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UMI
MICROELECTROMECHANICAL (MEMS) VLSI STRUCTURES FOR HEARING INSTRUMENTS

by

Sazzadur Chowdhury

A Thesis
Submitted to the College of Graduate Studies and Research
Through Electrical and Computer Engineering
In Partial Fulfillment of the Requirements for
The Degree of Master of Applied Science at the
University of Windsor

Windsor, Ontario, Canada

2000

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ABSTRACT

In this thesis two microelectromechanical (MEMS) structures have been designed for use in a hearing instrument. MEMS realization of a capacitive microphone has been designed that has a higher sensitivity in mV/Pa than the sensitivity of the existing MEMS microphones together with an extended audio frequency response range. The microphone is designed using a low residual stress polysilicon germanium diaphragm of thickness 0.8 µm and an area of 2.5 x 2.5 mm², an airgap of thickness 3.0 µm and a silicon nitride backplate of thickness 1.0 µm with 1280 acoustical ports. Each acoustical port has a dimension of 30 x 30 µm². The microphone is constructed using a combination of surface and bulk micromachining techniques in a single wafer process to achieve a sensitivity of 52 mV/Pa and a frequency response extending up to 18 kHz. The electromechanical analysis of the microphone structure is carried out using IntelliSuite MEMS software with an accuracy of convergence of 0.005 µm. The simulation results match very closely with the analytical values. MEMS realization of an electromagnetic microactuator features a novel surface micromachined modular fabrication technique. A magnetomotive force of 4.5 AT is realized by electrodeposition of 3200 copper coil turns and the associated supermalloy core segments in a modular fashion. The advantage of this modular implementation is the magnetomotive force characteristics of the microactuator can be modified by changing the required number of modules. At a maximum coil current of 1.4 mA the microactuator consumes a power of 5.5 mW. Acoustical signal from the external world will be captured by the microphone. The output signal from the microphone will undergo signal conditioning and processing in a VLSI signal processor. The output of the signal processor will drive the electromagnetic
microactuator to generate a time varying magnetic field that will interact with the magnetic field produced by a micro permanent magnet that has been implanted on the round window of the Cochlea and will force the permanent magnet into vibrational motion. The motion of the implanted magnet will develop traveling waves on the Basilar membrane inside the Cochlea to give a hearing capability.
To my parents
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Chapter 1

INTRODUCTION

1.1 Introduction

A significant amount of research work has been carried out to develop hearing instruments to improve hearing capability of the people suffering from hearing impairment of different kinds. A conventional hearing instrument essentially consists of a microphone, battery, an amplifier and a loud speaker. The acoustical signal captured by the microphone is amplified and is fed to the loud speaker. Recent digital hearing instruments using advanced digital signal processing like multi channel compression, multiple memory and intelligent signal processing has improved the performances of the hearing instruments and the satisfaction of the user [1]. With the great development of VLSI microelectronics technology, it is now possible to incorporate greater functional modules of electronics circuits in a very small area, and hearing instruments now can be positioned completely inside the ear canal [2].

These hearing instruments suffer a major disadvantage of back-to-front coupling of sound due to close proximity of the loudspeaker and the microphone that results in a high-pitched squeal. Complete blockage of the ear canal by categories of hearing instruments knows as CIC (completely in the canal), ITC (in the canal) or IEC (in the ear) [2] results in an effect known as occlusion where outside sounds such as music are overwhelmed by internal sounds such as breathing or talking. Because the ear canal’s natural resonance is
significantly altered, the resulting sound can be unnatural and highly distorted [3]. The use of some specific type of hearing instrument (CIC, ITC, IEC, etc.) also depends on the type of hearing loss, e.g. a CIC type hearing instruments can’t be used if the hearing loss is profound [4]. The reason is in such situations, the amplifier is required to have a high gain, and that will increase the back-to-front coupling or feedback.

Again, as the amplifier amplifies the desired acoustic signal, it also amplifies the noise. Recent multiple memory hearing instruments apply adaptive cancellation of noise, some of them dynamically and some need a manual setting change. These programmable advanced digital hearing instruments have steeper learning curve for fitting and their cost is high. Additionally, most programmable instruments require the clinic to invest in equipment dedicated to a specific hearing instrument [5].

A completely different approach to hearing instrument design is proposed in this research work where MEMS (micro electro mechanical systems) technology is utilized to build a hearing instrument based on the concept of acousto-magnetic transduction. The system incorporates a MEMS capacitive microphone, a VLSI signal processor, a MEMS electromagnetic microactuator and a permanent micro magnet. The capacitive microphone along with the VLSI signal processing unit and battery will be implanted in the ear canal. The electromagnetic microactuator will be implanted in the middle ear cavity and the permanent micro magnet will be implanted over the round window of the cochlea using a bio-compatible adhesive.

External sound captured by the MEMS capacitive microphone will be converted to an equivalent current by a VLSI signal processor. Additional advanced digital signal
processing circuit modules will also be included in the VLSI processor. Both the
capacitive microphone and the VLSI signal processor along with a very small battery will
be implanted inside the ear canal that will occupy a small volume inside the ear canal and
will not block it. The output of the VLSI signal processor will be transmitted to the
implanted electromagnetic microactuator by a special under skin very thin cable pair
through a micro plug-socket arrangement. The time varying magnetic field that will be
generated by the electromagnetic microactuator due to a time varying coil current, will
force the micro permanent magnet implant into vibrational motion. The motion of the
micro permanent magnet implant will provide the required pressure difference between
the oval and the round window of the cochlea to enable hearing [6].

In this method the main acoustic pathway through the ossicular chain is not disturbed.
Depending on the degree of hearing loss, the coil current can be adjusted so that the
combined volume velocity of the oval window and the round window of the cochlea will
be equal to that occurs during normal hearing process. As this method establishes a
second independent acoustical pathway through the external and middle ear, a generative
coupling between the normal hearing pathway and the acousto-magnetic transduction
pathway can be used to feed the inner ear.

As the MEMS components can be batch fabricated at a very low production cost like the
VLSI chips [7], these hearing instruments will be of low cost. Since there is no
loudspeaker in the system the problem of back-to-front coupling will be eliminated
completely and the problem of occlusion can be avoided. Different degrees of hearing
loss can be addressed just by adjusting the actuator coil current.
This research work concentrates on the design and fabrication methods of the MEMS components of the proposed system and the design parameters of the permanent magnet implant.

The MEMS capacitive microphone is designed using a poly silicon germanium diaphragm, an airgap and a backplate. The loss at high frequencies due to compression of the air in the airgap is minimized by providing acoustical ports in the backplate [8]. A sensitivity of 52 mV/Pa is expected for the microphone with a high frequency response extending to 18 kHz. The microphone has a foot area of 3 x 3 mm\(^2\) and the height is less than 100 microns without packaging.

The microphone is designed using a combination of surface and bulk micromachining techniques in a single wafer process. Using a polysilicon germanium diaphragm creates the CMOS integration capability using low temperature deposition techniques [9]. The output voltage signal is obtained from the capacitor microphone using a capacitive voltage divider network and amplified by a simple source follower. A 12 Volt DC voltage is used to bias the microphone. A Dickson type DC-DC voltage converter will be used for this DC bias [10].

The electromagnetic microactuator is designed using a novel modular fabrication method. Each module is implemented by electrodeposition of 200 turns of planar copper coil with Bisbenzo-cyclobutene (BCB) as inter-module and between-the-turns insulation and electrodeposition of high permeability supermalloy (80% Nickel) core. The actuator is constructed using the successive deposition of 16 identical modules repeating the process.
sequences used for the first. In this way, 3200 turn copper coil is realized to produce a magnetomotive force of 4.5 AT (Ampere-turns).

A very high coercivity and maximum energy product Neodymium-Iron-Boron type permanent magnet [11] is chosen for the implant over the round window. 1222 microns magnetic length of the permanent magnet will provide a magnetomotive force of nearly 21.5 AT. The total magnetomotive force of the coupled magnetic circuit at maximum coil current of 1.4 mA will be 26 AT accounting a 10% leakage flux. This amount of magnetomotive force is equivalent to a mechanical force of 250 μNewton. This force is equal to the force at the oval window of the cochlea when an acoustic wave of 94 dB SPL is incident at the ear entrance.

The MEMS capacitive microphone and the electromagnetic microactuator are simulated using IntelliSuite [12] from Intellisense Corporation of Massachusetts. The simulation results are consistent with the calculated values.

1.2 Thesis Organization

This thesis is divided into 7 chapters.

Chapter one provides an introduction for the research work carried out. It gives a general idea about the currently available technologies to improve hearing capabilities to the deaf people who suffer from moderate to profound hearing loss. A brief discussion on merits and limitations of these technologies is provided. The idea of hearing instruments developed on acousto-magnetic transduction system is presented.
Chapter two presents a brief overview of human ear anatomy, normal hearing process in human, general causes of hearing impairment and types of hearing losses and also presents the state-of-the-art in hearing instruments and their limitations.

Chapter three presents a brief discussion on MEMS technology. The idea of MEMS based hearing instrument is presented.

Chapter four presents the design requirements for the MEMS hearing instrument. The actual physical requirements for design of the electromagnetic microactuator and the permanent magnet implant is investigated and the required design parameters are set. Physical positioning of the implants are also discussed.

Chapter five is devoted to the design of the capacitive microphone. The principle of operation, design considerations, material selection, fabrication process, 3D FEA simulation, capacitive readout circuitry and required power supply circuitry are presented.

Chapter six presents the design of the magnetic circuit. The permanent magnet design, design of the electromagnetic microactuator, material selection, modular fabrication concept, fabrication process, are presented.

Chapter seven presents a concluding summary of the research work carried out, future possibilities and the suggestions and remarks.
Chapter 2

OVERVIEW OF THE HUMAN AUDITORY SYSTEM

This chapter provides a brief description of human ear anatomy, physiological process of hearing, general causes of hearing impairment and characterization of hearing loss. Present state-of-the-art of hearing instruments is given and the limitations of these hearing instruments are discussed.

2.1 Human Ear Anatomy

Human ear is composed of a number of internal organs and fluids and their functioning is really complex. In this section only the major sections of the ear is described with a brief description of their function (Figure 2-1)[13].

2.1.1 Outer Ear

The outer ear captures sound from the external media, amplifies it and sends it to the middle ear. The major organs that are included in the outer ear are:

*Pinna:* The pinna is composed of cartilage and has a relatively poor blood supply. Its presence on both sides of the head allows us to localize the source of sound from the front vs. the back. Our ability to localize from side to side depends on the relative intensity and relative phase of sound reaching each ear and the analysis of the phase/intensity differences within the brainstem.

*Cartilaginous portion of external ear canal:* The ceruminous and sebaceous glands in the cartilaginous portion of the ear canal combine to produce cerumen. The total length of
the ear canal in adults is approximately one inch, which gives it a resonance frequency of approximately 3400 Hz, an important frequency region for understanding speech.

**Bony portion of external ear canal:** The bony portion of the ear canal is surrounded by the mastoid bone, occupies the inner third, and is very tender. Occasionally, completely in the canal (CIC) hearing instruments will reach as far as this portion of the canal.

**Tympanic membrane:** The tympanic membrane actually has three layers, with the outer layer continuous with the skin of the outer ear canal. The upper portion of the TM is called the pars flaccida, while the lower portion is called the pars tensa. The central portion of the pars tensa provides the active vibrating area in response to sound. The TM is a continually growing structure, which allows it to close if it has a hole in it and to extrude a ventilation tube.

---

Figure 2-1  A simplified diagram of human ear showing major functional sections
**Mastoid air cells:** A portion of the temporal bone (surrounding the ear). Under normal circumstances this honeycombed area cells are filled with air. They can fill with fluid or pus when chronically infected.

### 2.1.2 Middle ear

The middle ear serves as an impedance-matching transformer, matching the impedance of air in the ear canal to the impedance of the perilymph of the inner ear.

**Malleus:** The malleus is the most lateral (toward the side of the head) of the three ear bones (ossicles) in the middle ear. The long process of the malleus is attached to the inner layer of tympanic membrane. When the TM vibrates in response to sound, the malleus vibrates in concert.

**Incus:** The incus is attached to the malleus, and so vibrates as the malleus vibrates. The long process of the incus is also attached to the head of the stapes. Because the long process of the incus is slightly shorter than the long process (manubrium) of the malleus, incoming sound is given a slight (2.5 dB) boost in energy. This is referred to as the lever advantage.

**Stapes:** The stapes has a footplate and a superstructure. Its footplate is seated in the oval window, which separates the middle ear from perilymph of the inner ear. As the long process of the incus vibrates, so does the footplate of the stapes. Because the vibrating area of the tympanic membrane is larger than the area of the stapes, incoming sound is given a significant boost in energy of over 20 dB. This is referred to as the hydraulic advantage.
Round Window Niche: The round window is located within the round window niche. The round window is the most basal end of the scala tympani, and allows release of hydraulic pressure of perilymph that is caused by vibration of the stapes within the oval window.

Eustachian Tube: The eustachian tube connects the middle ear with the nasopharynx of the throat. This tube "opens" with swallowing or coughing to equalize pressure between the middle ear and ambient pressure that is found in the throat.

2.1.3 Inner Ear

Cochlea: A snail shaped structure that is the sensory organ of hearing. The vibrational patterns that are initiated by vibration of the stapes footplate set up a traveling wave pattern within the cochlea. This wavelike pattern causes a shearing of the cilia of the outer and inner hair cells. This shearing causes hair cell depolarization resulting in all or none neural impulses that the brain interprets as sound.

Vestibular Labyrinth: The vestibular labyrinth is comprised of the saccule and utricle, sense organs of balance which inform our brain about our linear position in space. The horizontal, anterior, and posterior semi-circular canals are also part of our vestibular labyrinth, and inform our brains about rotational movement in space.

VIII Nerve: The VIII nerve, also known as the auditory nerve, transmits information from the cochlear and vestibular labyrinth to our brains. Essentially, it is the transmission line from the sense organs to the central processor of the brain.
Facial Nerve: The facial nerve is the VII cranial nerve, and travels in parallel with the VIII cranial nerve through the internal acoustic canal. The facial nerve innervates the face and provides both sensory and motor function to the face.

Scala Vestibuli: One of the three partitions within the cochlea that is filled with perilymph, which attaches via the cochlear aqueduct (not shown) to the subarachnoid space. Perilymph, therefore, is the same as cerebro spinal fluid (CSF). Perilymph has a low Potassium - K⁺ - concentration and a high Sodium - Na⁺ - concentration.

Scala Media: Also one of the three partitions within the cochlea, the scala media is filled with endolymph. In contrast to perilymph, endolymph has a high K⁺ concentration and a low Na⁺ concentration. The endolymph that is found within the scala media is continuous with the endolymph of the saccule, communicating through the ductus reuniens.

Scala tympani: The third partition, the scala tympani also contains perilymph. The scala vestibuli and the scala tympani are continuous with each other at the most apical end of the cochlear partition called the helicotrema.

Organ Of Corti: Within the scala media, the organ of Corti is the sense organ of hearing. The outer and inner hair cells of the Organ of Corti change vibrational energy into neural energy, that is transmitted via the VIII nerve to the brain.

Reissner’s membrane: Separates the endolymph of the scala media from the perilymph of the scala vestibuli.

Tectorial membrane: A delicate, flexible, gelatinous membrane overlying the sensory receptive inner and outer hair cells. The cilia (hair like tufts that extend from both outer
and inner hair cells) of the outer hair cells are embedded in the tectorial membrane. For inner hair cells, the cilia may or may not be embedded in the tectorial membrane. When the cochlear partition changes position in response to the traveling wave, the shearing of the cilia is thought to be the stimulus that causes depolarization of the hair cells to produce an action potential.

**Stria vascularis:** A highly vascularized layer of cells that is thought to secrete endolymph.

**Outer hair cells:** There are three rows of approximately 12000 outer hair cells. Although they are much greater in number than the inner hair cells, they receive only about 5% of the innervation of the nerve fibers from the acoustic portion of the VIII nerve. These cells contain muscle-like filaments that contract upon stimulation and fine tune the response of the basilar membrane to the movement of the traveling wave. Because of their tuned response, healthy outer hair cells will ring following stimulation. This "ringing" provides the sound source for Otoacoustic Emissions.

**Inner hair cell:** There is one row of approximately 3500 inner hair cells. These cells receive about 95% of the innervation from the nerve fibers from the acoustic portion of the VIII nerve. These cells have primary responsibility for producing our sensation of hearing. When lost or damaged, a severe to profound hearing loss usually occurs.

**Tunnel of Corti:** A space filled with endolymph that is bordered by the pillars of Corti and the basilar membrane.

**Basilar membrane:** A ribbon like structure upon which rests the organ of Corti. The basilar membrane is less wide but stiffer at its basal end toward the oval window, but is
wider and more compliant toward its apical end away from the oval window. Because of these two gradients of size and stiffness, high frequencies are coded at the basal end with low frequencies progressively coded toward the apical end.

**Osseous spiral lamina:** A delicate bony plate that extends from the mid portion of the coiled cochlea, helping to separate scala vestibuli from scala tympani.

