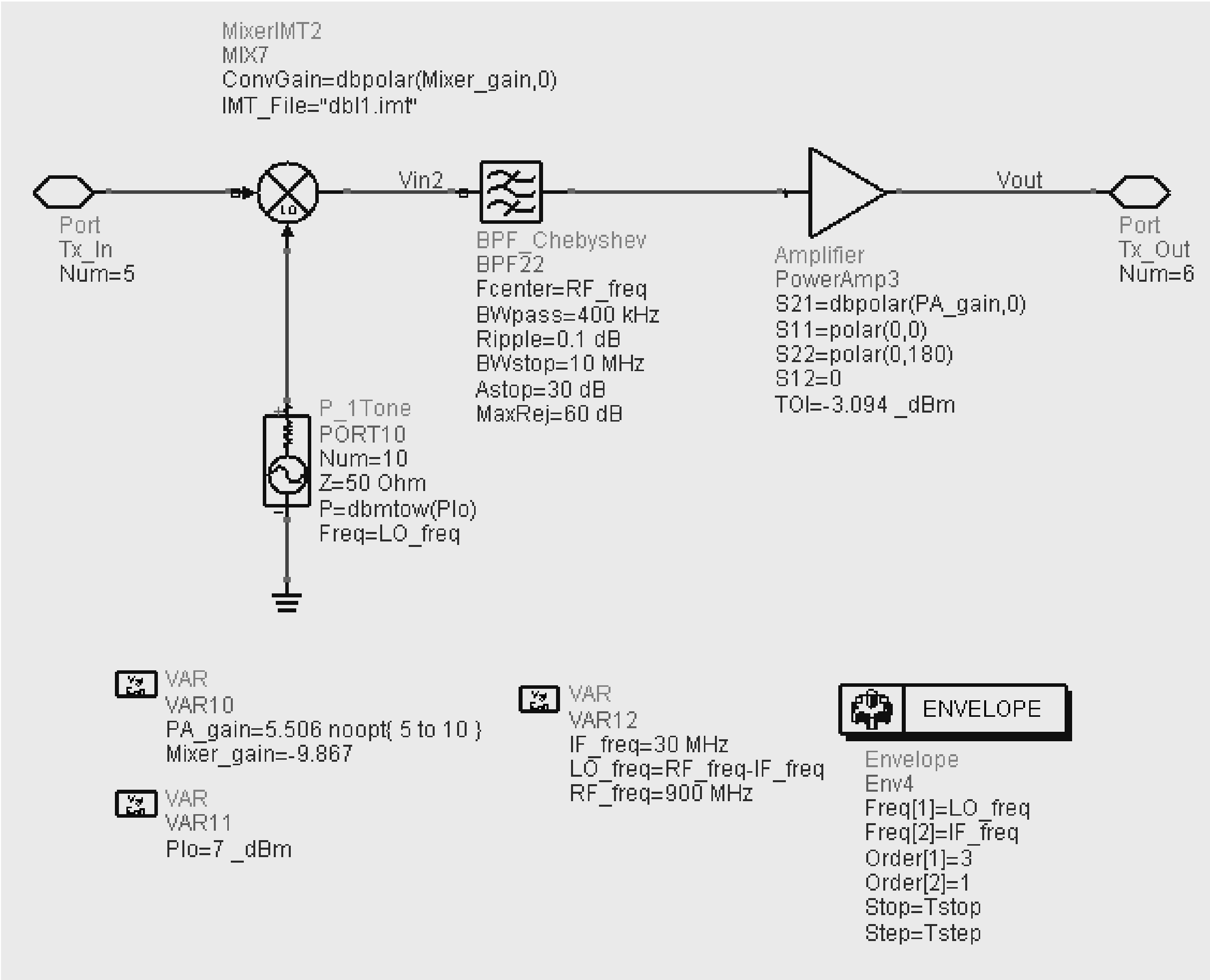

Fig. 4.8 Up-converter subsystem model.

In order for block parameter optimization, an additional schematic of transmitter is built on the base of Fig. 4.8. Reference values of key blocks (filter, power amplifier, up-conversion mixer) are set initially based on search results from references.

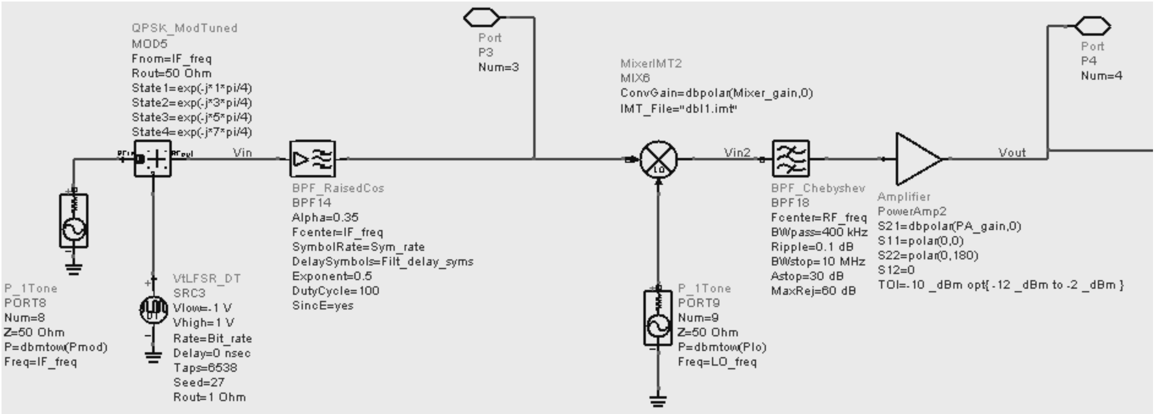

Fig. 4.9 Block schematic of RF transmitter for optimization.

The mixer is a behavioral mixer model, based on the data file *dbl1.imt*, which simulate the intermodulation of a double balanced mixer as shown in the figure.