**Pillars of corti:** Supporting cells that bound the tunnel of Corti. The tunnel of Corti runs the entire length of the cochlear partition.

**Spiral ganglia:** The cell bodies of nerve fibers that innervate the inner and outer hair cells.

**Inner sulcus:** Inert supporting cells to the inner hair cells.

**Dieter's cells:** Inert supporting cells of the outer hair cells.

**Hensen's cells:** Inert supporting cells.

**Claudius' cells:** Inert supporting cells.

### 2.2 Physiological Process of Hearing

The environmental sound reaches the cochlea in a variety of mechanisms but the primary acoustic pathway for sound conduction to the cochlea is through the ear canal, tympanic membrane, the ossicles and the stapes footplate. The environmental sound enters the ear canal through the concha and gets amplified due to a horn effect. The ear canal supports an infinite number of cross-sectional acoustical modes and the plane wave is the first mode of them. Since the human ear canal is long and slender, these modes are uncoupled and maintain their individual energies as they propagate with the rigid wall portions of
the canal. Below 18 kHz in an average size adult ear, the plane wave is the only mode that propagates along the length of the ear canal. The remainder modes are trapped near the concha and the tympanic membrane. As the amplified sound wave reaches the tympanic membrane, it sets the tympanic membrane into vibrational motion. When the sound pressure moves the tympanic membrane, motion of the middle ear ossicles causes the stapes footplate to create pressure changes in the cochlear fluids [14]. These pressure changes can be thought of as the combination of two components. The first component is the average of the scala tympani and scala vestibuli pressure and this average pressure component propagates down the cochlea from the stapes as a compressional wave in the fluid that travels at approximately the speed of sound in the air. This average pressure wave is believed to have little influence on the motion of the basilar membrane.

The second pressure component is the pressure difference between scala vestibuli and scala tympani and is the component of the pressure that moves the basilar membrane. This second component propagates at a speed that is much slower than the compressional waves and the speed depends on the cross-sectional area of the fluid compartments and the mechanical properties of the basilar membrane and organ of corti.

The pressure difference is created because the round window equalizes the pressure between scala tympani and the middle ear cavity at the base of the cochlea. The result is that pressure difference between scala vestibuli and scala tympani at the round window is approximately equal to the pressure produced in scala vestibuli by the stapes footplate. This pressure difference produced by the combination of the stapes and the round window causes the motion of the basilar membrane to take the form of a traveling wave that that propagates from base to apex. The cochlea is capable of propagating traveling
waves on the basilar membrane in both directions and it is only because the round window is located in the base that the sound induced waves travel towards the apex.

The traveling waves on the basilar membrane which is created by this pressure difference has many of the same characteristics as a surface wave on the ocean. Although the waves can travel long distances, the fluid itself only moves back and forth as the wave passes by. A major difference, however is that in the ocean surface wave, the restoring force acting on the surface is gravity acting on the water while in the cochlea the major restoring force is the stiffness of the basilar-membrane-organ of corti complex. The stiffer the organ, the faster the traveling wave. The propagation velocity of the basilar membrane traveling wave also depends on the fluid inertia in the scalae with the higher the inertia, the slower the traveling wave.

The mechanical action of the basilar membrane and the organ of corti is more complex. At acoustical frequencies, the inertial and the viscous forces are also important. The combination of the inertia and the stiffness of the basilar membrane and the organ of corti cause the organ to exhibit a resonant frequency that is graded from base to apex. The highest resonant frequencies are found in the base because the effective mass is lower and the effective stiffness higher that in the apex.

For sinusoidal stimuli, the traveling wave on the basilar membrane propagates towards the base until it nears the region with a resonant frequency near that of the stimulus frequency. The wave then slows, the magnitude increases to a peak and then the wave decays rapidly beyond the peak. The location of the peak is referred to as the “best place” or “characteristic place” for the stimulus frequency and the frequency that best excites a particular place is called the “best frequency” or “characteristic frequency”.
The frequency-place map and the height and shape of the traveling wave peak depend not only on the passive mechanical properties of the organ and surrounding fluids but also on the active properties of the outer hair cells. The outer hair cells are capable of both sensing the motion of the organ of Corti and producing forces that act back on the organ through a unique form of cellular motility. This motility takes the form of a voltage dependent length change where a change in the outer hair cell membrane potential results in a change in the length of the cell body (soma). Depolarization of the cell results in contraction and hyperpolarization results in extension and this process can function at frequencies in the range of 10's of kHz.

2.3 General Causes of Hearing Loss

General causes of hearing impairments are listed below [15]-

- Middle ear infection
- Exposure to loud or constant noise
- Heredity
- Illness or birth defects
- The natural aging process
- Traumatic injury
- Ototoxic medications
- Tumors
2.4 Types of Hearing Loss

Hearing loss occurs any time there is structural abnormality and/or dysfunction of any part of the hearing mechanism. Loss of hearing may occur in just one or both ears (unilaterally or bilaterally). It can occur suddenly, or fluctuate, or become more progressive. Other symptoms that may be associated with hearing loss include tinnitus, dizziness, ear "fullness" and increased loudness of sounds.

There are four general types of hearing loss (Figure 2-2) [16]. Damage or dysfunction in the outer or middle ear causes conductive hearing loss. Damage or dysfunction in the inner ear causes sensorineural hearing loss. If there are both conductive and sensorineural components to the loss, it is called a mixed hearing loss. Occasionally the

![Diagram of the ear showing sensory and neural loss with conductive and sensorineural loss]

Figure 2-2 The locations of different types of hearing losses

hearing mechanism in the outer, middle, and inner ear functions normally, but the brain
does not interpret sounds and/or speech accurately, this is potentially a central auditory processing disorder.

2.4.1 Conductive Hearing Loss

Conductive hearing loss results from a problem with the outer or middle ear. Very simply, something prevents sound from being conducted through the outer and middle ear to the inner ear. In conductive hearing loss, the sensory and neural components of the inner ear are completely functional. Common causes of conductive loss include ear canal blockage (by ear wax or foreign body), ear infections, eardrum perforation, tympanosclerosis (scarring of the ear drum), congenital malformation of the outer or middle ear structures, cholesteatoma, otosclerosis and ossicular discontinuity. Treatment for conductive hearing loss depends on the cause and the extent of the loss, and may include surgery or hearing instruments, while antibiotics are usually indicated for ear

Figure 2-3 An audiogram of the left ear showing threshold of normal hearing
infections. An audiogram of left ear showing the normal hearing in given in Figure 2-3 and an audiogram showing effect of conductive hearing loss [17] is given in Figure 2-4.

![Audiogram](image)

Figure 2-4 An audiogram of the left ear showing the effect of conductive hearing loss

### 2.4.2 Sensorineural Hearing Loss

Sensorineural hearing loss results from either damage to the cochlea (inner ear) or the auditory (hearing) nerve. This type of loss is sometimes referred to as "nerve" loss, although this term is usually inaccurate since most sensorineural hearing loss is sensory, not neural. Disease or degeneration most often damages the hair cells, not the nerve. When hearing is tested, both sensory and neural hearing loss patterns appear the same way. Thus, the term sensorineural hearing loss is used to describe both types of loss. An audiogram of left ear showing the effect of sensorineural hearing loss [17] is shown in Figure 2-5. Most sensorineural hearing losses are permanent and cannot be treated.
medically or surgically except in certain circumstances, for example, sudden hearing loss, perilymph fistula and acoustic neuroma.

Figure 2-5 An audiogram of the left ear showing the effect of sensorineural hearing loss

2.4.3 Mixed type hearing loss

When both the conductive and sensory neural losses occur in an individual, the loss is termed as mixed type hearing loss. An audiogram of left ear showing the effects of mixed type (conductive and sensorineural) hearing loss [17] is shown in Figure 2-6.

Sensory loss from cochlear damage can occur with excessive noise exposure, ingestion of ototoxic medications, Meniere’s disease, and some immune diseases. Hearing loss that is present at birth is usually sensorineural, but also can be conductive especially in patients born with facial deformities. Aging causes the most common sensory hearing loss,
presbycusis (decreased hearing with age). By the age of 65, one-third of all people have significant hearing loss.

![Frequency in Hertz (Hz)](image)

**Figure 2-6** An audiogram of the left ear showing the effects of mixed type (conductive and sensorineural) hearing loss

Individuals with sensorineural hearing loss can usually become excellent hearing instruments users. Ultimate success with hearing instruments may be influenced by a person's ability to recognize words. Some people's ears with sensorineural hearing loss are like radios that aren't quite tuned in—speech may be loud enough but not clear. Unfortunately, if a person's ears create distortions that cause poor word recognition, hearing instruments may not accommodate this deficit. That is, the sound that is delivered through the hearing instruments may be crystal clear, but that's not the way the way it sounds to the ear with the distortions!
2.4.4 Characterization Of Hearing Loss

Based on the level of hearing losses in terms of acoustic units, hearing losses are characterized as mild, moderate, moderately severe, severe and profound hearing loss.

Table 2-1 lists the acoustical units related to each type of hearing losses [18].

<table>
<thead>
<tr>
<th>Hearing loss</th>
<th>Characterization</th>
</tr>
</thead>
<tbody>
<tr>
<td>-10 dB to 25 dB</td>
<td>Normal range</td>
</tr>
<tr>
<td>26 dB to 40 dB</td>
<td>Mild hearing loss</td>
</tr>
<tr>
<td>41 dB to 55 dB</td>
<td>Moderate hearing loss</td>
</tr>
<tr>
<td>56 dB to 70 dB</td>
<td>Moderately severe hearing loss</td>
</tr>
<tr>
<td>71 dB to 90 dB</td>
<td>Severe hearing loss</td>
</tr>
<tr>
<td>Over 90 dB</td>
<td>Profound hearing loss</td>
</tr>
</tbody>
</table>

Table 2-1 Characterization of hearing loss in terms of acoustical units

2.5 Present State-of-the-Art of Hearing Instruments

The conventional hearing instruments, both analog and digital are comprised essentially of a microphone to capture sound from the external world, an amplifier to amplify the captured sound and a loudspeaker to feed the amplified signal to the inner ear and a battery for power supply.

Recent programmable digital hearing instruments use advanced signal processing that automatically adjusts the amount of amplification (gain) by employing compression technique to raise the level of low amplitude and to limit the level of high amplitude sound as feed from the microphone [19]. Different signal processing configurations can
vary the amount of compression, as well as the loudness level at which compression is activated.

Some programmable hearing instruments have multi-channel capability where each channel can be adjusted independently so that different advanced signal processing schemes can be applied to each frequency region.

Some programmable hearing instruments have multiple memories, allowing them to store more than one frequency response and allow the user to choose from different frequency responses and signal processing options with a remote control or by pressing a button on the hearing instrument. Some use automatic adaptive cancellation of environment and microphone noise or feedback. This is useful for those who communicate in many different listening situations or have fluctuating hearing loss. The audiologist and the hearing instrument wearer together decide which hearing instruments responses to store in memory. The programs chosen are based on the wearer's most frequent and demanding listening situations (for example, attending meetings, listening to music or going to a restaurant). Some programmable hearing instruments have two separate microphones, one that picks up sound from a broad area and one that picks up from a narrower range [19]. There are different kinds of hearing instruments commonly available based on their position in the ear. Major ones are listed below [20]:

2.5.1 Completely-in-the-canal hearing instruments (CIC)

Completely-in-the-canal hearing instrument (CIC) is the smallest least visible custom hearing device. CIC hearing instruments are very compact, yet they have enough power for mild to moderately severe hearing loss.
2.5.2 In-the-canal hearing instruments (ITC)

In-the-canal hearing instrument (ITC) is a small and cosmetically appealing custom device with ample power to benefit patients with a wide range of hearing loss.

2.5.3 In-the-ear hearing instruments (ITE)

In-the-ear hearing instrument (ITE) are often located in a custom-made ear shell that fits securely in the outer ear, offering easy positioning and adjustment by the user. It features a wide choice of options such as a telephone pick-up coil and user-operator tone controls.

2.5.4 Behind-the-ear hearing instruments (BTE)

Behind-the-ear hearing instrument (BTE) is connected to a custom-made ear shell and offers special programming and power as well as flexibility for use with telephones and assistive listening devices.

2.5.5 Body hearing instruments

Body hearing instrument is housed in a portable case that can be carried in a pocket. A cord connects the hearing instruments to a receiver in the ear. These are the most powerful hearing instruments and are used for the most severe hearing losses.

2.6 Limitations of Conventional Hearing Instruments

All the conventional hearing instruments suffers from a great disadvantage that due to the close proximity of the loudspeaker to the microphone, a front to back coupling of the sound energy occurs and that gives rise to a high pitch squeal or whistling [3], [5], [21].
Also the amplifier used in a hearing instrument amplifies a significant amount of the background noise as well as the desired signal.

A major drawback of all the hearing instruments except BTE mentioned above is the ear canal is completely obstructed with either all or part of the hearing instrument that creates an effect known as occlusion, where outside sounds such as music are overwhelmed by internal sounds such as breathing or talking. Because the ear canal's natural resonance is significantly altered, the resulting sound can be unnatural and highly distorted. Occlusion also results in the inability to differentiate the direction of sounds, as well as the inability to adequately differentiate between background noise and other more important sounds, such as conversation [3].

The disadvantages of advanced digital hearing instruments are their steeper learning curve for fitting and their cost. Additionally, most programmable instruments require the clinic to invest in equipment dedicated to a specific hearing instrument [5].

The ideal would be for the user to have the same experience in all the environment as does an individual with unimpaired hearing with no or minimum amount of user intervention.

A suitable solution to this problem can be using a different approach other than conventional one of using a microphone-loudspeaker pair with a signal processing unit in between of them.
Chapter 3

ACOUSTO-MAGNETIC TRANSDUCTION: SYSTEM CONCEPT

In this chapter the concept of a hearing instrument that can be realized using MEMS based acousto-magnetic transduction system is presented. First, a brief discussion on the newly evolving MEMS (Micro Electro Mechanical Systems) technology is given. Next the concept of acousto-magnetic transduction system is presented.

3.1 MEMS (Micro Electro Mechanical Systems) Overview

3.1.1 Introduction

MEMS (Micro Electro Mechanical Systems) is a new evolving technology. The primary goal of the MEMS initiative is to develop the technology to merge sensing, actuating, and computing in order to realize new systems. The philosophy behind MEMS is to incorporate microelectronics and micromechanical components in a single chip that will enable co-location of sense, compute, actuate, control, communicate and power functional units to bring enhanced levels of perception, control, and performance. Integrating a number of different fabrication technologies closed-loop, microscale control of electrical, thermal, fluid, magnetic, optical, and mass flux will be possible by this surface technology where a control phenomena on the microscale will cause large effects both on macroscale and microscale. MEMS will provide the advantages of small size,
low-power, low-mass, low-cost and high-functionality to integrated electromechanical systems both on the micro as well as on the macro scale [22].

3.1.2 Applications of MEMS technology

Applications of MEMS technology can be broadly categorized as 1) fluid sensing, control, and transport; 2) mass data storage, 3) optics and imaging, 4) inertial instruments, 5) radio frequency components and communication, 6) sensor and actuator networks or arrays, and 7) bio-medical appliances and systems [22].

3.1.3 Historical Background

The concept of making a mechanical part from a deposited layer of material by dissolving an underlying sacrificial layer thus freeing the element except where it is retained by an attachment to the base silicon surface was demonstrated first in the mid-1960's by Nathanson and coworkers at Westinghouse Research laboratory [23]. In the early 80's researchers at the University of California at Berkeley first fabricated polycrystalline silicon microstructures using a silicon dioxide sacrificial layer. Polycrystalline surface micromachining was quickly recognized as a promising technology and was employed at both academic and industrial laboratories. The successful operation of surface micromachined polysilicon electrostatic micromotors in the mid-1980's stimulated government and industrial funding for research in silicon MEMS. Then beginning in the 1990's a significant influx of government research capital (Japan, United States and Europe) promoted the technological revolution that resulted in the development of fully integrated complex microelectromechanical systems where sensors, actuators and control functions are cofabricated using micromachining and IC processing.
3.1.4 A Basic MEMS Design: A Cantilever Beam

The simplest MEMS structure is a cantilever beam. Over a silicon wafer, a thick layer of SiO\textsubscript{2} is deposited by thermal process. This SiO\textsubscript{2} layer is then patterned by Ultra-violet ray using an appropriate mask and etched by buffered Hydrofluoric acid (BOE) to form an anchor hole. Over the SiO\textsubscript{2} a thin layer of Polysilicon (a few microns thick) is then deposited by Low Pressure Chemical Vapor Deposition (LPCVD) process and patterned by a photoresist and etched. The whole structure is then immersed in a solution of buffered Hydrofluoric acid (BOE), which dissolves the SiO\textsubscript{2} sacrificially. The polysilicon layer then becomes suspended like a cantilever beam connected to the base silicon wafer through the anchor hole. When this structure is set in any industrial structure where a pressure or vibration measurement or actuation is necessary, the polysilicon cantilever beam will vibrate according to external actuation and this vibration will be converted to an electrical signal capable to drive/actuate some mechanism. The control and drive CMOS electronics can be accommodated in the same chip and batch fabricated.