The *QPSKTuned_mod* is a QPSK modulator model in A/RF schematic. This QPSK modulator is included to resemble the QPSK_modulator in Ptolemy and do performance measurement and optimization in this schematic. When integrated into the whole system, the RF transmitter will be like Fig. 4.8

The expected design goal is to fulfill the ACPR and output power, A few measurement blocks need to be added as follow:

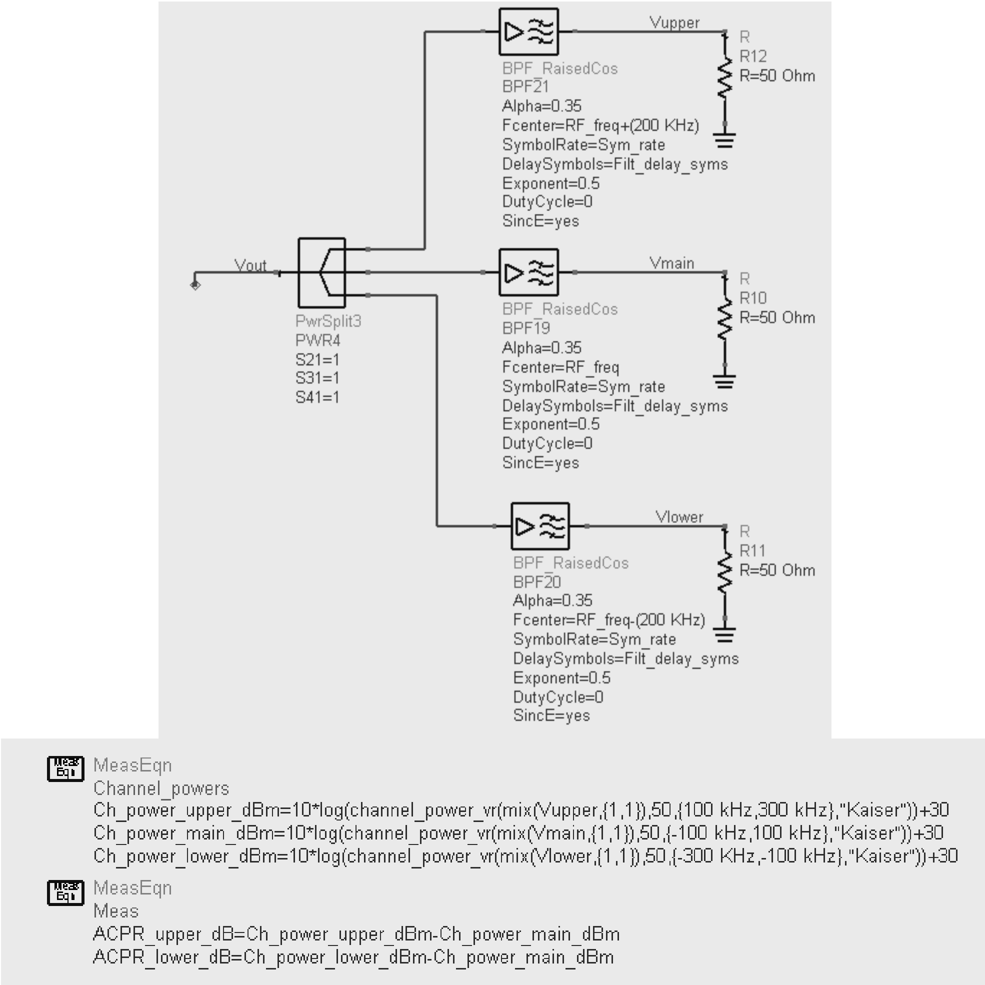

Fig. 4.10 Additional simulation setup of RF transmitter.

In Fig. 4.10, the *PwrSplit3* is added to the RF output and duplicate the signal at its input to 3 outputs. The *BPF_RaisedCos* is the band pass raised cosine filter help to filter out signals in different frequency band, each 200 KHz wide (the proposed bandwidth). The measurement equations listed above in the schematic will be used in simulation data display.

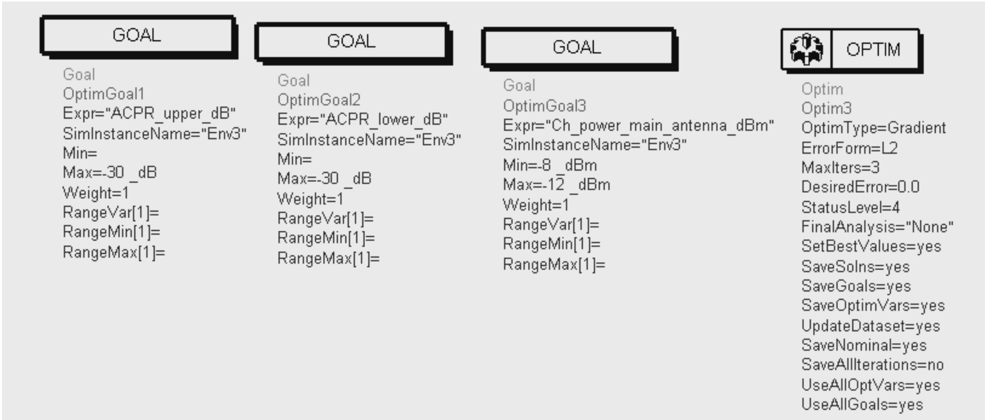

Fig. 4.11 Optimization goal and controller of RF transmitter.

Fig. 4.11 shows the optimization simulation controller setup. The expected ACPR is 30dBc, and the expected output power is between -8dBm and -12dBm.

4.1.5. RF Receiver

The RF signal propagation channel is simulated using two antenna model (A6, A7) and a GSM channel model (P2) in Fig. 4.12. As analyzed in section 3.4.1, no multi-path fading effect is considered in our setup as set in P2.

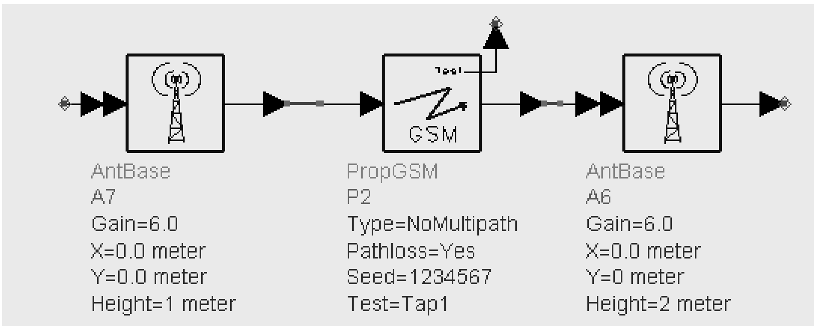

Fig. 4.12 Propagation channel simulation setup

For the modeling of the earpiece comprising a RF receiver and an earphone, a super-heterodyne receiver structure and QPSK digital modulation scheme are considered [44].

RF signal is picked up and down converted to 30 MHz signal by the *RF_Rx_IFout1_New* block and then further demodulated by passing through the demodulation sub-block (*QPSK_Demod*). The following five sub-blocks serve to convert the demodulated I/Q signals to digital signals required for further processing. As the actual simulation delay along each RF block may affect correct extraction from I/Q signals, it is necessary to place the *delay* blocks before the *BinaryCombiner*

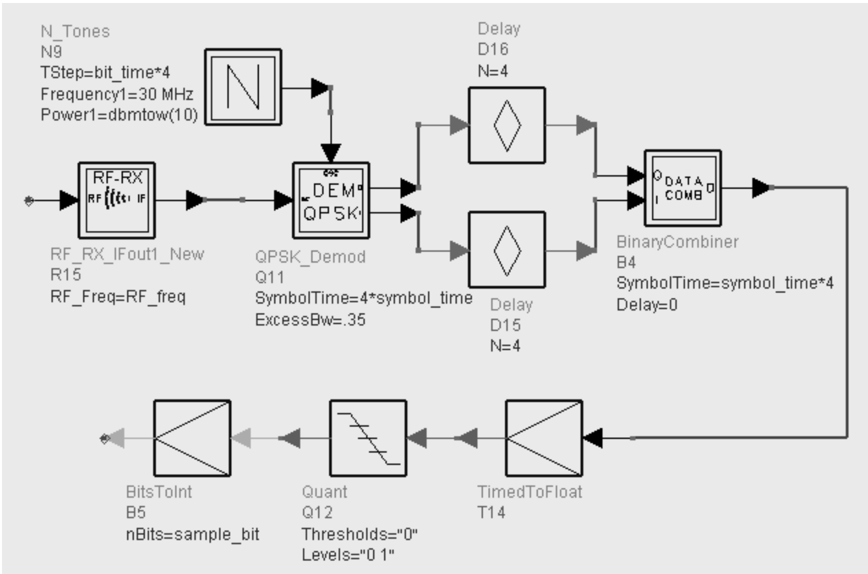

Fig. 4.13 RF Receiver model (system level).

In order for block parameter optimization, additional components for BER measurement need to be added. And the schematic of receiver has to be redrawn in Fig. 4.14. The two *berIS* blocks measure bit error rate at both I and Q channel. The actual BER value is to be determined using the following equation:

$$BER = IChannelBER/2 + QChannelBER/2 - QChannelBER * IChannelBER/4 .$$

The optimization goals and controller are added as shown in Fig. 4.15.

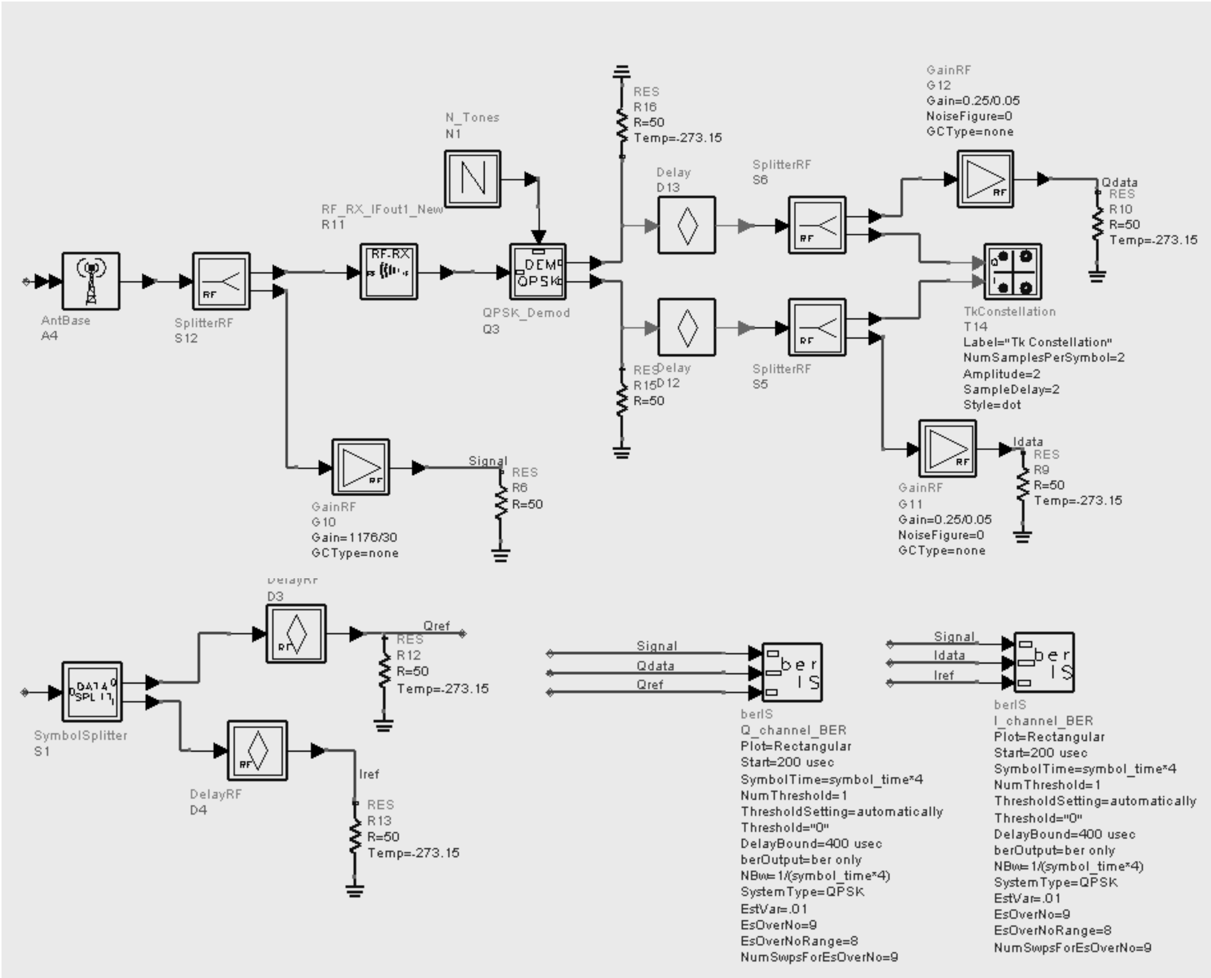

Fig. 4.14 Simulation setup for BER measurement of RF receiver.