3.1.5 MEMS Materials

Although it was started with <100> oriented Silicon wafer as the substrate, polysilicon as the structural material, SiO\textsubscript{2} and PSG (Phosphosilicate glass) as sacrificial or mask layer, and silicon nitride (Si\textsubscript{3}N\textsubscript{4}) as passivation layer, now different metals and compounds like Ferro silicon (Electromagnetic structure), Copper (inductor/micro coil), Gold, Aluminum, Titanium, Chromium, alloys of Nickel, Porous silicon, etc. are used to build MEMS structures.
3.1.6 Micromachining Processes

Surface epitaxy like conventional VLSI is used some times but the three major techniques for development of MEMS structure are- surface micromachining, bulk micromachining and wafer bonding. Deposition of different layers are carried out by thermal evaporation, electron beam evaporation, plasma deposition, Low Pressure Chemical Vapor Deposition (LPCVD), Atmospheric Pressure Chemical Vapor Deposition (APCVD), Plasma Enhanced Chemical Vapor Deposition (PECVD), Argon ambient magnetron sputtering, electrodeposition, spin coating, lift-off, etc.

3.1.7 LIGA

It is also possible to develop only a single micron thick- several microns high microstructures (high aspect ratio vertical structure) over a Silicon wafer using the expensive and novel process known as LIGA- the German word for (Lithographie, Galvaniformung, Abformung). In LIGA a thick photoresist is patterned by a deep X-Ray radiation from a synchrotron. After the photoresist is developed a mold is formed. The mold is then filled up by electrodeposition (electroplating) of the required metal. The photoresist is then stripped. Finally the top unwanted portion of the metal is micro milled by a precision milling machine. Or after stripping the photoresist, plastic or glass can be deposited and the metal can be etched sacrificially to obtain a plastic or glass structure [24]. In this way, micro gears, wobble motors, crankshafts, gratings, mirror arrays, turbine shafts and blades can be built in the microdomain.
3.1.8 Etching Processes

Etching processes include both isotropic (rounded, due to equal etch rates in all directions) and anisotropic (orientation dependent that etch much faster in one direction than in another resulting in perfectly flat surfaces and well-defined sharp angles) etching. Another key distinguishable feature of etchants is the phase of the reactants: liquid (wet etching), vapor and plasma (dry etching).

The most common isotropic wet silicon etchant is HNA - a mixture of hydrofluoric acid (HF), Nitric acid (HNO₃) and Acetic acid (CH₃COOH). Anisotropic wet Si etchants include: 1) Alkali hydroxide etchants (KOH, NaOH, CsOH, and RbOH) that use SiO₂ or Si₃N₄ as mask material, 2) Tetra methyl Ammonium Hydroxide (TMAH), 3) Ethylene Diamine Pyrocatechol (EDP). A mixture of H₂PO₄, C₂H₄O₂ and HNO₃ is used to wet etch metals. SiO₂ (typically used as a mask for Si etch or as a sacrificial layer) is etched by using a hydrofluoric acid (HF) solution buffered with ammonium fluoride -a process known as Buffered Oxide Etch (BOE).

Isotropic non-plasma dry etch of Si is carried out by XeF₂ (Xenon difluoride). Laser-driven-vapor-phase etching of Si is done using a technique known as LACE which employs a typical Laser wavelength of 500 nm. The most common form of dry anisotropic etching of bulk Si are Plasma etching and RIE (Reactive Ion Etching) where external energy in the form of radio frequency (RF) power drives chemical reactions in low-pressure reaction chambers at a relatively low temperature. Common reactants for plasma and RIE etching are chlorofluorocarbon (CFC) gases, freon, sulphur hexafluoride (SF₆), bromine compounds and oxygen.
High aspect ratio dry etching methods known as Deep RIE relies on a high-density (inductively coupled) plasma source and an alternating process of etching and protective polymer deposition. The etching step uses SF₆/Ar with a substrate bias of -5V to -30V while a mixture of trifluoromethane (CHF₃) and Ar is used for polymerization and all sidewalls (exposed surfaces) are coated with Teflon like polymerized (CF₂) thin polymer layer [23].

Another technique known as sputter etch uses reactive ion sputtering from a magnetron to remove a very thin layer from the wafer.

In practice the above etching techniques are used selectively in different combinations to build a complete complex microstructure.

3.2 Acousto-Magnetic Transduction: System Concept

3.2.1 A View into the Middle Ear

From chapter 2 it appears that the oval and round window of the cochlea are the locations where external acoustomechanical signals act on cochlear fluid. If the response of the cochlea is linearly dependent on its inputs, then with sinusoidal pressure pressures $P_{OW}$ and $P_{RW}$ outside the oval and round windows, respectively, the cochlear response can be described as the sum of two terms:

$$\text{Cochlear response} = D (P_{OW} - P_{RW}) + \frac{1}{2} C (P_{OW} + P_{RW}),$$

where the first term in the right hand side of the equation is the difference mode term and is proportional to the difference in sound pressures at the cochlear windows. and the second term in the right hand side of the equation is the common-mode term and is
proportional to the sum of the sound pressure at the cochlear windows [25]. The complex coefficients D and C are the difference mode and common mode gains. If the cochlea is sensitive only to difference mode stimuli, i.e., the common mode gain C is zero, equation (3-1) simplifies to:

\[
\text{Cochlear response} = D (P_{\text{OW}} - P_{\text{RW}})
\]

(3-2)

If equation (3-2) holds and the cochlear system are reciprocal, the volume velocities at the oval and round windows must be equal in magnitude and opposite in phase, i.e., an inward stapes displacement of a volume of perilymph produces an outward round-window displacement of identical volume. Thus the difference mode term represents mechanisms that involve equal inner-ear fluid flow from the one window to the other [25].

Non-zero common mode term could arise if (1) intracochlear pressure is a stimulus to the receptors and/or (2) fluids flow through paths that don’t connect the two cochlear windows. Mechanisms that account for such fluid flow (“third window effects”) include flow out of the cochlea through the perilymphatic or endolymphatic ducts and flow produced by compression of cochlear structures (e.g., nerves, blood vessels, or scala media).

Voss, Rowsoski and Peake carried out measurements in cats to identify the complex coefficients C and D to determine the respective effects on cochlear fluid due to stimulation of either the oval window or the round window. Simultaneous stimulation of the oval and round windows [25]. Their results support the idea that the cochlea responds primarily to the difference in pressure between the oval and round windows (Figure 3-1).
The common mode response is difficult to measure because it is generally much smaller than the difference mode response [25].

The dominance of the difference mode response supports the idea of increasing difference mode input to cochlea to improve hearing. The work of Voss, Rowsoski and Peake suggests that if by some artificial means the round window of the cochlea can be pulled towards the tympanic cavity by an amount sufficient to compensate for the pressure shortage at the oval window, it is possible to create the required pressure
difference between the oval and round window of the cochlea as required by the normal hearing process.

This artificial means can be implemented using an electromagnetic microactuator positioned very close to the round window of the cochlea while a micro permanent magnet will be implanted on the round window of the cochlea using some biocompatible adhesive. Acoustical signal from the external world will be captured by a small microphone implanted in the ear canal. The output signal from the microphone will undergo signal conditioning and processing in a VLSI signal processor. The output of the signal processor that will be an analog time varying signal, having same shape characteristic as the external acoustical signal will drive the electromagnetic actuator to generate a time varying magnetic field. By this way an acousto-magnetic transduction can be realized to have a magnetic field that has a time-varying characteristic as an acoustical signal. This time-varying magnetic field will interact with the magnetic field produced by the micro permanent magnet that has been implanted on the round window of the cochlea and will force the micro permanent magnet into motion in a manner like the external acoustical signal. Eventually the round window will be forced into motion as it is a thin membrane structure and is tacked to the micro permanent magnet implant. Thus it will be possible to create the required pressure difference between the oval and the round window of the cochlea using an artificial manner. As the nature of this pressure difference will be very much similar to the pressure difference created during normal hearing process, the same way the cochlear fluid will flow to develop traveling waves in the basilar membrane. This eventually will excite the inner and the outer hair cells to send neuro-signal to the respective section of the brain.
Due to the small volume of the tympanic cavity, a MEMS realization of an electromagnetic microactuator will be best suited for this application. For completely implantable and to cover a wide range of audio spectrum, a MEMS realization of a capacitive microphone will also be the best. A block diagram of the proposed system is given in Figure 3-2.

**Acousto-Magnetic Transduction System**

**Block Diagram**

![Block Diagram](image)

Figure 3-2  Block diagram of the acousto-magnetic transduction system
Chapter 4

DESIGN REQUIREMENTS FOR MEMS IMPLANTS

One of the basic requirements for the proposed acousto-magnetic transduction system is that the magnetic circuit comprised of the electromagnetic microactuator and a permanent magnet implant must develop a time varying magnetic force on the round window of the cochlea so that the required pressure difference between the oval and the round window of the cochlea is established. An analog electrical model of the external and middle ear function developed by Rowsoski provides a means to calculate pressure at different locations inside the ear including the cochlea entrance. In this chapter the Rowsoski model is described and using the Rowsoski model the pressure/force at the oval window of the cochlea is measured analytically. Next, the considerations for placing external devices inside the tympanic cavity are discussed.

4.1 Analog Electrical Model of Human Ear

4.1.1 Acousto-Mechanical View of the Hearing Process

Environmental sound reaches the inner ear through a variety of mechanisms, but the primary path for sound conduction to the cochlea is through the coupled motion of the tympanic membrane, ossicles, and stapes foot plate. The input to the ear can be defined in terms of sound pressure $P_{PW}$ and direction (noted by the effective azimuth $\theta$ and elevation $\phi$) of an equivalent uniform plane wave. Diffraction of the plane wave by the head, body and pinna results in a sound pressure $P_{EX}$ and the volume velocity $U_{EX}$ at the
concha entrance that are dependent on direction (Figure 4-1). Waves of sound pressure and volume velocity travel through the concha and ear canal and interact with the mechanics of the tympanic membrane and middle ear to produce a pressure $P_T$ and volume velocity $U_T$ acting on the membrane. The directional dependence observed at the entrance to the ear canal remains essentially unaltered throughout later signal transmission stages. The tympanic membrane converts the pressure and volume velocity acting on its lateral surface into an effective force $F_U$ ON and mechanical velocity $V_U$ of the umbo (the umbo is the tip of the malleus handle, which is embedded in the tympanic membrane near the membrane's center). The umbo velocity is converted by translations and rotations of the ossicular chain into a piston-like motion of the stapes of velocity $V_S$.
and a force on the stapes footplate \( F_s \). The stapes velocity and force are integrated over the area of the footplate to produce a volume velocity \( U_s \) and pressure \( P_s \), which work against the acoustic impedance of the annular ligament and cochlea to produce sound pressure within the cochlear vestibule \( P_C \). The sound pressure and volume velocity within the vestibule produce the inner ear fluid motions associated with the transduction of sound to the hair cells and neural responses.

4.1.2 Acoustic Wave Characteristics

The human auditory area covers three decades of frequency (20 Hz to 20 kHz) and six orders of magnitude in sound pressure (2 x 10^{-5} to 20 Pa). The average displacement of air particles in an audible plane wave stimulus can vary from subatomic dimensions of less than 10^{-10} m to motions that are nearly of macroscopic magnitude 10^{-4} m. Given a sound propagation velocity of 343 m/sec in air, the wavelength of audible sounds varies from 17m to 1.7mm. This range includes wavelengths \( \lambda \) that are much larger than the dimensions of many ear structures through wavelengths that are much smaller than many ear dimensions.

4.1.3 The Air Characteristics in hearing process

As air has a low viscosity, the two physical features that play the largest role in determining how sound propagates within it are its static density \( \rho_0 \) and bulk modulus of compressibility \( B \). The compressibility of a fluid depends on whether the temperature of the fluid changes as it is rarefied or compressed. In the case of isothermal pressure changes, any change in the heat content of the fluid is immediately corrected by heat transfer to and from the fluid's surroundings. In such conditions, the compressibility of
the fluid is determined by the isothermal bulk modulus $B_I$. In cases in which no heat is 
allowed to flow between the fluid and the surroundings, the temperature of the fluid 
increase and decreases with each compression and rarefaction, and the compressibility of 
the fluid is determined by the adiabatic bulk modulus $B_A$. For sound of all but the lowest 
audible frequencies, the changes in temperature generated by sound pressure changes 
alternate so rapidly with each compression and rarefaction of the media that there is no 
time for heat to flow between the fluid and the surround, and we can consider the 
pressure changes to be adiabatic. The propagation velocity of sound $c$ in a low-viscosity 
fluid can be calculated from its adiabatic bulk modulus and density: 

$$ c = \sqrt{\frac{B_A}{\rho_0}} $$ 

where $B_A = \gamma B_I$ and $\gamma$ is the ratio of the specific heats. In the case of an ideal gas such as air, 
$B_I = P_0$, the static pressure, and $\gamma = 1.4$.

Because of the assumptions of low viscosity and adiabatic compression hold for most 
situations in hearing, we can generally think of the air in the external ear tube and middle 
ear spaces as a lossless media [6].

4.2 The Rowsoski Model

Based on the acousto-mechanical view of the hearing process described in section 4.1 
Rowsoski formulated an analog electrical circuit equivalent of external and middle ear 
mechanism [6]. The model is shown is Figure 4-2. The Figure 4-2 (A) describes the 
middle and inner ear mechanism as functional blocks and Figure 4-2 (B) elaborates the 
details of the blocks after the ear canal tube. In Figure 4-2(A), the left most block 
contains a Thevenin equivalent sound source where the equivalent source pressure is the
product of the sound pressure of the uniform plane wave stimulus and a directional term \( G_S^{oc} \) that accounts for the diffraction and scattering of sound by the head, body and pinna [26].

Assuming that the head is a rigid sphere, the source directional term \( G_S^{oc} \) is equal to the open-circuit pressure gain that relates the sound pressure on the surface of the rigid sphere \( P_{ss} \) with the sound pressure of a plane wave stimulus \( P_{pw} \). From this simplified assumption, the directional term \( G_S^{oc} \) can be expressed as [27]:

\[
G_S^{oc} (\omega, \theta) = \frac{P_{ss}(\omega, \theta)}{P_{pw}(\omega)} = \frac{1}{(ka)^2} \sum_{m=0}^{\infty} j^{m+1} L_m (\cos \theta) (2m + 1)
\]

\[ - j m (ka) - j N_m (ka) \]  

(4-1)

Where

\[ \Theta \] = Angle describing the position on the sphere relative to the direction of propagation of the plane wave

\[ \omega = 2\pi f \] , The radian frequency of the stimulus

\[ k = \frac{\omega}{c} \] , The wave number

\[ a \] = Radius of the sphere

\[ L_m \] = Legendre polynomial of order \( m \)

\[ J_m \] = Derivative of the \( m^{th} \) order Bessel function

\[ N_m \] = Derivative of the \( m^{th} \) order spherical Neumann function

The sound pressure at the opening of the external ear \( P_{ex} \) can be expressed in terms of the Thevenin equivalent pressure \( P_{pw} G_S^{oc} \) as:

40
\[ P_{EX} = P_{pw} G_s^{oc} \frac{Z_{EX}}{Z_{EX} + Z_R} \]  

(4-2)

Where \( Z_{EX} = P_{EX}/U_{EX} \), the acoustic impedance looking into the ear opening, \( P_{EX} \) and \( U_{EX} \) are the sound pressure and volume velocity at the entrance to the ear, and \( Z_R \) is the Thevenin impedance of the equivalent source and is equivalent to the radiation impedance looking out from the ear opening into the environment [26].

The next block of Figure 4-2(A) models the concha and external ear canal as a horn and tube of dimensions similar to those of the human external ear. Assuming that the pinna has a little effect on power gathering, the modeled horn and the tube transform \( P_{EX} \) and \( U_{EX} \) into \( P_T \), the pressure outside the tympanic membrane and the tympanic membrane volume velocity \( U_T \). The ratio of \( P_T \) and \( U_T \) defines the middle ear input impedance \( Z_T = P_T/U_T \), where this impedance represents the combination of the impedances of the tympanic membrane, ossicles, cochlea and the middle ear air space or cavity. The air spaces contribute to this impedance because the motion of the tympanic membrane changes the volume of the closed air space behind it (the Eustachian tube is normally closed), there by producing a sound pressure \( P_{CAV} \), within the space that pushes on the medial surface of the tympanic membrane. The acoustical impedance of the middle-ear cavity \( Z_{CAV} \) is expressed as

\[ Z_T = Z_T^{oc} + Z_{CAV} \]  

(4-3)

Where \( Z_T^{oc} \) is the input impedance of the tympanic membrane, ossicles and cochlea measured with "open cavities."
The pressure difference on both sides of the tympanic membrane $P_T - P_{CAV}$, is somewhat smaller in magnitude than $P_T$ and acts as the drive to the rest of the middle ear. Equation (4-3) points out that the effect of the middle ear air spaces on middle ear function depends on the relative magnitude of $Z_T^{OC}$ and $Z_{CAV}$. In humans, $Z_{CAV} > Z_T^{OC}$ and the effect of the cavities is small.