Fig. 4.15 Optimization goal and controller of RF receiver.

4.1.6. Filter Bank

X6 in Fig. 4.1 is the inbuilt filter bank in the earpiece. It works to modify the frequency spectrum of output signal so as to compensate the patient's hearing loss diagram. For simplicity and simulation time consideration, it is built up using 5 sub-band finite impulse response (FIR) filters [12]. The center frequency of these band pass filters are set to be 250 Hz, 500 Hz, 1 kHz, 2 kHz and 4 kHz respectively. As shown in Fig. 4.16, an ideal *gain* component is placed in each channel. By changing the gain of each sub-band, the filter bank is programmable in simulation against different hear loss types.

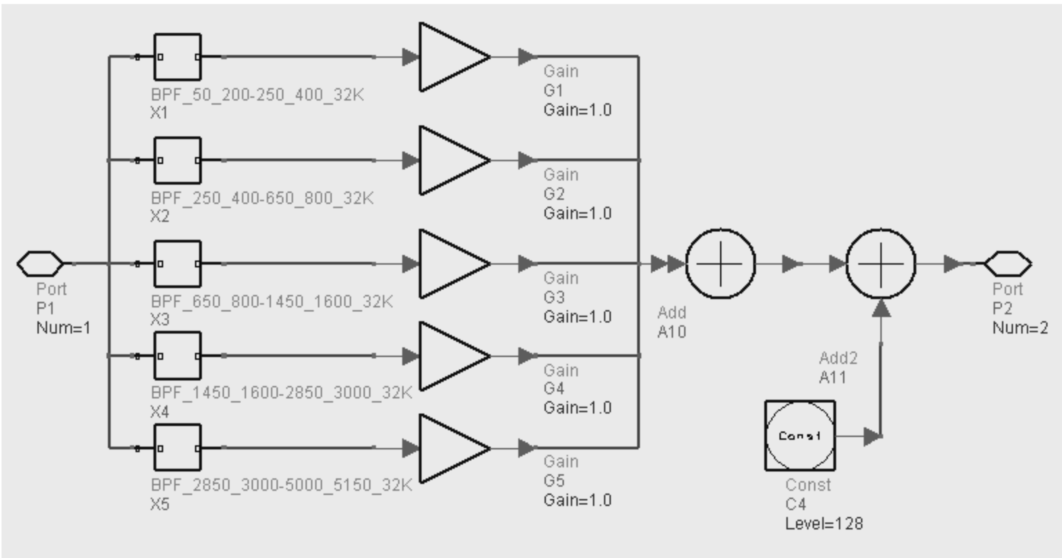

Fig. 4.16 Filter bank simulation setup.

4.1.7. Output Stage

Signal is finally converted to sound by passing through D/A converter, power amplifier and earphone model (X7) and is displayed using data display block (*TimedSink*) in Fig. 4.17.

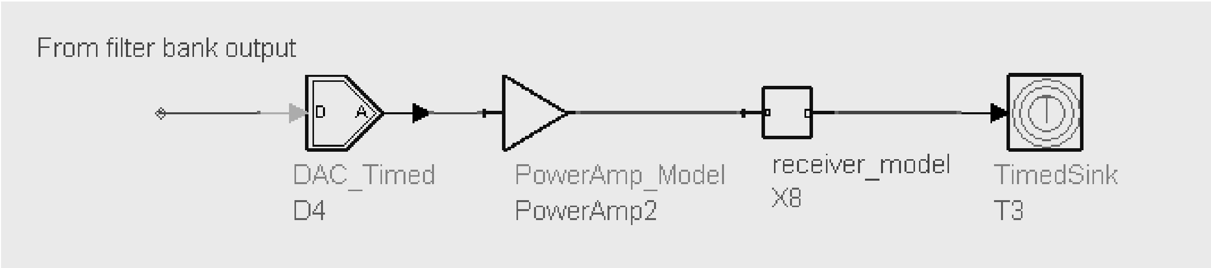

Fig. 4.17 Output stage model setup.

4.2. Parameter Setting

Behavioral models for sub-blocks and transducers system simulation on proposed wireless hearing aid is performed and a few simulation results will be given in next section. Some of the key system settings are listed as follow. The full-on gain of preamplifier block is 40dB. Preamplifier circuit noise is set according to [17]. The center frequency of noise cancellation block is 3 kHz as discussed in section 3.2. The A/D, D/A converters are set to have a peak value of 1V and sampling bit of 12. The sampling frequency is 32 kHz. The carrier frequency of RF transmitter is chosen to be 900 MHz with QPSK as modulation and demodulation scheme. Based on these inputs to the system, the complete model of the system is simulated using ADS, some of the results are described below.

4.3. Simulation Results for Baseband Blocks

4.3.1. Noise Cancellation

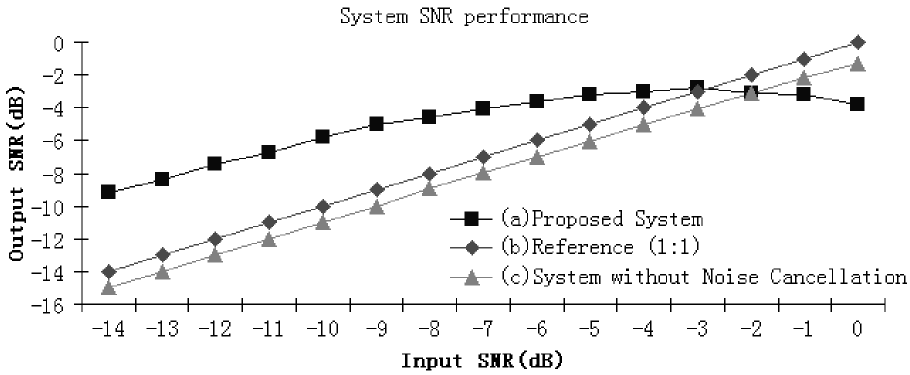

Fig. 4.18 Simulation data of system output SNR.

Noise cancellation performance is shown in Fig. 4.18 and Fig. 4.19. The output SNR and input SNR relation of proposed system in Fig. 4.18 (Curve a) meets the theoretical analysis above well.