The middle ear cavity is modeled as a two port network and the pressure difference across the tympanic membrane $P_T - P_{CAV}$ acts as the drive on the rest of the model.

The first block of Figure 4-2(B) is a detailed representation of the block “middle ear cavity” in Figure 4-2(A). The block contains four acoustic elements: a capacitor that represents the compliance of the tympanic air space located directly behind the tympanic membrane $C_{TC}$, another capacitor to represent the compliance of the mastoid air volume $C_{MC}$ that is separated from the other space by aditus ad antrum, an acoustic inerntance that approximates the inerntance of the connecting aditus $L_A$ and a resistance $R_A$ associated with viscous losses in the aditus and mastoid air spaces.

The next block in Figure 4-2(B) models the tympanic membrane and the mallear attachment. The pressure difference $P_T - P_{CAV}$ causes the tympanic membrane to convert the pressure $P_T$ and volume velocity $U_T$ at the tympanic membrane into force $F_U$ and velocity $V_U$ at the umbo. A series branch with two compliances $C_T$ and $C_{T2}$, two resistances $R_T$ and $R_{T2}$ and two inertances $L_T$ and $L_{T1}$ accounts for the damping, mass and stiffness associated with “coupled” motions of the tympanic membrane and malleus. An ideal transformer $I: A_{TM}$ with units of inverse inertia converts the acoustic variables of so-
Figure 4-2 The Rowsoski analog electrical model of external and middle ear function.
und $P$, volume velocity $U$ and acoustic impedance $Z = P/U$ to mechanical variables of force $F$, velocity $V$ and mechanical impedance $Z^M = F/V$. The two mechanical elements of the parallel branch are a compliance $C^M_{75}$ and a resistance $R^M_{75}$ that account for relative (uncoupled) motions between the tympanic membrane and the malleus. In the block the capacitors are related to the tympanic membrane compliance ($1/$stiffness), the resistors are associated with losses and damping, the inductors are associated with acoustic and mechanical masses and the transformer is associated with the tympanic membrane.

The next block in Figure 4-2(B) converts the motion of the umbo $V_U$ to a piston like volume velocity of the stapes $U_S$ and is characterized by the transfer function $U_S/V_U$. The circuit element representation of this block includes: (a) a series resistance, a compliance and a mass that models the damping $R^M_{MI}$, compliance $C^M_{MI}$ and mass $L^M_{MI}$ within the malleus, incus and their supporting ligaments, (b) a dimensionless transformer $l_I : l_M$ that accounts for rotation of the malleus and incus complex, (c) a shunt branch with capacitor $C^M_I$ and resistor $R^M_I$ to account for the loss of stapes velocity from compression of the ossicular joints, (d) the mass of the stapes $L^S_M$, and (e) a second transformer $A_{FP} : 1$. The second transformer is the stapes footplate area that converts the force $F_S$ and piston like velocity of the stapes $V_S$ into pressure $P_S$ and volume velocity $U_S$.

The last two blocks in Figure 4-2(B) models the annular ligament and the cochlea. Both the impedances of the annular ligament and the cochlea are included as one port networks. The pressure $P_S$ acting on the ligament and the pressure $P_C$ at the cochlear
entrance depends on frequency and series combination of the ligament and cochlea acts as a voltage divider. At low frequencies, $P_s \gg P_c$ and at high frequencies $P_s = P_c$.

The values and units of each element of the model are shown in Table 4-1.

<table>
<thead>
<tr>
<th>Components/Block</th>
<th>Symbol</th>
<th>Value</th>
<th>Unit</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Transformers</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Tympanic transformer</td>
<td>$l:A_{TM}$</td>
<td>$1/(60 \times 10^{-5}) \text{ m}^2$</td>
<td>Inverse area</td>
</tr>
<tr>
<td>Ossicular transformer</td>
<td>$l_l : l_m$</td>
<td>$1/1.3$</td>
<td>Dimensionless</td>
</tr>
<tr>
<td>The footplate transformer</td>
<td>$A_{FP} : l$</td>
<td>$3.2 \times 10^{-6} \text{ m}^2$</td>
<td>Area</td>
</tr>
<tr>
<td><strong>Middle ear cavity</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Compliance of the mastoid air space</td>
<td>$C_{MC}$</td>
<td>$3.9 \times 10^{-11}$</td>
<td>m$^5$N$^{-1}$</td>
</tr>
<tr>
<td>Compliance of the tympanic air space</td>
<td>$C_{TC}$</td>
<td>$4 \times 10^{-12}$</td>
<td>m$^5$N$^{-1}$</td>
</tr>
<tr>
<td>Resistance of the aditus that connects two airspaces</td>
<td>$R_A$</td>
<td>$6 \times 10^6$</td>
<td>Pa s m$^{-3}$</td>
</tr>
<tr>
<td>Inertance of the aditus</td>
<td>$L_A$</td>
<td>100</td>
<td>Kgm$^{-4}$</td>
</tr>
<tr>
<td>Components/Block</td>
<td>Symbol</td>
<td>Value</td>
<td>Unit</td>
</tr>
<tr>
<td>------------------</td>
<td>--------</td>
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<td>------</td>
</tr>
<tr>
<td><strong>Tympanic membrane and the mallear attachment</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Series branch</td>
<td>$L_{TI}$</td>
<td>750</td>
<td>Kgm$^{-1}$</td>
</tr>
<tr>
<td></td>
<td>$L_T$</td>
<td>$6.6 \times 10^3$</td>
<td>Kgm$^{-1}$</td>
</tr>
<tr>
<td></td>
<td>$C_T$</td>
<td>$3 \times 10^{-12}$</td>
<td>m$^5$N$^{-1}$</td>
</tr>
<tr>
<td></td>
<td>$C_{T2}$</td>
<td>$1.3 \times 10^{-11}$</td>
<td>m$^5$N$^{-1}$</td>
</tr>
<tr>
<td></td>
<td>$R_T$</td>
<td>$2 \times 10^7$</td>
<td>Pa s m$^{-3}$</td>
</tr>
<tr>
<td></td>
<td>$R_{T2}$</td>
<td>$1.2 \times 10^7$</td>
<td>Pa s m$^{-3}$</td>
</tr>
<tr>
<td>Parallel branch</td>
<td>$C_{TS}^M$</td>
<td>$1.1 \times 10^{-3}$</td>
<td>mN$^{-1}$</td>
</tr>
<tr>
<td></td>
<td>$R_{TS}^M$</td>
<td>$4.3 \times 10^{-2}$</td>
<td>N s m$^{-1}$</td>
</tr>
<tr>
<td><strong>The ossicles and their supporting ligaments</strong></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Malleus and incus</td>
<td>$C_{MI}^M$</td>
<td>$\infty$</td>
<td></td>
</tr>
<tr>
<td></td>
<td>$L_{MI}^M$</td>
<td>$7.9 \times 10^{-6}$</td>
<td>Kg</td>
</tr>
<tr>
<td></td>
<td>$R_{MI}^M$</td>
<td>$7.2 \times 10^{-3}$</td>
<td>N s m$^{-1}$</td>
</tr>
<tr>
<td>Components/Block</td>
<td>Symbol</td>
<td>Value</td>
<td>Unit</td>
</tr>
<tr>
<td>-------------------------------</td>
<td>--------</td>
<td>-----------</td>
<td>---------</td>
</tr>
<tr>
<td>Stapes</td>
<td>$L^M_5$</td>
<td>$3.0 \times 10^{-6}$</td>
<td>Kg</td>
</tr>
<tr>
<td>Compression of ossicular</td>
<td>$R^M_j$</td>
<td>3.6</td>
<td>N s m$^{-1}$</td>
</tr>
<tr>
<td>joints</td>
<td>$C^M_j$</td>
<td>$4.9 \times 10^{-4}$</td>
<td>m N$^{-1}$</td>
</tr>
</tbody>
</table>

### Annular ligament

<p>| | | | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>$C_{AL}$</td>
<td>$9.4 \times 10^{-15}$</td>
<td>m$^2$N$^{-1}$</td>
<td></td>
</tr>
<tr>
<td>$R_{AL}$</td>
<td>0</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

### Cochlea

<p>| | | | |</p>
<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td>$R_C$</td>
<td>$2.0 \times 10^{10}$</td>
<td>Pa s m$^{-3}$</td>
<td></td>
</tr>
<tr>
<td>$L_C$</td>
<td>$2.4 \times 10^6$</td>
<td>Kg m$^{-4}$</td>
<td></td>
</tr>
</tbody>
</table>

Table 4-1  The element values of the Rowsoski analog electrical model of external and middle ear function

In formulating the model, Rowsoski assumed that the cochlea is a one port network such that all the volume velocity that enters the cochlea via the oval window leaves by way of the round window.
4.3 Applicability and Limitations of Rowsoski Model

The total external ear gain predicted by the Rowsoski model is similar to the measurements done physically by Shaw [28]. The middle ear input impedance $P_T/U_T$ and stapes motion $U_s/P_T$ predicted by the model match measurements quite well [29]. The predicted middle-ear pressure transfer ratio $P_P/P_T$ is similar to that measured in animals [30].

One big assumption was made in building the model is to lump all the directional responses into a single source term $G_{5oc}$. This assumption is consistent with the notion that the directional response is produced by structures peripheral to the canal but ignores the role of the concha in producing elevation and azimuth cues for localization.

Due to ethical reason, measurement of the exact pressure at different sections inside the ear in a living human being is not possible and similar experiments were carried out in cats and cadavers. In that sense the applicability of the model to identify the exact hearing process can only be on an approximation and more detailed association between structure and functions are being investigated.

On this basis, the Rowsoski model can be used as the basis to calculate the pressure exerted on the stapes footplate due to a sound wave incident at the pinna.

4.4 Other Mechanisms for Sensing Sound in Humans

The Rowsoski model of Figure 4-2 implies that stapes motion is the only input to the inner ear, yet it is well known that interruption of the incudostapedial joint produces only a 50—60 dB hearing loss. Clearly, there must be other mechanisms for sound to enter the inner ear. Those “other” mechanisms are thought to include bone conduction [31] and
direct acoustic stimulation of cochlear windows [32] [33]. The greatly decreased hearing sensitivity after interruption of the ossicular chain and model analyses [32] suggest that these other paths are not significant in normal ears, but can become important in middle ear pathology.

4.5 Considerations for Physical Placement of the Implants

To set the oval window of the cochlea into motion by coupled action of an electromagnetic microactuator and a micro permanent magnet implant, it is required to tack the micro permanent magnet implant over the round window of the cochlea by a biocompatible adhesive.

If the distance between the electromagnetic actuator and the permanent magnet implant is too large (for instance if the actuator is placed in the ear canal outside the tympanic membrane), the required magnetomotive force to produce the desired force will be very large. In that case the airgap length between the actuator and permanent magnet implant pole faces will be more than one inch and that will cause a considerable amount of magnetic flux to be lost in the form of leakage flux. To compensate for this leakage flux, an actuator with much higher magnetomotive force (almost 50%) will be required. To generate a magnetomotive force of that magnitude a current more than 200 mA will be required and the power consumption will be considerably high. Associated dissipation of heat will be too much for such a device to be placed inside the ear canal. Based on that consideration, it is decided to place the electromagnetic actuator inside the tympanic cavity. An investigation is required to carry out now to determine if the placing of these external devices may affect the natural resonance characteristic of the tympanic cavity or alter its effect in normal hearing.
Equation (4-3) points out that the effect of the middle ear air spaces on middle ear function depends on the relative magnitude of $Z_T^{oc}$ (input impedance of the tympanic membrane, ossicles and cochlea measured with "open cavities") and $Z_{CAV}$. In humans, $Z_T^{oc} > Z_{CAV}$ and the effect of the cavities is small [6]. The impedance measurements in cats and gerbils indicate that except for a slight damping of sharp middle ear cavity resonances, the resistance of the cavity is insignificant except at very low frequencies. Also the malleus motion measured in human temporal bones is not greatly affected by opening the middle ear cavity [6]. Thus placing of the external devices in the tympanic cavity will not produce any significant variations in the middle ear function.

Again, to minimize the leakage flux as most as possible, it is desirable to make the airgap length between the actuator and permanent magnet implant as small as possible. An airgap length of 500 micron seems to be a good solution. The actuator is to be placed inside the tympanic cavity by some mechanical means so that the pole faces of actuator remain firmly fixed at a distance of 500 microns when the actuator is not energized and there is no lateral movement of the actuator with respect to the permanent magnet implant. The actuator can be secured in this position using a thin support rod of very strong material (silicon carbide/titanium) and cementing the rod to the temporal bone nearby.
4.6 Force Measurements on the Stapes and the Scala Vestibuli

The Rowsoski model of external and middle ear function can be applied effectively to measure the sound pressure level at different locations inside the ear. Our particular objective is the measurement of force at the stapes footplate, i.e. $P_S$ or $P_C$. The Matlab

![Graph showing sound pressure level variation with frequency in different parts of the ear](image)

**Figure 4-3** Sound pressure measurement at different points inside the ear as obtained from Matlab implementation of the Rowsoski model of External and middle ear function

code generated from the Rowsoski model is listed in Appendix I. Figure 4-3 shows the resulting pressure distribution inside the ear as obtained from the Rowsoski model. A plain wave pressure of 1 Pa (94 dB SPL) was applied at the entrance of pinna at zero degree elevation and zero degree azimuth. It is known that 120 dB SPL is the highest sound pressure a human auditory system can sustain. But it was reported that if a human
is exposed to a continuous sound pressure higher than 90 dB SPL the person may become deaf. On this consideration a pressure of 1 Pa (94 dB SPL) is used as the highest level of sound pressure for our design purpose. This sound pressure level also reduces the required actuator size and the actuator coil current to a low value. The VLSI signal conditioner will limit any sound having a SPL higher than 94 dB.

4.7 Design Requirements for the Implants

From the Figure 4-3, it is evident that the maximum force applied at the stapes footplate is about $250 \times 10^{-6}$ Newton when the incident sound wave is 94 dB SPL and the minimum force is about $4.8 \times 10^{-9}$ Newton. Thus our design criteria is now:

1. To build a MEMS capacitive microphone suitable for use in a hearing instrument and

2. A magnetic circuit comprised of a electromagnetic microactuator and a permanent magnet implant to obtain the mentioned force variations.
Chapter 5

THE MEMS CAPACITIVE MICROPHONE

In this chapter the design and simulation of the MEMS implementation of a capacitive microphone is discussed. First, a brief discussion of the design parameters is given. Then the design process of the microphone is described. Next the simulation of the fabrication processes of the microphone is presented. Finally the electromechanical simulation of the microphone performances is described.

5.1 Lumped Acoustic and Mechanical Elements and Their Electrical Analogs

Lumped acoustic and mechanical elements provide a shorthand notation for describing the equations of motions in a fluid or a mechanical system. The force needed to compress a spring is proportional to the spring’s displacement \( f(t) = \int v(t)dt / C^M \), the force needed to compress a shock absorber or a dashpot (a mechanical resistor) is proportional to the velocity of compression, \( f(t) = v(t)R^M \), while the force needed to displace a mass is proportional to the acceleration, \( f(t) = L^M dv(t)/dt \), where \( C^M \) is the mechanical compliance of the spring, \( R^M \) is the mechanical resistance of the dashpot, and \( L^M \) is the mechanical inertness or mass. With sinusoidal stimuli these relationships can be written in terms of complex amplitudes as given in Table 5-1.

Similar relationship between pressure and volume displacement, volume velocity and volume acceleration can be used to define acoustic compliances, resistances and
<table>
<thead>
<tr>
<th>Generalized force and motion</th>
<th>Mechanical</th>
<th>Acoustical</th>
<th>Electrical</th>
</tr>
</thead>
<tbody>
<tr>
<td>Force: f(t), F(ω)</td>
<td>Pressure: p(t), P(ω)</td>
<td>Voltage: e(t), E(ω)</td>
<td></td>
</tr>
<tr>
<td>Velocity: v(t), V(ω)</td>
<td>Volume velocity: u(t), U(ω)</td>
<td>Current: I(t), I(ω)</td>
<td></td>
</tr>
</tbody>
</table>

| Impedance analogies        | Mechanical impedance: Z^M(ω) = F(ω)/V(ω) | Acoustical impedance: Z^A(ω) = P(ω)/U(ω) | Electrical impedance: Z^E(ω) = E(ω)/I(ω) |

| Ideal element values       | Compliance of a spring: F(ω) = V(ω)/jωC^M | An acoustical compliance: P(ω) = U(ω)/jωC^A | A capacitor: E(ω) = I(ω)/jωC^E |
|                            | Inertance of a mass: F(ω) = jωL^M V(ω) | An acoustic inertance: P(ω) = jωL^A U(ω) | An electrical inductance: E(ω) = jωL^E I(ω) |
| Transformation ratio of an ideal lever: T^M = F_1/F_2 = V_2/V_1 | Transformation ratio of an ideal pneumatic piston system: T^A = P_1/P_2 = U_2/U_1 | Transformation ratio of an ideal electric transformer: T^E = E_1/E_2 = I_2/I_1 |

Table 5-1 The mechanical, electrical and acoustical analogies

Inertances. Thus we can find the analogies between compliances and an electrical capacitors, between electrical, mechanical and acoustic resistances and between the
inertances and inductance. There are also simple analogies between ideal transformers in
the three systems.

Acoustic compliances are of two general types, (i) the "series" compliances associated
with elastically supported membranes or pistons that influence the flow of volume
velocity within open tubes and air spaces, and (ii) the "grounded" compliances associated
with the compression of air in enclosed spaces. In the series type of acoustic compliance,
volume velocity flows across the membrane and displacement of the membrane cause
equal displacements in the fluid on either side. In the "grounded" type of compliance,
volume velocity is expended in compressing the air within the closed space and flows
nowhere else; it is as if one side of the compliance were attached to the ground. Acoustic
resistors can be constructed from screens, slits or long narrow tubes in which the
viscosity of air dominates the force that impedes the flow of volume velocity, while
acoustic inertances can be constructed from short, wide, open tubes where much of the
sound pressure is used to accelerate the mass of air within the tube.

5.2 The Capacitive Microphone Design Considerations

The design of a high-sensitivity MEMS based capacitive microphone is presented in this
section using a polysilicon Germanium diaphragm, an airgap and a gold-coated nitride
backplate. A high mechanical sensitivity to dynamic changes in air pressure requires the
use of a low stress material to construct a thin membrane with a large surface area and
small airgap between the diaphragm and the backplate. Losses at high frequencies, due
to the compression of air in the airgap, can be minimized by providing holes or acoustical
ports in the backplate. A low damping property can be obtained using a highly perforated
backplate. The microphone is designed to be fabricated in a single wafer process using combined surface and bulk micromachining technologies.

The shape of the frequency response of the microphone is determined by the damping and resonance behavior of the microphone structure, which depends mainly on the size and stress of the diaphragm. Damping is caused by losses in the diaphragm and the viscous losses associated with the air streaming in and out of the airgap. The perforation of the backplate to create acoustical ports provides a means to control streaming losses and therefore the damping characteristics of the microphone structure.

The dynamic behavior of the microphone can be calculated using an equivalent analog electrical circuit model [8] as given in Figure 5-1. The acoustic force, $F_{\text{sound}}$, and the flow velocity $V_m$, are modeled as equivalent voltage and current sources, respectively. The air radiative resistance is defined as $R_r$, and the air mass is defined as $M_r$. The diaphragm mechanical mass is $M_m$ and its compliance is $C_m$. The airgap and back-vent losses are represented by viscous resistances $R_g$ and $R_h$, respectively. The airgap compliance is given by $C_a$.

The diaphragm compliance depends on its flexural rigidity, $D$, and tension, $T$. The flexural rigidity of the diaphragm is given [34] by:
\[ D = \frac{E t^3}{12(1 - \nu^2)} \quad (5-1) \]

where \( E \) is Young's modulus of elasticity, \( t \) is the diaphragm thickness and \( \nu \) is Poisson's ratio. The tension, \( T \), is determined by the residual stresses of the diaphragm material, which depends primarily on the deposition technique, the temperature and the crystalline structure of the deposited thin film is given by:

\[ T = \sigma, t \quad (5-2) \]

The diaphragm deflection, \( W \), can be approximated from the theory of vibration for a clamped edge rectangular plate as given by

\[ -D \nabla^4 W + T \nabla^2 W = \rho \frac{\partial^2 W}{\partial \tau^2} \quad (5-3) \]

where \( \rho \) is the mass per unit area (area density) of the diaphragm for a given diaphragm thickness.

The deflection of the square diaphragm for the first fundamental mode is:

\[ W(x, y, \tau) = A \sin \frac{\pi x}{a} \sin \frac{\pi y}{a} e^{-j2\pi \tau} \quad (5-4) \]

Where \( a \) is the diaphragm edge width.

The first resonant frequency of the diaphragm then becomes:

\[ f_{res} = \sqrt{\frac{1}{\rho} \left( \frac{D \pi^2}{a^4} + \frac{T}{2a^2} \right)} \quad (5-5) \]
The acoustical impedance of the air in contact with the vibrating diaphragm has a radiative resistance, $R_r$, and an acoustical mass, $M_r$. For a square membrane, these can be approximated by:

$$R_r = \frac{\rho_o a^4 \omega^3}{2\pi c}, \text{ and } M_r = \frac{8\rho_o a^3}{3\pi \sqrt{\pi}} \quad (5\text{-}6)$$

where $\rho_o$ is the air density, $c$ is velocity of sound and $\omega$ is the angular vibration frequency.

The diaphragm compliance is equal to the average diaphragm deflection (volume velocity) divided by the applied force. The compliance can be approximated using a Rayleigh energy method as:

$$C_m = \frac{32 a^2}{\pi^6 \left( 2\pi^2 D + a^2 T \right)} \quad (5\text{-}7)$$

The equivalent acoustical mass element, $M_m$, is derived from the kinetic energy of the square diaphragm under the uniform loading as given by:

$$M_m = \frac{\pi^4 \rho (2\pi^2 D + a^2 T)}{64 T} \quad (5\text{-}8)$$

The airgap viscosity loss, $R_x$, and its compliance, $C_a$, are given by [35]:

$$R_x = \frac{12\eta a^2}{nd^3 \pi \left( \frac{\alpha}{2} - \frac{\alpha^2}{8} - \frac{\ln \alpha}{4} - \frac{3}{8} \right)} \quad (5\text{-}9)$$

and

$$C_a = \frac{d}{\rho_o c^2 \alpha^2 a^2} \quad (5\text{-}10)$$
where \( n \) is the hole density in the backplate, \( \alpha \) is the surface fraction occupied by the holes, \( \eta \) is the air viscosity coefficient, \( d \) is the average airgap distance, and \( \rho_o \) is the air density.

The viscosity loss of the backplate holes can be given approximately as:

\[
R_h = \frac{8\eta ha^2}{mp \pi r^4}
\]  \hspace{1cm} (5-11)

where \( h \) is the backplate thickness and \( r \) is the radius of the backplate hole.

The sensitivity of the microphone is defined as the output voltage produced per unit of acoustical pressure applied to the diaphragm and is given by:

\[
S = \frac{V_o}{P} = \frac{V_b a^2}{j \omega d Z_i}
\]  \hspace{1cm} (5-12)

where \( P \) is the applied sound pressure, \( V_b \) is the bias voltage between the diaphragm and the backplate. \( Z_i \) is the total equivalent impedance of the circuit shown in Figure 1 and is given by:

\[
Z_i = R_r + j\omega(M_r + M_m) + \frac{I}{j\omega C_m} + \frac{R_f + R_h}{1 + j\omega(R_f + R_h)C_a}
\]  \hspace{1cm} (5-13)

The principal design variables are thus, the diaphragm size \( a \), the diaphragm thickness \( t \), the backplate thickness \( h \), the airgap thickness \( d \), the backplate hole radius \( r \), and the surface area fraction occupied by the holes \( \alpha \).

In [36], it is shown that the cutoff frequency, \( f_c \), of the microphone is a function of the stiffness of the diaphragm and inversely proportional to the air streaming resistance \( R_a \),
i.e. $R_e + R_h$, so that either increasing the stiffness of the diaphragm or decreasing $R_a$ will increase the cutoff frequency. Since an increase in stiffness of the diaphragm will effect the sensitivity, a good method to increase the cut-off frequency is to decrease $R_a$ by increasing the acoustic hole density $n$.

At low frequencies, the sensitivity of the microphone can be approximated as:

$$S_o = \frac{32V_ha^2}{\pi^6Td} \quad (5-14)$$

The pull-in voltage for a clamped rectangular elastic plate under tension is approximately [37] given by:

$$V_p = \frac{64}{7} \sqrt{\frac{Et^3d^3}{5(1-v^2)}} \sigma_e \epsilon^2 \left( I + \frac{2}{9} (1-v^2) \right) \frac{\sigma_e a^2}{Et^2} \quad (5-15)$$

If $t<0.01a$, $V_p$ reduces to $V_p = \frac{64}{7} \sqrt{\frac{2}{45}} \sqrt{\frac{Td}{\epsilon_o a^2}} \quad (5-16)$

Therefore, the sensitivity can be expressed in terms of the pull-in voltage as:

$$S_o = \frac{\kappa}{\epsilon_o} \left( \frac{V_h}{V_p^2} \right) d^2 \text{ where } \kappa \text{ is a constant.} \quad (5-17)$$

From the above equation, we can see that sensitivity is a function of $d^2$ and is proportional to the ratio of the bias voltage to the square of the pull-in voltage. Thus, by minimizing $V_p$ and maximizing $d$ the sensitivity can be increased.

The capacitance between the diaphragm and the backplate can be expressed as:

$$C = \frac{\epsilon_o a^2}{d} \quad (5-18)$$
In order to maximize $C$, either $a$ can be increased or $d$ can be decreased. If one parameter ($a$ or $d$) is maximized based on physical constraints like fabrication, or size requirements, etc, the other one ($a$ or $d$) can be fixed. Due to fabrication constraints associated with micromachining, efficient etching requirements for the sacrificial layer sets an upper limit on the size of $d$. Thus, $a$ becomes a variable and the specific capacitive readout circuit used determines the lower limit of $a$.

Now $V_p$ can be expressed as in terms of $C$ as:

$$V_p = \frac{64}{7} \sqrt{\frac{2}{45} \frac{Td^2}{C}}$$

(5-19)

For a desired $C$, the pull-in voltage can thus be minimized by decreasing the tensile force per unit length of the diaphragm.

The design morphology to maximize the sensitivity of a MEMS realization of a capacitive microphone consists of:

- Finding the optimum airgap thickness permitted by the fabrication process.
- Minimizing the tensile force per unit length of the diaphragm by using a low stress material and applying special techniques like Phosphorous doping, high temperature annealing, etc.
- Defining a minimum diaphragm size to satisfy the constraints associated with the hearing instrument dimensions.
- Maximizing the ratio of holes to backplate areas by increasing the acoustic hole density to increase the cutoff frequency.
Based on these criteria the optimum design parameters and outputs for the microphone are shown in Table 5-2. The calculated frequency response is given in Figure 5-2.

<table>
<thead>
<tr>
<th>Design Parameters</th>
<th>Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Diaphragm side length and width, $a$</td>
<td>2.5 mm</td>
</tr>
<tr>
<td>Diaphragm thickness, $t$</td>
<td>0.8 μm</td>
</tr>
<tr>
<td>Airgap thickness, $d$</td>
<td>3 μm</td>
</tr>
<tr>
<td>Backplate hole size, $r$</td>
<td>30 μm x 30 μm</td>
</tr>
<tr>
<td>Number of backplate holes, $n$</td>
<td>1280</td>
</tr>
<tr>
<td>Surface fraction occupied by holes, $\alpha$</td>
<td>0.18</td>
</tr>
<tr>
<td>Backplate thickness, $h$</td>
<td>1 μm</td>
</tr>
<tr>
<td>Diaphragm material</td>
<td>Poly-SiGe</td>
</tr>
<tr>
<td>Young’s modulus of Poly-SiGe, $E$</td>
<td>154.61 GPa</td>
</tr>
<tr>
<td>Poly-SiGe residual stress, $\sigma$</td>
<td>20 MPa</td>
</tr>
<tr>
<td>Backplate material</td>
<td>Silicon nitride (Si$_3$N$_4$)</td>
</tr>
<tr>
<td>Substrate</td>
<td>Si Czochralski $&lt;$100$&gt;$</td>
</tr>
<tr>
<td>Bias voltage, $V_b$</td>
<td>12V</td>
</tr>
<tr>
<td>Resulted sensitivity at 1 kHz, $S$</td>
<td>52 mV/Pa</td>
</tr>
<tr>
<td>Cut-off frequency, $f_c$</td>
<td>18 kHz</td>
</tr>
<tr>
<td>Microphone capacitance, $C$</td>
<td>18 pF</td>
</tr>
<tr>
<td>Pull-in voltage $V_p$</td>
<td>5.38 V</td>
</tr>
</tbody>
</table>

Table 5-2 Design and output parameter values of the capacitive microphone
The Matlab code used to calculate the frequency response of the microphone is given in Appendix II.

5.3 Microphone Material Characteristics

It was shown earlier that the residual stresses in the material used to make the diaphragm mainly influence the microphone's first resonant frequency, cutoff frequency and sensitivity. The residual stresses in the thin films are the sum of thermal and intrinsic stresses. Thermal stress is caused by a difference in the expansion coefficient of the substrate and of the thin film. Intrinsic stress has two components: the first originates from the volume contraction associated with crystallization and is tensile; and the second one is compressive and is due to the existence of a preferred growth orientation, disorder
at the grain boundary, effects related to different deposition rates or the incorporation of impurity atoms. It is known that the dominant component of stress is the intrinsic one [9]. The stress gradient along the z-axis direction of a thin film also contributes to the residual stress [38].

There are some methods available to control the residual stress. One of them is to anneal the thin film at a high temperature between 900°C to 1100°C in nitrogen after ion implantation by Boron, Phosphorous or Arsenic.

Different types of materials have been reported for the fabrication of the diaphragm, viz., silicon nitride, annealed polysilicon after ion implantation, polyamide, monocrystalline silicon, etc. The residual stresses of the above films range from 110 MPa (PECVD nitride) to 20 MPa (high temperature annealed LPCVD polysilicon ion implanted with Phosphorous). The need for high temperature annealing of a diaphragm made of polysilicon prevents the micromachined structure to be built on the same wafer where the electronic circuitry is already fabricated due to thermal damage. Since the high tensile stress of PECVD nitride limits the microphone sensitivity, a newly developed alloy - polycrystalline Silicon Germanium (Poly-SiGe) appears to be an excellent alternative for the diaphragm construction.

It is found that the poly-SiGe system offers several advantages over the conventional polysilicon-and-oxide system [39]. Poly-SiGe can be conformally deposited and doped at much lower temperatures (less than 500°C) than polysilicon. This eases integration with CMOS devices and allows for surface passivation to reduce stiction effects. Poly-SiGe with a high Ge content can be etched in solutions of NH₄OH and H₂O₂ that will not
damage the on-chip CMOS devices. Poly-SiGe structural and sacrificial layers can be deposited in the same chamber by changing the percentage of Ge content.

Poly-SiGe has a lower strain and strain gradient than polysilicon. High temperature annealing is, therefore not necessary in all cases. The Poly Ge sacrificial layer is highly conformal and thus allows the structure to be an accurate reproduction of the mold wafer. Instead of a Poly Ge sacrificial layer, a conventional SiO₂ sacrificial layer can as well be used. It is reported [9] that Poly-SiGe shows nearly flat surface planes and this is a clear indication of the uniformity of stress (low stress gradient) along the direction of growth. The stress induced decrease in the airgap thickness associated with a microphone having a polyamide diaphragm (resulting in a decreased sensitivity) will not occur with a microphone having a Poly-SiGe diaphragm [10].

1 Microphone Fabrication

The process starts with a <100> oriented n-type Czochralski silicon wafer. A 1.0 micron thick silicon nitride, Si₃N₄, layer is deposited by LPCVD process. Over the nitride layer, a 30 nm thick layer of titanium is sputtered, which is followed by the evaporation-
deposition of a 30 nm thick layer of gold. Next, a 3 microns thick layer of SiO₂ is deposited by LPCVD process. The oxide layer is patterned and then etched using buffered hydrofluoric acid (Figure 5-3).

This is next followed by the deposition of a 300 nm thick layer of low-stress PECVD SiN layer. This nitride layer will act as an isolator between the Poly-SiGe diaphragm and the backplate electrode. The PECVD nitride layer is then patterned and etched in hot phosphoric acid using a .5 μm thick layer of low temperature oxide (Figure 5-4).

In the next step a 40 nanometer thick layer of polysilicon is deposited by LPCVD process. This polysilicon layer will act as the nucleation layer for the subsequent
deposition of Poly-SiGe. The Poly-SiGe (30% Ge) is deposited by LPCVD process using SiH₄ and GeH₄ gases (Figure 5-5).

The Poly-SiGe and the thin polysilicon layers are then ion-implanted with phosphorous and annealed at 925°C to reduce the residual stress. The Poly-SiGe and the thin polysilicon layers are then patterned and etched using reactive ion etching with SF₆/CH₄ plasma (Figure 5-6).

![Figure 5-6 Poly-SiGe and polysilicon layers are RIE etched](image)

After that a thin layer of low-temperature oxide (LTO) is deposited, patterned and etched in Buffered Hydrofluoric acid (BOE) to define the contact hole for the backplate (gold electrode). The nitride over the contact area is then etched by hot phosphoric acid. Next, the backside of the wafer is patterned using a 0.5 μm thick layer of LTO. The silicon wafer is then RIE etched using SF₆/O₂ plasma (Figure 5-7).

![Figure 5-7 Pattern and etching of PECVD nitride layer and the substrate is patterned and RIE etched](image)
Next, the silicon nitride backplate is patterned using LTO and the silicon nitride layer and titanium layer are plasma etched using CF$_4$/O$_2$. The gold layer is then RIE etched using Cl$_2$ (Figure 5-8).