A reference curve (Curve b in Fig. 4.18) is used to indicate the behavior of the system assuming noiseless circuit components and having no provision for noise cancellation. Thus, the gradient of Curve b is unity since in such case $SNR_{in} = SNR_{out}$ stays true. Output SNR performance of our system without noise cancellation unit but having all types of system noise has also been simulated and represented by Curve c. It is found around 1dB below the reference curve due to internal circuit noise, quantification error and communication errors.

Once noise cancellation unit is added, the output SNR is no long linearly related to input SNR. When noise canceling is working, output SNR (Curve a) is increased and stay above to Curve c. Since for the same input SNR, effective noise suppression only happens when Curve a is above Curve c, Fig. 4.18 indicates improvement of signal SNR is achievable using proposed system, which is believed to be beneficial to the hearing impaired.

However, the improvement decreases with the increase of input SNR as depicted in Curve a. The input SNR threshold for effective speech enhancement can be read out at the point where Curve a and Curve c join, which is around -2dB. As analyzed in section 3.2, the system output signal-to-noise spectrum density ratio is reverse proportional to the input signal-to-noise spectrum density ratio. Thus a threshold will occur but still proves the applicability of this system and beamforming algorithm as further analyzed in [36]. Other noise cancellation algorithms can be implemented in the proposed system for better SNR improvement which is out of our scope in this thesis.

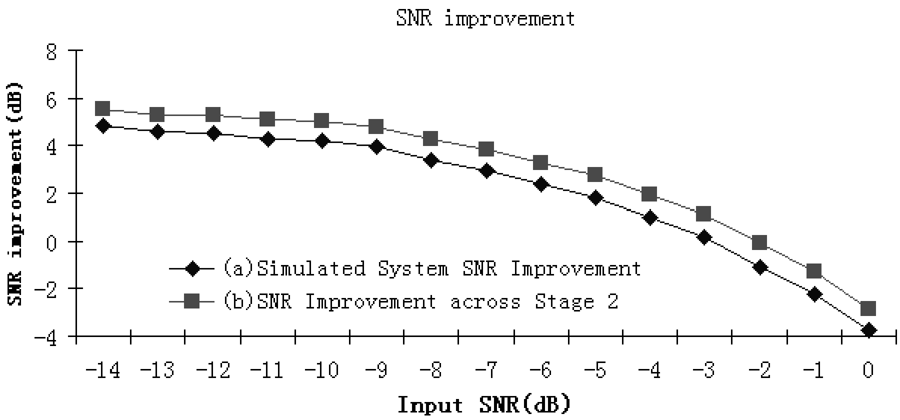

Fig. 4.19 SNR improvement across stage 2 and system.

Comparison between a reference system with noise cancellation but no circuit noise and the proposed system is also simulated and the results are shown in Fig. 4.19. The SNR improvement of the proposed system (Curve a) is 1 dB lower than the SNR improvement (Curve b) of the reference system which is comprised of stage 2 and other noiseless circuit blocks. It proves that this 1 dB degradation is due to combinational effect of all types of internal noises..

4.3.2. System Frequency Response

The frequency response in term of the input and output sound pressure level is a key parameter evaluating hearing aid system performance. Amplification in the SPL is simulated and the corresponding plots with the input frequency, along with gain of behavioral model for microphone, earphone in are given in Fig. 4.20.

It can be seen that the frequency response of the whole system is dominated by frequency response of earphone and microphone. However, the circuit plays an important role primarily in gain enhancement, control, and SNR improvement.

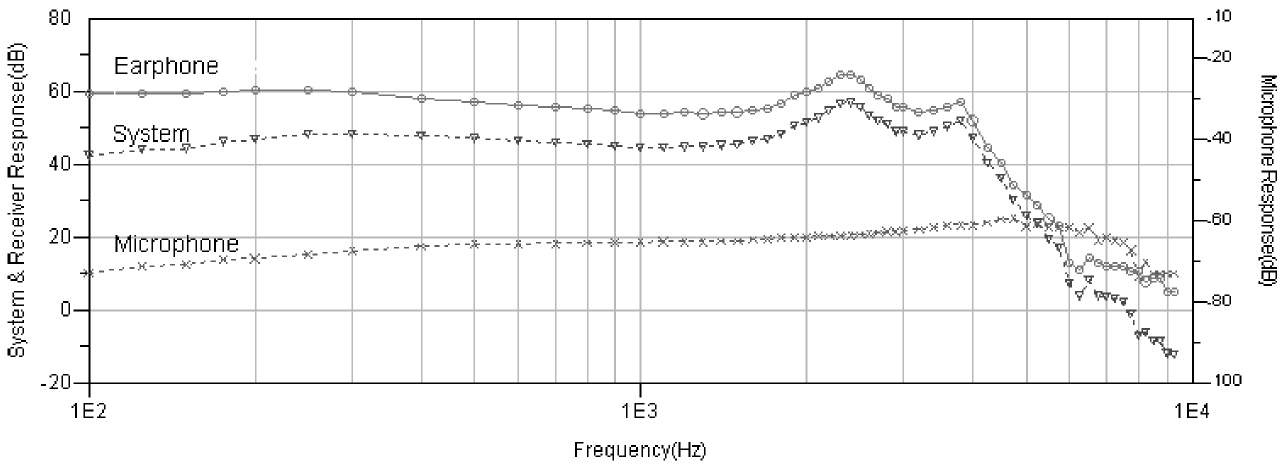

Fig. 4.20 System frequency response.

4.3.3. Auto Gain Control

Fig. 12 shows the static property of amplitude compression capability. A non-linear compression mode is used which is also programmable in addition to the programmable filter bank. Fig. 4.21 shows four curves with different CK points ranging from 50 dB to 80 dB.

It can be clearly seen that system gain is 40dB when input SPL is lower than the knee point. As input SPL increases further, compression occurs and the gain deceases gradually. Output SPL is clipped at 120dB when input SPL becomes excessively loud, since the uncomfortable loudness level

(UCL) of the patient is usually 120dB. This level is controlled by the limiting factor of the limiter in the system.

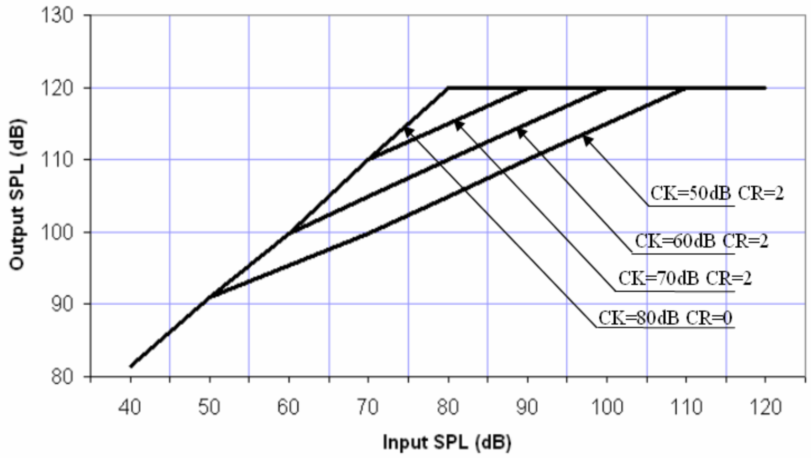

Fig. 4.21 Static property of AGC.

(CR: compression ratio; CK: compression knee point)

Dynamic property of AGC is also obtained through simulation. The simulated attack time is around 9 ms and release time is 150 ms, both of which are within the range of commercial hearing aid.

4.4. RF Transceiver Specification Freezing

The RF sub-blocks in body-unit and earpiece can be specified via system simulation. In this simulation, 900MHz is chosen from ISM (Industrial, Scientific and Medical) bands as carrier frequency and 30MHz as IF frequency. The RF bandwidth is determined by data transfer rate of the one-way RF link. In order to keep the RF bandwidth low, the 12 bit sample data must be compressed to 8 bit before modulation. The gross bit rate can be calculated by multiplying the sampling frequency with the bit number 8. The baud rate of RF link is half of the gross bit rate in the case of QPSK modulation and is calculated to be 144k baud/sec. Thus a RF bandwidth as low as 200 kHz is sufficient to meet the data transfer needs.

As there is no existing regulation on wireless hearing aid system, we use Federal Communication Commission (FCC)'s regulation on low power radio service (LPRS) [50] as a reference standard. The targeted transmission power is set 0.1mW. The minimal adjacent channel power ratio (ACPR) is aimed to be 30dBc. As per FCC, the SNR threshold at the receiver input should not be less than 13dB when Bit Error Rate equals 0.1%. The general RF block parameters considered here are listed in Table 4.1.

Table 4.1 General design reference of RF transceiver.

Parameters	Value
Carrier frequency	900 MHz
RF bandwidth	200 KHz
IF frequency	30 MHz
Modulation type	QPSK
Baud rate (m=4)	144kbaud/sec
Transmit power	<-10dBm
Transmit APCR	>=30dBc
Receiver BER	<0.1% (input SNR=13dB)
Receiver input power	min. -70dBm max. -30dBm

4.4.1. Transmitter Specification Freezing

Optimization simulations are carried out to reach the optimization targets using the simulation setup depicted in section 4.1.4. Final values given by simulation are listed in Table 4.2. The simulation aims mainly on the gain and non-linearity of Mixer, Filter and Power amplifier. It is assumed that the noise figure of transmitter blocks is less important in system level since the received RF signal is accompanied by channel noise mainly. The parameters of other blocks are expected to be optimized through circuit level simulation.

Table 4.2 Parameter values of transmitter blocks after optimization.

Optimized Value	Mixer	Filter	Power Amp

Gain (dB)	5.5	0	6
TOI3(dBm)	-2.154	-	15.986

APCR value can be measured and calculated from Fig. 4.22 to be 32dBc, which is well beyond the minimal requirement. The transmission power is -8.26dBm. The output spectrum of RF signal is also simulated and shown in Fig. 4.23. The bandwidth shown is around 150 kHz and provides design margin comparing to the estimated 200 kHz bandwidth.

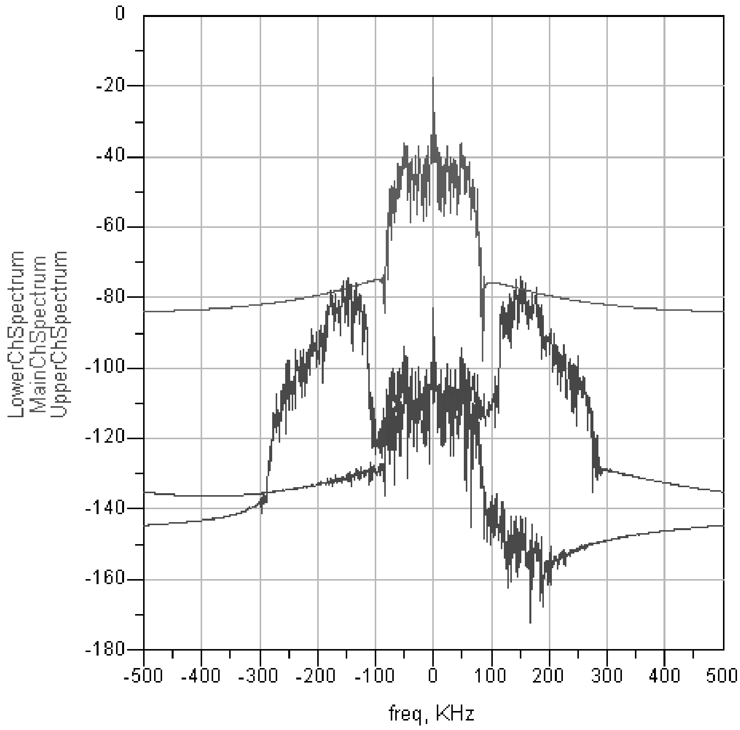

Fig. 4.22 ACPR measurement of optimized transmitter.