![Figure 5-8 The silicon nitride backplate, titanium and gold layers are patterned and etched dry](image)

The protective LTO is then stripped. Chromium and gold are next evaporated to form contact pads with a thickness of 50 and 400 nm, respectively. The contact pads are then patterned using photoresist and wet etched (Figure 5-9).

![Figure 5-9 Chromium and gold contact pads are formed](image)

Gold

Chromium
Finally, the sacrificial oxide layer is etched by BOE to release the diaphragm and then the wafer is freeze-dried using cyclohexane at atmospheric pressure under a continuous flow of nitrogen (Figure 5-10).

![Figure 5-10 The sacrificial oxide layer is etched BOE to release the diaphragm](image)

### 5.5 Operation and Readout Circuitry

The microphone requires a bias voltage $V_{bias}$ of 12 volts to obtain a linear operating range. This voltage can be supplied using a Dickson type DC-DC voltage converter with a relatively low supply voltage of about 2 volts and having a 10 stage voltage multiplier circuit (Figure 5-11)[10].

![Figure 5-11 Dickson type DC-DC voltage converter](image)
As the capacitance variation of the microphone structure is relatively low, a readout circuit with high gain and low input noise properties is required. A simple source follower can be used as a buffer amplifier to convert the current flow due to capacitance change into a usable ac voltage [40] as shown in Figure 5-12.

The input bias voltage of the amplifier, $V_{\text{amp}}$, can be set through an isolated minimum geometry diode. Thus the voltage at the backplate of the microphone can be given by:

$$V_{\text{mic}} = V_{\text{amp}} + V_a$$  \hspace{1cm} (5-20)

where $V_a$ is the ac voltage generated by the acoustic signal. The current, $i_{\text{mic}}$, flowing through the combination of the microphone capacitance and the capacitance associated

\begin{figure}[h]
\centering
\includegraphics[width=0.5\textwidth]{readout_circuit.pdf}
\caption{Readout circuit for the capacitive microphone}
\end{figure}

with the output pad to the backplate is thereby injected into the gate of the source follower and can be expressed by:

$$i_{\text{mic}} = \frac{d}{dt} (Q) = \frac{d}{dt} \left[ (C_{\text{mic}} + \delta C_{\text{mic}} + C_{\text{pad}})(V_{\text{bias}} - V_{\text{amp}} - V_a) \right]$$  \hspace{1cm} (5-21)
where $C_{mic}$ and $\delta C_{mic}$ are the average and time varying components, respectively, of the capacitance between the microphone electrodes. If the input capacitance of the buffer amplifier is $C_{inp}$, $v_a$ can be expressed as:

$$v_a = \frac{\delta C_{mic}}{C_{mic} + C_{pad} + C_{inp}} (v_{bias} - V_{ump})$$  \hspace{1cm} (5-22)

5.6 Simulation

5.6.1 Simulation of fabrication process

After the microphone design parameters are fixed, the fabrication processes of the microphone structure was simulated using IntelliSuite MEMS software. The software has the capability to simulate the actual fabrication processes to give a visualization of the fabrication processes step by step. After the final process step is executed, the final 3D structure of the MEMS device can be visualized in the visualization window. In the package a material is characterized by specifying its physical properties, viz, density, Young modulus, Poisson ratio, thermal conductivity, coefficient of thermal expansion, dielectric constant, etc. Up to 24 physical properties can be entered. Some materials we have used to fabricate the microphone were not available in the material database supplied with the software package. The materials Poly-SiGe, titanium and chromium were created entering the physical properties obtained from different authentic resources. As the thin film properties vary considerably depending on the deposition techniques and process parameters, the appropriate processes were then created for the deposition of LPCVD Poly-SiGe, sputtering of titanium and electron beam evaporation of chromium. After defining all the materials and processes necessary for the fabrication of the
microphone, appropriate masks are then designed using IntelliSuite mask editor. A total of nine masks were used to complete the fabrication process. In patterning of the thin films, the industry standard high precision Karl Suss Contact Aligner was simulated. For photoresists, another industry standard Shipley photoresist series S-1800 was used. S-1800 series photoresists are wideband UV lithography compatible and are ranging from 324 to 436 nanometers in wavelengths. After stripping of the photoresists after developing, the structure was RCA cleaned for subsequent processing. After the completion of the simulation of the fabrication processes, a 3D model of the microphone structure was generated by the software. The 3D model of the microphone is shown in Figure 5-13. The complete process sequence is given in Appendix III. No process incompatibility was reported.

Figure 5-13 3D model of the capacitive microphone in the IntelliSuite design environment after simulating the fabrication processes
5.6.2 Performance Simulation of the microphone

After successful completion of simulation of the fabrication processes, the 3D model of the microphone was exported to the electromechanical analysis module of IntelliSuite to carry out the electromechanical analysis.

5.6.2.1 Simulation Setup.

5.6.2.1.1. The Electromechanical simulation method used in IntelliSuite.

A solution method called “Relaxation” is used to solve coupled electromechanical problems. The method implemented in IntelliSuite uses an electrostatic boundary element solver to extract capacitance and electrostatic forces, and a finite element solver to extract mechanical deformations. Each domain solver is called in turn, updating the values of its associated variables. After a check for convergence, this sequence is repeated. The convergence criteria can be set by the user to a desired value depending on the precision of work required [41].

IntelliSuite uses 20-node brick elements because of the high quality and accuracy of these elements. After extensive numerical testing, linear elements (such as 8-node brick elements) were found to be inefficient for MEMS simulations. These linear elements usually require a prohibitively fine discretization (especially through the thickness) which leads to large numerical models.

Tests on the tetrahedral family of elements also showed that a larger number of nodes and elements are required to achieve the same accuracy as a mesh consisting of 20-node brick elements. In MEMS applications especially, the tetrahedral elements in a mesh comprised solely of tetrahedrons can become highly distorted leading to numerical inaccuracies.
5.6.2.1.2. Simulation constraints

A solution method that engages an electrostatic boundary element solver and a finite element solver is very much time consuming. The accuracy of the result depends largely on the number of nodes used to solve the problem, i.e. the finer the mesh size, the more accurate will be the result and at some specific size of the mesh the results doesn’t vary. This is known as mesh convergence criteria. Also the mesh over the whole structure needs not to be refined. A mesh refinement is necessary only at the particular part of the structure that has some mechanical movement, vibration or undergoes mechanical deformation. In IntelliSuite besides mechanical mesh refinement, there is also provisions for electrical mesh refinement. Instead of refining the volume mesh, Elect._mesh can be used to refine only the electrostatic surface mesh on chosen Exposed Faces. The advantage of this method is that the surface mesh used in the electrostatic analysis is separated from the mechanical volume mesh while assuring full compatibility between the two. The EFM method results in smaller computational models while improving the numerical accuracy in MEMS simulation.

To carry out the simulation, the 3D model of the microphone was modified. First, to keep the problem size small, the titanium adhesion layer and the gold contact layer were removed in that sense that in IntelliSuite while a voltage is applied to an entity, IntelliSuite interprets it as a conductor. This saves a considerable amount of computation time. Second the PECVD nitride layer was removed to visualize the deformation and the pressure and the charge distribution after the simulation process was finished. Third, the mechanical volume mesh in the Poly-SiGe diaphragm was refined to obtain the desired
accuracy and also to keep the numerical problem size small. The 3D model of the microphone with removed layers is shown in figure 5-14.

The space within the silicon nitride backplate and the Poly-SiGe diaphragm is interpreted by IntelliSuite as an airgap and the dielectric constant of air is used as default value of insulation between the diaphragm and the backplate.

5.6.2.1.3 Applied electrical and mechanical loads

The physical properties of materials used for this electromechanical simulation are, the dielectric constant of the materials, the resistivity, the Young’s modulus, Poisson ratio, density, coefficient of thermal expansion, and thermal conductivity. After checking of the appropriate value of the materials, necessary loads are applied as below:
1. Voltage:
   Poly-SiGe diaphragm: 0 V DC
   Nitride back plate: 12 V DC

2. Pressure:
   Poly-SiGe diaphragm: 1 Pascal, a standard pressure applied to test MEMS microphone [34].

5.6.2.1.4. Applying appropriate boundary conditions

All the four lateral faces of the diaphragm were fixed to emulate a clamped membrane. The top and the bottom faces of the nitride backplate were fixed. Also the 4 outer and 4 interior walls of the silicon wafer were fixed. As the wafer was through etched using RIE, the etch profile was almost vertical. Thus there was no need for node adjustment as that would be required if the etching was carried out using KOH.

5.6.2.1.5 Mesh generation

After all the boundary conditions are set, a coarse mesh was created using default mode. In IntelliSuite current version, the recommended mesh size is 2-10% of the XY dimension of the device. The number of nodes is limited to a maximum of 20,000. However, if the workstation has enough memory (96 MB or more) and available hard disk space (1 gigabyte or more), the system may be able to handle larger problems (more than 20,000 nodes). As the mesh is required to be conformal in all the three X-Y, Y-Z and X-Z dimensions, the smallest mesh size for the backplate is automatically set at 30 microns. As both the X and Y dimensions of the diaphragm is 2500 microns a mesh size of 100 microns was selected for the diaphragm that is 4% of the X or Y dimension to stay closer to the lower limit of the recommended value. A total of 26,000 nodes were created
after the mesh refinement. A memory of 30 Megabytes was set for the prememory and 10 megabytes for the prebuffer after trial and error.

The analysis start option was set for an undeformed option. Next the analysis option was set at large displacement to carry out a nonlinear mechanical analysis accounting residual stresses to simulate a practical problem.

As it was stated earlier, the electrostatic boundary element solver is coupled with finite element solver to continue looping through solution algorithms until a desired accuracy of convergence is met, in the next step the convergence criteria for displacement was set at 0.005 micron. An iteration number of 20 was set arbitrarily since the number of iteration or the convergence criteria which one is met first, the software will stop iterating. After completion of all of these setup procedures, the relaxation process was activated.

5.6.2 Simulation Results

After the electromechanical analysis was completed, an examination of the FEA-JOB message file shows that yet, the iteration number was set at 20. at the 14th iteration, the convergence criteria of 0.005 micron was satisfied and the program stops executing. No warning or error messages were generated during the simulation process for wrong user input or during the analysis. The FEA_JOB message file is shown in Appendix IV.

The capacitance between the diaphragm and the back plate was simulated as 16.09 pico farad. The result is very much consistent with the calculated value that is 18 picofarad. The plot of Capacitance Vs. Iterations is shown in Figure 5-15.
The Z-axis displacement of the diaphragm due to coupled electrostatic force and the applied mechanical force is shown in Figure 5-16. The displacement is completely symmetrical with a maximum at the center node with a value of \(-1.51\) micron. Since the airgap thickness is 3 micron, there is no possibility of contact between the diaphragm and the back plate at the applied sound pressure and bias voltages.
Figure 5-16 Diaphragm deflection due to applied mechanical force of 1 Pascal and the electrostatic attraction force after biasing. The bias voltage is 12 V DC.

The convergence of the simulation process can be observed from Z-axis-displacement vs. iterations number that is shown in Figure 5-17. It is clear that the results obtained after iteration No. 12, 13 and 14 are very close to each other and between iteration 13 and 14 the difference of maximum deflection of the diaphragm is less than 0.005 microns. Thus the accuracy criteria were satisfied and the simulation process terminates.

The pressure distribution on the top face and the bottom face of the diaphragm are given in Figure 5-18 and Figure 5-19. A 3D view of the deformed diaphragm is shown in Figure 5-20.
Figure 5-17 Plot of Z axis displacement of the microphone diaphragm vs. iterations
Figure 5-18 Pressure distribution on the top surface of the diaphragm
Figure 5-19  Pressure distribution on the bottom surface of the diaphragm
Figure 5-20 3D view of the microphone showing deformation in the Z-axis direction
5.7 Summary of the Microphone Properties and Specifications

A capacitor microphone has been designed using a polysilicon-germanium diaphragm with an airgap and a perforated silicon nitride backplate. The losses at high frequencies, due to compression of the air in the airgap, have been minimized by providing acoustical ports in the backplate. The microphone is constructed using a combination of surface and bulk micromachining techniques in a single wafer process. The microphone diaphragm has a thickness of 0.8μm, an area of 2.5 x 2.5 mm², an airgap of thickness 3.0 μm and the backplate that accommodates 1280 acoustical ports is 1.0 μm thick. Each acoustical port has a dimension of 30 x 30 μm². A buffer amplifier made of a simple source follower with high gain and low input noise properties can be used to convert the current flow due to capacitance change into a usable ac voltage. A 12.0 volt DC bias voltage from a Dickson type DC-DC voltage converter is provided between the diaphragm and the backplate. A sensitivity of about 52.0 mV/Pa is expected for the microphone, with a high frequency response extending to 18 kHz.
Chapter 6

MAGNETIC STRUCTURES DESIGN AND IMPLEMENTATION

The design and fabrication of the MEMS realization of an electromagnetic microactuator and the design of a permanent micro magnet is discussed in this chapter. These two magnetic structures form a magnetic circuit to be used in the proposed acousto-magnetic transduction system. At first the required magnetic formulations are given, then the permanent magnet design considerations are discussed. Next the magnetic circuit design procedure is described. The design of the electromagnetic microactuator is given next and finally the fabrication of the electromagnetic microactuator is presented.

6.1 Magnetic Analysis

A small rectangular loop of wire (of area $A$) carrying an electrical current $i$ when is placed in a uniform magnetic field $B$, it will experience a torque $T$ that tends to rotate the plane of the current loop until it is perpendicular to the direction of $B$. This torque is calculated from:

$$T = iAB \sin \theta$$  \hspace{1cm} (6-1)

The magnetic dipole moment $m$ linked to a microscopic current loop of area $A$ and carrying a current $i$ is defined as:

$$m = iA$$ \hspace{1cm} (6-2)

The torque that tends to rotate a magnetic dipole moment into alignment with an applied magnetic field thus can be expressed as:
\[ T = mB \sin \theta \quad (6-3) \]

The work done \( E \) in rotating a magnetic dipole moment into alignment with an applied magnetic field is given by:

\[ E = \int T \, d\theta = mB \int \sin \theta \, d\theta = -mB \cos \theta \quad (6-4) \]

The magnetization \( M \) is defined as the magnetic dipole moment per unit (elemental) volume and is given by:

\[ M = \lim_{\Delta V \to 0} \frac{\sum m}{\Delta V} \quad (6-5) \]

In an elemental volume in which there is only a \( z \)-directed magnetization \( M_z \), the sum of the moments is:

\[ \sum m_z = M_z \delta x \delta y \delta z \quad (6-6) \]

From the definition of magnetic dipole moment:

\[ \sum m_z = i \delta x \delta y \quad (6-7) \]

Thus:

\[ i = M_z \delta_z \quad (6-8) \]

In the adjacent elemental volume in which there is also only a \( z \)-directed magnetization \( M'_z \) with its equivalent current \( i' \), the change in current is found to be:

\[ i' = i + \frac{\partial i}{\partial x} \delta x = i + \frac{\partial M_z}{\partial x} \delta x \delta z \]

Thus:

\[ i - i' = -\frac{\partial M_z}{\partial x} \delta x \delta z \quad (6-9) \]
Thus an equivalent current \( i - i' \), in other words a current density \( J_y = -\frac{\partial M}{\partial x} \) flows through the boundary of two elemental current loops [42].

The magnetic vector potential [43] \( A \) at a point \((x', y', z')\) due to an infinitesimal magnetic dipole moment \( m \) at point \((x', y', z')\) at a distance \( R \) away is given by:

\[
A = \frac{\mu_0}{4\pi} \nabla \times \frac{m}{R}
\]  

(6-10)

where \( \mu_0 \) is the permeability of free space. The contribution to the vector potential from the magnetic polarization of a material body is given by:

\[
A(x, y, z) = \frac{\mu_0}{4\pi} \int_{V'} \nabla' \times \frac{M(x', y', z')}{R} dV' + \frac{\mu_0}{4\pi} \oint_S \frac{M \times n}{R} dS'
\]  

(6-11)

where \( n \) is the unit outward normal to the surface \( S \) of the body, \( M \) is the dipole moment per unit volume and \( \nabla' \) indicates the differentiation with respect to the source variables \( x', y', z' \). The volume density of the polarization current \( J_m \) is given by:

\[
J_m = \nabla' \times M
\]  

(6-12)

and the surface polarization current \( J_{ms} \) is expressed as:

\[
J_{ms} = M \times n
\]  

(6-13)

If the magnetization of the body is uniform, \( M \) is constant throughout, thus the curl of \( M \) is zero and the equivalent volume polarization current is zero. The surface polarization current vanishes only when \( M \) is perpendicular to the surface so that \( M \times n \) is zero.