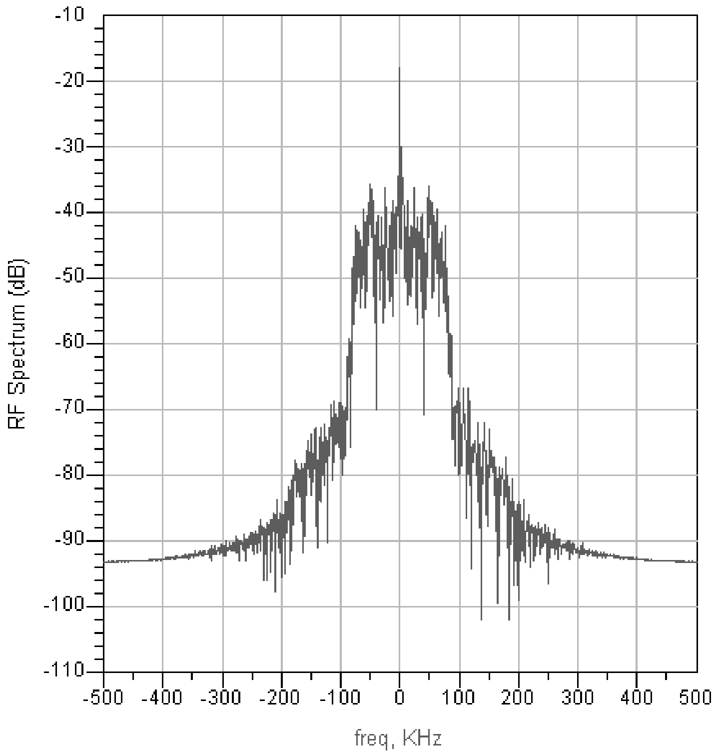

Fig. 4.23 Output frequency spectrum of optimized transmitter.

4.4.2. Receiver Specification Freezing

The receiver specification optimization is performed using the simulation setup depicted in section 4.1.5. Final values given by simulation are listed in Table 4.3. The simulation aims mainly on the gain, noise figure and non-linearity of Low Noise Amplifier (LNA), first mixer, IF filter and the demodulation unit. The parameters of other blocks are expected to be optimized through circuit level simulation.

Table 4.3 Frozen specification of RF receiver by ADS simulation.

	Image Filter	LNA	1st Mixer	IF filter	2nd Mix/AMP
NF(dB)	-	3	5	-	6.4
Gain(dB)	0	12	8	-1	48.5
IP3out(dBm)	-	12.7	15	-	4

The final Bit Error Rate diagram is given below to examine the receiver performance Fig. 4.24. As can be seen from the curve, when P_e=0.1%, the input SNR at the receiver input point is around 11dB, which exceeds the estimated requirement of 13dB mentioned previously. Thus the specification described in Table 4.3 is acceptable and can be used for circuit level design. The final constellation figure is shown in Fig. 4.25.

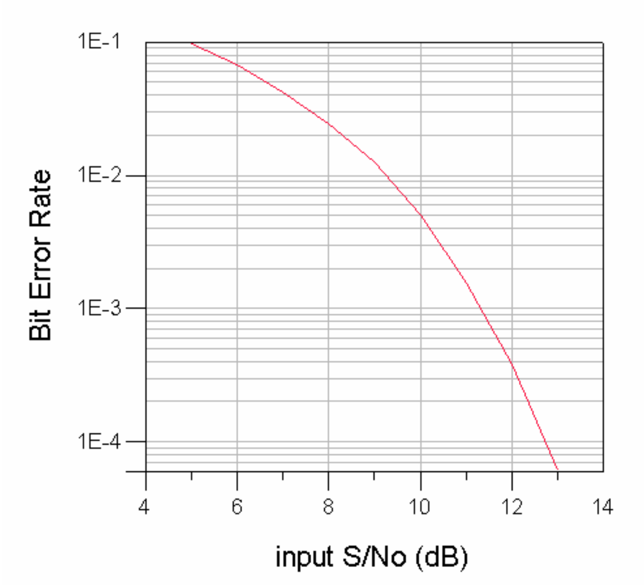

Fig. 4.24 BER performance of receiver after optimization

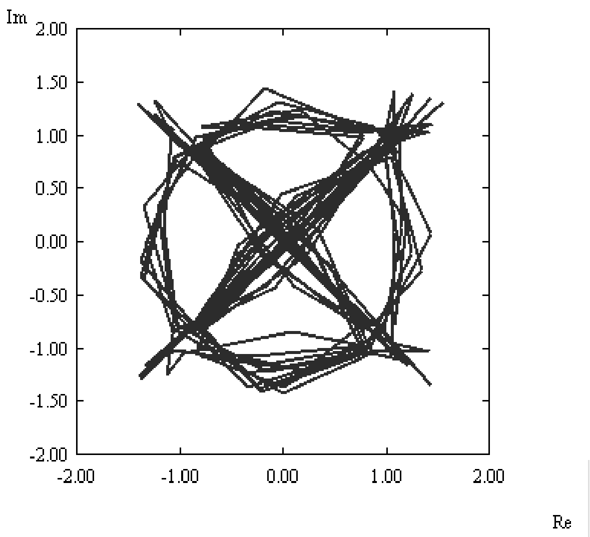

Fig. 4.25 RF signal constellation plot of RF receiver.

4.5. Simulated System Parameter

A few system performance parameters for the proposed wireless hearing aid are tabulated in Table 4.4. The earpiece power consumption is calculated based on A675 battery [51] and a typical current drain for CMOS receiver [43].

Table 4.4 General system parameters.

System Specifications	Value
System gain (Full On)	40 dB
System frequency range	100Hz-6kHz
Input SPL	40-140 dB
Output SPL	80-120 dB
Sampling frequency	32 KHz
Sampling bit	12
Data transfer rate	144k baud
RF carrier frequency	900MHz
AGC type	Non-linear
Attack time	9ms
Release time	150ms
Battery voltage	1.2V
Earpiece power consumption	2.5mW
Battery type	A675 (600mAH)
Estimated battery life	288h
Theoretical distance between Microphones	5.57cm

Chapter 5. Conclusions and Future Work

5.1. Main Conclusions

In this work, wireless hearing aid system architecture has been introduced. Its main features are:

1) Only one way data transfer between body unit and the earpiece,
2) Earpiece does not need a RF transmitter,
3) DSP and CMOS compatible,
4) Reverberation/Interference cancellation,
5) Possible integration with handheld portable devices.

A two-element beamformer based dual-microphone noise cancellation method is also proposed. Its usage in the wireless hearing aid application is demonstrated through system level simulation. The theoretical effect of the noise/reverberation canceling across the system has been analyzed using the concept of noise factor which has not been reported before.

To check and ensure the functional behavior of the wireless hearing aid system, a behavioral modeling approach is introduced. An ADS compatible system simulation set up is developed. It includes the behavioral model building for all the key blocks, (e.g. transducers, noise canceling unit, AGC, RF transceiver).

Final simulation results are presented. The system performance obtained proves that our proposed system is able to suppress background noise with less consideration on power consumption and circuit area.

5.2. Future Work

To bypass the limitation of ADS Ptolemy simulation which helps little in simulating power consumption in system level, analog simulation using circuit level models may be a supplementary method in the future.

Physical implementation of proposed wireless hearing aid system is needed for future work. Actual measurement on prototype is needed so that performance comparison between our system and commercial products can be performed. Clinical experiments and tests on hearing impaired subjects are needed to better examine the noise canceling effect and system performance. The constraints of proposed system usage in daily life and child patients due to RF nature remain to be investigated.

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APPENDICES

A. Frequency Response Data File for Microphone Model

```
REM this model data file is for EK3024microphone
REM Freq in Hz
REM Gain in dB(relative to 1V), Gain is the ratio of output voltage to Input SPL.

BEGIN DSCRDATA
% INDEX    Freq Gain
1 100 -73.0
2 125 -71.4
3 150 -70.8
4 175 -69.7
5 200 -69.4
6 250 -68.3
7 300 -67.4
8 400 -66.3
9 500 -65.7
10 600 -65.7
11 700 -65.6
12 800 -65.5
13 900 -65.4
14 1000 -65.3
15 1100 -65.2
16 1200 -65.2
17 1300 -65.2
18 1400 -65.0
19 1500 -64.9
20 1600 -64.6
21 1700 -64.5
22 1800 -64.3
23 1900 -64.1
24 2000 -64.2
25 2100 -63.9
26 2200 -63.8
27 2300 -63.7
28 2400 -63.5
29 2500 -63.4
30 2600 -63.2
31 2700 -62.9
```

32 2800 -62.8
33 2900 -62.7
34 3000 -62.6
35 3200 -62.1
36 3400 -61.6
37 3600 -61.3
38 3800 -61.0
39 4000 -61.0
40 4250 -60.3
41 4500 -59.5
42 4750 -59.4
43 5000 -61.4
44 5250 -60.7
45 5500 -61.6
46 5750 -61.8
47 6000 -61.6
48 6250 -62.9
49 6500 -61.9
50 6750 -64.8
51 7000 -64.2
52 7250 -64.8
53 7500 -65.5
54 7750 -67.4
55 8000 -73.7
56 8250 -70.2
57 8500 -73.1
58 8750 -73.0
59 9000 -73.0
60 9250 -73.1
61 9500 -73.2
END

B. Frequency Response Data File for Receiver Model

REM this model data file is for BK1600 receiver model
REM 2rd Version
REM set frequency correspondent to the mic para.
REM Freq in Hz
REM Gain in dB(relative to 1V), Gain is the ratio of output voltage to Input SPL.

BEGIN DSCRDATA
% INDEX Freq Gain
1 100 59.39
2 125 59.35
3 150 59.14
4 175 59.86
5 200 60.29
6 250 60.41
7 300 59.60
8 400 58.05
9 500 56.76
10 600 56.00
11 700 55.56
12 800 55.12
13 900 54.53
14 1000 53.85
15 1100 53.61
16 1200 54.01
17 1300 53.96
18 1400 54.23
19 1500 54.35
20 1600 54.80
21 1700 55.22
22 1800 56.51
23 1900 58.77
24 2000 59.62
25 2100 60.60
26 2200 62.60
27 2300 64.33
28 2400 64.34
29 2500 63.07
30 2600 60.65
31 2700 58.73
32 2800 57.91
33 2900 55.61
34 3000 55.58
35 3200 54.09

36 3400 54.58
37 3600 55.58
38 3800 56.80
39 4000 52.09
40 4250 44.53
41 4500 40.26
42 4750 34.09
43 5000 31.42
44 5250 28.43
45 5500 25.12
46 5750 22.94
47 6000 12.95
48 6250 10.82
49 6500 14.10
50 6750 12.63
51 7000 11.74
52 7250 11.90
53 7500 11.90
54 7750 10.29
55 8000 10.48
56 8250 7.91
57 8500 8.57
58 8750 8.57
59 9000 5.01
60 9250 4.80
61 9500 4.63
END

C. Data File for Transmitter's Mixer

```
! DBL1.IMT
! @(#) $Source: /cvs/sr/src/geminiui/templates/dbl1.imt,v $ $Revision: 1.3 $ $Date: 2000/05/25
17:51:32 $
! Intermodulation table for double balanced mixer #1
! Signal Level (dBm)  LO Level (dBm)
     -10            7
!  M x LO ( Horizontal )  N x Signal (Vertical )
!\ 0  1  2  3  4  5  6  7  8  9  10  11  12  13  14  15
!
 99 26 35 39 50 41 53 49 51 45 65 55 75 65 85 99
 24  0 35 13 40 24 45 28 49 35 55 45 65 55 99
 73 73 74 70 71 64 69 64 69 65 75 75 85 99
 67 64 69 50 77 47 74 44 74 45 75 55 99
 86 90 86 88 88 85 86 85 90 85 85 99
 90 80 90 71 90 68 90 65 88 65 99
 90 90 90 90 90 90 90 90 90 99
 90 90 90 90 90 87 90 90 99
 99 95 99 95 99 95 99 99
 90 95 90 95 90 99 99
 99 99 99 99 99 99
 90 99 99 99 99
 99 99 99 99
 99 99 99
 99 99
 99
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D. Author’s Related Publications

1. Ram Singh Rana, Garg Hari Krishna, Zhang Liang and **Tang Bin,** “Hearing Aid Devices A Few Selected Research Issues,” *the 8th World Multi-Conference on Systemic, Cybernetics and Informatics*, pp. 80-84, Jul. 2004, U.S..

2. Ram Singh Rana, Zhang Liang, **Tang Bin** and Garg Hari Krishna, “An Enhanced Method and Behavioral Model for Noise Cancellation in Audio Devices,” *IEEE International Workshop on Biomedical Circuits & Systems*, pp. S2.6-11-S2.6-14, Dec. 2004, Singapore.

3. Ram Singh Rana, **Tang Bin**, Zhang Liang, Garg Hari Krishna ,and Wang De Yun, “Wireless Hearing Aid System Simulations using Advanced Design System™: A Behavioral Modeling Approach,” *the 27th Annual International Conference of the IEEE Engineering in Medicine and Biology*, pp. 5.3.1-6, Sept.2005, Shanghai, China.

4. **Tang Bin**, Hari Krishna Garg, Zhang Liang and Ram Singh Rana, “Wireless Hearing Aids System Simulation”, *the 39th Annual Asilomar Conference on Signals, Systems and Computers*, Oct. 2005, accepted for presentation and publication.