From Ampère's Law, the magnetic field \( B \) rotating around a conductor carrying a real electrical current density \( J \) is expressed as:

\[
\mu_0 J = \nabla \times B
\]  

(6-14)
This may equally well refer to a magnet's equivalent current density $J_m$, or to a system in which there exists both electrical currents and magnetic materials, in which case:

$$\mu_0 (J + J_m) = \nabla \times B$$  \hspace{1cm} (6-15)

Substituting for $J_m$ from equation (6-12), we get $\mu_0 (J + \nabla \times M) = \nabla \times B$, or $\mu_0 J = \nabla \times (B - \mu_0 M)$ and finally:

$$J = \nabla \times \left( \frac{B}{\mu_0} - \frac{M}{\mu_0} \right)$$  \hspace{1cm} (6-16)

Substituting $\left( \frac{B}{\mu_0} - \frac{M}{\mu_0} \right)$ by $H$, the magnetic field intensity vector, we obtain:

$$J = \nabla \times H$$  \hspace{1cm} (6-17)

Excluding the ferromagnetic material, $M$ is directly proportional to the applied external field $B$ and thus $M$ is directly proportional to $H$ also. Thus:

$$M = \chi_m H$$  \hspace{1cm} (6-18)

where $\chi_m$ is the magnetic susceptibility. Thus we get:

$$B = \mu_0 (1 + \chi_m) H = \mu H$$  \hspace{1cm} (6-19)

where:

$$\mu = \mu_0 (1 + \chi_m)$$  \hspace{1cm} (6-20)

is the magnetic permeability.
6.2 The permanent Magnet Considerations

![Diagram showing elementary current loops](image)

Figure 6-1 Cross-section of a permanent magnet showing elementary current loops

6.2.1 Permanent magnet is a solenoid

In the interior of a bar permanent magnet that has a uniform magnetization (magnetic dipole moment per unit volume) equal to $M_{sat}$, the volume density of polarization current is given by:

$$J_m = \nabla \times M_{sat}$$  \hspace{1cm} (6-21)

Since $M_{sat}$ is constant, the circulating infinitesimal current loops all carry the same current and hence all the interior currents effectively cancel out so that $J_m = 0$ (Figure 6-1(a)). As along the outer boundary there is no cancellation of the currents of
each adjacent small loops, all the individual current loops combine to produce a net
surface polarization current flowing along the surface of the bar magnet (Figure 6-1(b)).

The value of this current is equal to:

\[ J_{sat} = M_{sat} \times n = M_{sat} a_z \times a_r = M_{sat} a_\phi \]  \hspace{1cm} (6-22)

where \( a_r \) and \( a_\phi \) are the unit vectors in cylindrical co-ordinates and this current flows
circumferentially around the bar magnet.

Thus the field from the permanent bar magnet will be same as that from an equivalent
solenoid having an effective surface current of \( M_{sat} \) amperes per meter \([44]\). A closely
wound solenoid of \( N \) turns per meter carrying a current \( I \) would be equivalent to the
permanent bar magnet if:

\[ NI = M_{sat} \]  \hspace{1cm} (6-23)

6.2.2 Magnetocrystalline Anisotropy

To rotate a magnetic dipole moment \( m \), into alignment with the magnetic flux density \( B \).
the work done \( E \) is expressed as:

\[ E = -mB \cos \theta \]  \hspace{1cm} (6-24)

and substituting for \( B \):

\[ E = -\mu_0 mH \cos \theta - \mu_0 mH \sin \theta \]  \hspace{1cm} (6-25)

The first term of the energy \( E \) expression, represents the work done \( E_f \) by a field \( H \) upon
an individual moment \( m \), while the second term represents the work done \( E_k \), by the net
distribution of moments (i.e. the magnetization) upon an individual moment \( m \).
From above equation the magnetocrystalline anisotropy energy $E_k$, the minimization of which expresses any preferred direction for an individual moment $m$ within a crystal structure having a net magnetization $M$ can be expressed as:

$$E_k = 2\mu_0 M M \sin^2 \frac{\theta}{2}$$  \hspace{1cm} (6-26)

Conversely, the maximization of this $E_k$ function indicates when the orientation of an individual moment $m$ is unstable within a crystal structure. The function is maximized at $180^\circ$, being the unstable condition when $m$ is directly opposed to the net $M$ (the stable state is when $m$ is aligned with $M$, at $0^\circ$ or $360^\circ$).

Considering iron that has a crystal structure which places individual moments $m$ in a "body-centered-cubic" lattice, the stable states for an individual moment $m$ are whenever it is aligned with any of the planes that occur every $90^\circ$, i.e. the unstable condition first occurs at $45^\circ$ rather than $180^\circ$. Applying this condition and defining a crystalloographic constant for the material $K_i = 8\mu_0 M M$, we get:

$$E_k = \frac{K_i}{4} \sin^2 2\theta$$  \hspace{1cm} (6-27A)

The work done $E_f$ by an externally applied field $H$ that try to rotate $M_{sat}$ (saturation magnetization) away from one of the preferred crystal planes and into alignment with $H$ can be expressed as:

$$E_f = -\mu_0 M_{sat} H \cos(\theta_0 - \theta)$$  \hspace{1cm} (6-27B)

where $\theta_0$ is the angle of applied field $H$.

Thus:

$$E = \frac{K_i}{4} \sin^2 2\theta - \mu_0 M_{sat} H \cos(\theta_0 - \theta)$$  \hspace{1cm} (6-28)
Differentiating $E$ to find the minimum total energy, then differentiating a second time and setting the result $= 0$ we get the reversal of the rate of change of energy:

$$\frac{\partial^2 E}{\partial \theta^2} = 2K_1 \cos 4\theta - \mu_0 M_{\text{sat}} H \cos \theta = 0 \quad (6-29)$$

The solution to this equation yields a unique value for a critical field that is just strong enough to flip over $M_{\text{sat}}$ into the reverse direction, is the intrinsic coercivity $H_{ci}$ of the magnet [45]. The higher this $H_{ci}$ the less the magnet will be at the risk of reversal of polarization due to an externally applied field.

For the implant we need a high intrinsic coercivity permanent magnet for this reason.

### 6.2.3 Stress Anisotropy

A magnetic material that is mechanically strained when the direction or magnitude of its magnetization changes is said to be magnetostrictive. Conversely, a change in the mechanical strain of a magnetostrictive material can affect its magnetization [46]. Through magnetostriction the stress in a magnetic material generates a magnetic anisotropy known as stress anisotropy. The general relationship between stress and the magnetic-stress anisotropy energy density in an isotropic magnetic material is given by:

$$W_{\text{stress}} = K_{\text{stress}} \sin^2 \theta = \frac{3}{2} \lambda \sigma \sin^2 \theta \quad (6-30)$$

with magnetostriction coefficient $\lambda$ and stress $\sigma$. Typically the magnetostriction of NiFe alloys run between $\lambda = \pm 30 \times 10^{-6} \lambda$. This means that for an anisotropic stress level as high as 1 GPa, $K_{\text{stress}}$ can be as large as $4.5 \times 10^4 \text{ J/m}^3$. 

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6.2.4 The Demagnetization Curve.

The demagnetization curve using the $B=\mu_0(H+M)$ relationship for a permanent magnet converts the intrinsic $M$ vs. $H$ magnetization characteristic of the magnet into the "normal" $B$ vs. $H$ magnetization characteristic that is useful for practical design [45]. The typical demagnetization curve of a permanent magnet is shown in Figure 6-2.

Remanence is the intercept of the $B$ vs. $H$ curve on the positive $B$ axis. For the ideal material, its value $B_r=\mu_0M_{sat}$ but $B_r$ is always the value of flux density for the condition when a magnet develops no magnetizing force ($H=0$).

Coercivity is the intercept of the $B$ vs. $H$ curve on the negative $H$ axis. Its value $-H_c$ is the

![Figure 6-2 The demagnetization curve of a permanent magnet](image)

magnetizing force required to reduce the magnet's flux density $B$ to zero, which on this ideal curve is $-H_c=M_{sat}$. A comparison of the "normal" and intrinsic characteristics of a permanent magnet shows that the values of $-H_c$ and $-H_{ci}$ are not the same, i.e. the
magnetizing force required to make \( B = 0 \) may be less than that required to reverse the direction of the material's magnetization.

Maximum energy product \((BH)_{\text{max}}\) is the point on the second quadrant of the \( B \) vs. \( H \) curve at which the product of \( B \) and \( H \) for the magnet are maximized. On this ideal curve, it is located exactly halfway down the second quadrant line, with a value of:

\[
(BH)_{\text{max}} = \mu_0 (\frac{1}{2}M_{\text{sat}})^2
\]

From the manufacturer's data of \((BH)_{\text{max}}\) it is thus possible to calculate the magnetic dipole moment per unit volume when a magnetic material is permanently magnetized.

### 6.2.5 Neodymium-Iron-Boron Magnet

Neodymium-Iron-Boron material was developed in the beginning of 80's. NdFeB magnets are the most powerful ones commercially available today. They show a combination of very high remanence, coercivity and maximum energy product, superior to all other magnet materials [11].

NdFeB magnets are manufactured by pressing and sintering of metallic powder, and are magnetically anisotropic. After sintering and heat treatment they are normally ground to final dimensions and surface coated with Ni, Zn or Sn for corrosion protection. NdFeB magnets are available in shapes of blocks, plates, rings, rods and segments. A relatively large amount of standard dimensions are accessible, but magnets with special dimensions can be manufactured by slicing. Tolerances in surface coated magnets are normally \( \pm 0.1 \) mm, but even tighter tolerances are possible. Typical demagnetization curves of Neodymium-Iron-Boron type permanent magnet is shown in Figure 6-3.
Three types of NdFeB magnets are available:

Sintered: There are several different routes by which Neodymium-Iron-Boron magnets are manufactured. The milled powder being formed into a fully dense anisotropic magnet by compaction in an orienting field and then sintering. This produces a nucleation-type, in which the grain boundaries are composed of deviations from the primary Nd$_2$Fe$_{14}$B composition, providing pinning of the domain walls.

Rapidly quenched: A completely different process route involves rapid quenching of the molten Neodymium-Iron-Boron alloy, using a "melt spinning" technique to produce a ribbon that is then milled to powder. While the crushed ribbon yields relatively large platelet-shaped powder particles, rapid quenching provides them with the extremely fine microstructure that conforms to the single domain model. Neodymium-Iron-Boron
magnets that are made from rapidly quenched powder base their permanent magnetism on magnetocrystalline anisotropy, and require a much greater applied field to initially align the grains' magnetizations and magnetize this material to its saturation level. Since it is not practical to mill this powder to single domain size, rapidly quenched powder is inherently isotropic. However, it can be consolidated into a fully dense anisotropic magnet by the plastic deformation that occurs in hot pressing. The natural protection afforded the grain boundaries by the fine microstructure also makes this powder very stable against oxidation, so it is easy to blend and form into any type of isotropic bonded magnet.

**HDDR:** The HDDR process (hydrogenation, disproportionation, desorption and recombination), gives Neodymium-Iron-Boron powder an ultrafine structure with grains about the size of a single domain, and the powder can be milled into particles of around this size. HDDR prepared Neodymium-Iron-Boron powder is therefore inherently anisotropic, and magnets made from it base their permanent magnetism on magnetocrystalline anisotropy. HDDR powder can be hot pressed into a fully dense anisotropic magnet, or it can be blended and molded into an anisotropic bonded magnet.

Neodymium-Iron-Boron magnets demagnetization curves shows well-defined "knees" at which $M_{sat}$ reverses as an applied field of $-H_{ci}$ is approached.
6.2.6 Magnetic properties Neodymium-Iron-Boron magnet

Magnetic properties of some commercial grade of Neodymium-Iron-Boron magnet are listed in table 6-1.

<table>
<thead>
<tr>
<th>Grade</th>
<th>Remanence (Br)</th>
<th>Coercivity (bHc)</th>
<th>Intrinsic Coercivity (iHc)</th>
<th>Maximum Energy Product (BH)(\text{max})</th>
<th>Maximum Working Temperature °C</th>
</tr>
</thead>
<tbody>
<tr>
<td>NA27</td>
<td>1.02 - 1.10</td>
<td>764 - 836</td>
<td>Min 955</td>
<td>200 - 224</td>
<td>80</td>
</tr>
<tr>
<td>NA30</td>
<td>1.08 - 1.15</td>
<td>796 - 859</td>
<td>Min 955</td>
<td>224 - 248</td>
<td>80</td>
</tr>
<tr>
<td>NA33</td>
<td>1.13 - 1.17</td>
<td>843 - 883</td>
<td>Min 955</td>
<td>248 - 264</td>
<td>80</td>
</tr>
<tr>
<td>NA35</td>
<td>1.17 - 1.22</td>
<td>875 - 915</td>
<td>Min 955</td>
<td>264 - 280</td>
<td>80</td>
</tr>
<tr>
<td>NA37</td>
<td>1.20 - 1.28</td>
<td>899 - 955</td>
<td>Min 955</td>
<td>280 - 304</td>
<td>80</td>
</tr>
<tr>
<td>NA27H</td>
<td>1.02 - 1.10</td>
<td>764 - 836</td>
<td>Min 1353</td>
<td>200 - 224</td>
<td>120</td>
</tr>
<tr>
<td>NA30H</td>
<td>1.08 - 1.15</td>
<td>796 - 859</td>
<td>Min 1353</td>
<td>224 - 248</td>
<td>120</td>
</tr>
<tr>
<td>NA32H</td>
<td>1.12 - 1.17</td>
<td>812 - 883</td>
<td>Min 1353</td>
<td>248 - 264</td>
<td>120</td>
</tr>
<tr>
<td>NA27SH</td>
<td>1.02 - 1.10</td>
<td>764 - 836</td>
<td>Min 1591</td>
<td>200 - 224</td>
<td>150</td>
</tr>
<tr>
<td>NA30SH</td>
<td>1.08 - 1.15</td>
<td>796 - 859</td>
<td>Min 1591</td>
<td>224 - 248</td>
<td>150</td>
</tr>
<tr>
<td>NA25UH</td>
<td>0.97 - 1.05</td>
<td>748 - 812</td>
<td>Min 1910</td>
<td>184 - 208</td>
<td>160</td>
</tr>
</tbody>
</table>

Table 6-1 Magnetic properties of commercial grade Neodymium-Iron-Boron permanent magnet
6.2.7 Physical Properties of NdFeB Magnets

The major physical properties of Neodymium-Iron-Boron type permanent magnet are given in Table 6-2.

<table>
<thead>
<tr>
<th>Property Name</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Temperature Coefficient of Br</td>
<td>-0.11% / °C</td>
</tr>
<tr>
<td>Density</td>
<td>7.4 g/cm³</td>
</tr>
<tr>
<td>Vickers Hardness</td>
<td>600 Hv</td>
</tr>
<tr>
<td>Tensile Strength</td>
<td>8.0 Kg / mm²</td>
</tr>
<tr>
<td>Specific Heat</td>
<td>0.12 KCal / (Kg. °C)</td>
</tr>
<tr>
<td>Young’s Modules</td>
<td>1.6 x 10¹¹ N / m²</td>
</tr>
<tr>
<td>Poisson’s Ratio</td>
<td>0.24</td>
</tr>
<tr>
<td>Curie Temperature</td>
<td>310 - 340 °C</td>
</tr>
<tr>
<td>Temperature Coefficient of iHc</td>
<td>-0.60 % / °C</td>
</tr>
<tr>
<td>Electrical Resistivity</td>
<td>144 m W / cm</td>
</tr>
<tr>
<td>Flexural Strength</td>
<td>25 Kg / mm</td>
</tr>
<tr>
<td>Coefficient of Thermal Expansion</td>
<td>4 x 10⁻⁶ / °C</td>
</tr>
<tr>
<td>Thermal Conductivity</td>
<td>7.7 KCal / (m.h.°C)</td>
</tr>
<tr>
<td>Rigidity</td>
<td>0.64 N / m²</td>
</tr>
<tr>
<td>Compressibility</td>
<td>9.8 x 10⁻¹² m² / N</td>
</tr>
</tbody>
</table>

Table 6-2 Physical properties of Neodymium-Iron-Boron permanent magnet
6.3 The Electromagnetic Microactuator Design

6.3.1 Magnetic Force Calculation

The block diagram of the magnetic circuit is shown in Figure 6-4. The U core Electromagnetic actuator has \( N_1 \) turns of copper coil through which a current of \( I_1 \) flows. The permanent magnet is formed as a thin film U shaped magnet with short pole pieces and the equivalent magnetomotive force of the permanent magnet is derived from

![Diagram of U-Core Electromagnet and Equivalent Circuit](image)

Figure 6-4 The magnetic circuit showing the permanent magnet and the electromagnetic microactuator and the equivalent circuit

Equations (6-23) and (6-31) as \( N_2 I_2 \). Both the magnets are separated by an airgap of length \( l \). The pole pieces of the actuator are enlarged to increase the effective area of coupling and have a cross sectional area of \( A \). The length of the actuator core is \( l_c \) and the length of the permanent magnet is \( l_p \) The relative permeability of the actuator core is \( \mu_c \). The total flux flowing through the airgap is \( \Phi \). Thus we have:

\[
\Phi = \frac{N_1 I_1 + N_2 I_2}{R_1 + R_2 + R_c} \tag{6-32}
\]
Where the airgap reluctances are given by:

\[ R = R_1 = R_2 = \frac{l}{\mu_0 A} \]  \hspace{1cm} (6-33)

The reluctance of the magnetic path through the actuator core is:

\[ R_c = \frac{l_c}{\mu_0 \mu_c A} \]  \hspace{1cm} (6-34)

Due to very high relative permeability of the supermalloy core, \( R_c \) becomes negligible in comparison to \( R_1 \) and \( R_2 \). Thus the flux density flowing through the airgap can be expressed as:

\[ \Phi = \frac{N_1 I_1 + N_2 I_2}{2R} \]  \hspace{1cm} (6-35)

Letting

\[ N_1 I_1 + N_2 I_2 = NI \]  \hspace{1cm} (6-36)

and substituting for \( R_1 \) and \( R_2 \), we obtain:

\[ \Phi = \frac{\mu_0 A N I}{2l} \]  \hspace{1cm} (6-37)

The inductance of the magnetic circuit under the given condition is given by:

\[ L = \frac{N\Phi}{l} = \frac{\mu_0 A N^2}{2l} \]  \hspace{1cm} (6-38)

If the permanent magnet is displaced by a small amount \( dl \) in a time interval of \( dt \), the flux \( \Phi \) changes by an amount:

\[ d\Phi = \left. \frac{d\Phi}{dl} \right|_{l=\text{const}} \Phi dl = \frac{\Phi dl}{l} \]  \hspace{1cm} (6-39)

This results in an induced voltage \( \varepsilon \) in the equivalent coil having \( N \) turns and is expressed by:

100
\[ \varepsilon = -I \frac{dL}{dt} \]  \hspace{2cm} (6-40)

In order to keep the current \( I \) constant, a voltage \(-\varepsilon\) must be applied that does the work of amount:

\[ dW_i = -\varepsilon \, ldL = I^2 dL \]  \hspace{2cm} (6-41)

in the time interval of \( dt \). During this displacement, the energy in the magnetic field changes by an amount \( dW_m = I^2 dL/2 \). If the field exerts a force \( F \) on the permanent magnet, a force \(-F\) must be applied in order to increase the airgap by an amount \( dl \).

During the displacement, the work done is of amount \( dW = -Fdl \). Equating the work done on the system to the change in field energy, we get:

\[ DW + dW_i = -Fdl + I^2 dL = \frac{1}{2} I^2 dL \]  \hspace{2cm} (6-42)

and hence

\[ F = \frac{1}{2} I^2 \frac{dL}{dl} \]  \hspace{2cm} (6-43)

Replacing \( IL \) by \( N\Phi \) and using equation (6-39) we obtain:

\[ \frac{F}{2} = -\frac{NI\Phi}{4l} \]  \hspace{2cm} (6-44)

for the force acting on one pole of the permanent magnet. Substituting for \( \Phi \) and simplifying, we obtain the total force acting on the permanent magnet as [47]:

\[ F = -\frac{N^2 I^2 \mu_0 A}{4l^2} \]  \hspace{2cm} (6-45)

where the negative sign indicates that the force is attractive.
6.3.2 Design Procedure

In Chapter 3 it was stated that the output from the VLSI signal processor will drive an electromagnetic actuator that will develop a time-varying magnetic field. This time varying magnetic field will interact with a high coercivity permanent micromagnet that has been implanted on the round window of the Cochlea and will set the micromagnet into motion.

In chapter 4, the maximum force needed to be exerted on the round window of the cochlea was determined as \( 250 \times 10^{-6} \) Newton at a sound pressure of 94 dB SPL and the minimum force at 0 dB SPL as \( 4.8 \times 10^{-9} \) Newton. Based on the maximum force requirement, and assuming an air-gap length of 0.5mm between the actuator pole pieces and the implanted permanent magnet, we have a required magnetomotive force of \( NI = 0.014 / \sqrt{A} \), where \( A \) is the core cross-sectional area. The actuator pole faces were enlarged to an area 600 x 600 micron\(^2\) in order to improve the magnetic coupling with the permanent magnet. This value of \( A \) when substituted into equation (6-45) yields a maximum magnetomotive force of 23.5 AT (ampere-turns).

Due to the requirements of high coercivity and a high maximum energy product the permanent magnet is required to have a high magnetic strength/area ratio. Neodymium-Iron–Boron permanent magnets have a very high intrinsic coercivity (\( iH_c \)) and a very high maximum energy product (\( BH_{max} \)). The magnetic material designated as grade NA35 has an intrinsic coercivity, \( iH_c = 955KA/m \) (min) and a \( BH_{max} = 264-280 \) KJ/m\(^3\) (Table 6-1). This grade is chosen for the implanted magnet attached to the round window. The implanted magnet has the geometry of thin film formed to create a
rectangular U-shaped bar magnet with short pole pieces. The implant’s corresponding
magnetic path length is equal to 1220 microns. From equations (6-23) and (6-31) this
dimension will make the implanted magnet equivalent to an air-core solenoid having a
magnetomotive force of about 19 AT.

The additional ampere-turns required (23.5-19 = 4.5) must be supplied by the
electromagnetic actuator. Due to thermal and power consumption concerns the current in
the actuator coils should be kept to a minimum. Thus, an actuator with a higher number
of turns is desirable.

For a maximum coil current of 1.4 mA, a total of 3200 turns will be required to produce a
magnetomotive force of 4.5 AT. Thus, the number of turns per arm will be 1600. The
1600 turns can be realized by the electrodeposition of 16 levels of copper with bisbenzo-
cyclobutene (BCB) as inter layer insulation. A coil with turns that have a maximum
dimension (height) of 10 microns together with a BCB insulation layer thickness of 5
microns between each coil requires a net winding depth of 1000 microns. If the cross-
sectional area of the pole core is chosen as 200x200 micron², the resulting total base area
of the actuator is approximately 4.54 mm. x 4.54 mm.

The complete design specifications for the actuator is summarized in Table 6-3. Each
layer of the actuator is implemented as an individual module and the connection between
the individual modules is provided by a via-interconnect layer. This method simplifies
the overall design and fabrication process and provides a flexible and simple design and
fabrication methodology.
<table>
<thead>
<tr>
<th>Design parameters</th>
<th>Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum electromagnetic force required, $F$</td>
<td>250 $\mu$Newton</td>
</tr>
<tr>
<td>Air-gap length, $l$</td>
<td>500 $\mu$m</td>
</tr>
<tr>
<td>Cross-sectional area of a pole face, $A$</td>
<td>$600 \mu$m x $600 \mu$m</td>
</tr>
<tr>
<td>Initial $NI$</td>
<td>23.5 AT</td>
</tr>
<tr>
<td>Length of permanent magnet, $l_p$</td>
<td>1220 $\mu$m</td>
</tr>
<tr>
<td>Equivalent magnetomotive force of permanent magnet $N_2l_2$</td>
<td>19AT</td>
</tr>
<tr>
<td>Required magnetomotive force from the actuator, $N_1l_1$</td>
<td>4.5 AT</td>
</tr>
<tr>
<td>Reluctance of an air-gap path, $R_I$</td>
<td>$1.1 \times 10^9$ MKS units</td>
</tr>
<tr>
<td>Total air-gap reluctance, $R=R_I+R_2$</td>
<td>$2.21 \times 10^9$ MKS units</td>
</tr>
<tr>
<td>Maximum magnetic flux, $\varphi$</td>
<td>$1.06 \times 10^{-8}$ Weber</td>
</tr>
<tr>
<td>Maximum magnetic flux, $\varphi$ accounting 10% leakage</td>
<td>$1.17 \times 10^{-8}$ Weber</td>
</tr>
<tr>
<td>Maximum flux density, $B$</td>
<td>0.032 Weber/meter$^2$</td>
</tr>
<tr>
<td>Required magnetomotive force accounting leakage, $NI$</td>
<td>26 AT</td>
</tr>
<tr>
<td>Additional 2.5 AT is to be supplied by permanent magnet</td>
<td>21.5 AT</td>
</tr>
<tr>
<td>implant, so $N_2l_2$</td>
<td></td>
</tr>
<tr>
<td>Modified permanent magnet length, $l_{pm}$</td>
<td>1222 $\mu$m</td>
</tr>
<tr>
<td>Total number of turns of copper coil of the actuator, $N_I$</td>
<td>3200</td>
</tr>
<tr>
<td>Maximum coil current $I_{coil}$</td>
<td>1.4 mA</td>
</tr>
<tr>
<td>Number of module</td>
<td>16</td>
</tr>
<tr>
<td>Turns per module</td>
<td>200</td>
</tr>
<tr>
<td>Turns per-arm of a module</td>
<td>100</td>
</tr>
<tr>
<td><strong>Design parameters</strong></td>
<td><strong>Values</strong></td>
</tr>
<tr>
<td>--------------------------------------------------------------------------------------</td>
<td>----------------------------</td>
</tr>
<tr>
<td>Coil cross-sectional dimensions (width x height)</td>
<td>5 μm x 10 μm</td>
</tr>
<tr>
<td>Spacing between two coil turns</td>
<td>5 μm</td>
</tr>
<tr>
<td>Mean length of coil per turn</td>
<td>4.88 mm</td>
</tr>
<tr>
<td>Coil resistance of each micro actuator cell</td>
<td>350 Ω</td>
</tr>
<tr>
<td>Total coil resistance</td>
<td>5600 Ω</td>
</tr>
<tr>
<td>Each via resistance</td>
<td>15 mΩ</td>
</tr>
<tr>
<td>Total number of via</td>
<td>30</td>
</tr>
<tr>
<td>Resistance of via-interconnect</td>
<td>.45 Ω</td>
</tr>
<tr>
<td>Total resistance of via layer</td>
<td>10 Ω</td>
</tr>
<tr>
<td>Total via resistance</td>
<td>10.45 Ω</td>
</tr>
<tr>
<td>Total resistance</td>
<td>5610.45 Ω</td>
</tr>
<tr>
<td>Maximum power dissipation at $I_{\text{coil}}=1.4$ mA</td>
<td>5.5 mw</td>
</tr>
</tbody>
</table>

Table 6-3 Design specifications of the electromagnetic microactuator
6.4 The Electromagnetic Microactuator Fabrication

6.4.1 Fabrication Requirements

The fabrication of the electromagnetic microactuator requires a magnetic core material to generate the magnetic force, copper coils, insulation material between successive layers of copper coils and to insulate between the turns, a thick photoresist to create a mold for the electrodeposition of copper coils and a seed layer for electrodeposition.

6.4.1.1 Ferro-Silicon substrate

The ferrosilicon substrate is used instead of conventional silicon wafer for a couple of reasons. It will provide a magnetic path between the two arms of the magnetic core deposited by electrodeposition process and as well as will act as the anchor or base plate to support the whole structure. The magnetic properties of the FeSi substrate is given in Table 6-4.

<table>
<thead>
<tr>
<th>$H_c$</th>
<th>$B_{sat}$</th>
<th>$\mu_{max}$</th>
<th>$T_c$</th>
<th>CTE</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.2A/cm</td>
<td>2.03T</td>
<td>30000</td>
<td>750 °C</td>
<td>12 ppm/°C</td>
</tr>
</tbody>
</table>

Table 6-4 Magnetic properties of FeSi substrate

6.4.1.2 Magnetic Core Material

The magnetic materials that are most commonly used in microfabrication include nickel, iron, cobalt, and their alloys with other materials such as chromium and platinum. Nickel-Iron, which is frequently used by the magnetic-recording industry in microfabricated thin-film magnetic-recording heads, was the magnetic core material chosen for the actuator. Since the magnetic properties of a NiFe alloy are very sensitive, its atomic compositions and their impact on device performances must be analyzed.
It is known that stress-induced magnetic anisotropy, is linearly proportional to magnetostriction and the stress in the film. Since the stress in an electroplated film can be large and difficult to control precisely, its effect on the direction of the overall effective easy axis may not be predictable. In general, a deviation from the intended direction of the overall effective easy axis can greatly increase losses and even make the device inoperable. To improve the yield of properly fabricated devices, both the stress and the magnetostriction are required to be minimized.

The magnetostriction of randomly oriented cubic crystallites $\lambda_s$ [48] can be written as

$$\lambda_s = \frac{2\lambda_{100} + 3\lambda_{111}}{5}$$  \hspace{1cm} (6-46)

![Graph showing magnetostriiction vs. weight percent Ni.](image)

**Figure 6-5** Saturation magnetostriiction of NiFe
with the saturation-magnetostriction constants \( \lambda_{100} \) and \( \lambda_{111} \) in the \( <100> \) and \( <111> \) directions respectively. A plot of the saturation magnetostriction of NiFe as a function of nickel content is shown in Figure 6-5. Near 81% Nickel, the magnetostriction and hence the stress-induced magnetic anisotropy are reduced to zero.

The commercially available grade named supermalloy has a composition of 80% Ni and presents exceptionally high initial permeability. For this reason this material is chosen for the core material. The typical magnetization curve of supermalloy [49] is shown in Figure 6-6.

![Figure 6-6 Typical magnetization curves of supermalloy](image)

6.4.1.3 \textit{Thick Photoresist}

The actuator planar coils are to be realized by electrodeposition of copper. To realize the process we need a suitable mold formed by deposition of a photoresist by spin coating method and then to etch it after patterning. The planer copper coil layer is 10 micron high and each coil turn is 5 micron in width with a separation of 5 microns. To realize this high aspect ratio microstructure, a thick photoresist is required that will provide near
perfect vertical sidewalls after etching. Highly expensive LIGA process that involves deep X-ray radiation from a synchrotron to pattern a photoresist provides nearly perfect vertical sidewalls for the mold. With the advent of new technology and newly developed photoresists it is now possible to pattern the photoresists using UV lithography with precision vertical sidewalls requirements at low expenses. As the diffusion of the injected UV ray through the photoresist decreases with increasing depth due to refractive index change, it is a common problem to pattern thick films using UV that results in a etched profile that has a widening near the bottom surface of the film [50]. This non-uniform vertical profile of the mold consequently results in enlarged bottom area for the deposited metal [51]. Obviously this non-uniformity in vertical profile of a microstructure will result in a degraded performance of the fabricated device. A comparative study of different thick photoresists applicable for high aspect ratio UV lithography shows that SU-8, a negative tone EPON epoxy based resin photoresist offers the smallest feature resolution and also requires the lowest exposure dose [52]. The profiles are nearly vertical and the linearity is the most consistent between the top down and cross sectional analysis. For this reason SU-8 is chosen to form the mold for the copper layers deposition. This brand of photoresist provides a uniformity of +/- 3% for an approximately 150 micro thick film that is also a required feature. Thus a uniformity of higher precision can be obtained from a thickness of 10 microns.

6.4.1.4 BCB (bisbenzo-cyclobutene) Insulation

BCB (bisbenzo-cyclobutene), an advanced electronics resin is a polymer derived from B-staged bisbenzocyclobutene monomers. BCB is ideal for applications where a thin dielectric layer is required at the wafer level [53]. A coating of 1.0 to 26.0 microns can
be achieved in a single spin coat. BCB has a low dielectric constant (2.65 which is essentially independent of temperature and frequency), simple and flexible processing, low level of ionics (ppm), low moisture uptake, low cure temperature, rapid thermal curing, high optical transparency, and very low outgassing. BCB can also be dry etched using a plasma process containing both oxygen and a fluorine-containing gas. Due to this reasons, BCB makes itself a good candidate for applications where design and process compatibility is coupled with application durability and hypoallergenic environmental activity.

6.4.1.5 Titanium
Titanium will protect the FeSi substrate from corrosion during subsequent etching of the deposited layers and also will act as a seed layer for electrodeposition of NiFe core.

6.4.1.6 Gold
Gold being a good conductor, a thin layer of gold is used to act as a seed layer for electrodeposition of the copper coils and also used in conjunction with chromium to form the contact points.

6.4.1.7 Amorphous Silicon
A very thin layer of amorphous silicon (α-silicon) is used to break-up the flux path through the core material in between deposition of two successive modules. These insulating layers minimizes eddy-current and hysteresis losses in the core material.
6.4.2 Fabrication

The fabrication process starts on a 0.4 mm thick ferromagnetic substrate. On the substrate, a Ti layer is first sputtered to protect the substrate from corrosion during processing. Then a 10 μm thick layer of BCB (bisbenzo-cyclobutene) is spin-coated and cured. The BCB layer is then patterned and dry etched (Figure 6-7).

![Figure 6-7 Cross-section of the FeSi substrate after deposition of the titanium layer and etched BCB layer](image)

Next, the gold seed layer is deposited using e-beam evaporation technology. A 12 μm thick layer of SU-8 photoresist is spin deposited and then cured, patterned and developed to create a mold for the copper coil turns (Figure 6-8).

![Figure 6-8 A mold is formed using SU-8 photoresist for electrodeposition of the first copper coil layer over the gold seed layer](image)
Next, the first copper coil layer is realized by electrodepositing copper into the photoresist mold. One hundred turns of rectangular copper wire are deposited with a width 5 µm, a depth of 10 µm and a separation of 5 µm between adjacent turns. A 10 µm gap is provided between the pole piece and the closest copper turns. The SU-8 mold is then stripped and the gold seed layer is etched to isolate the coil turns electrically (Figure 6-9).

![Copper coil turns](image)

Figure 6-9 First copper coil layer is realized by electrodeposition of copper in the SU-8 photoresist mold and the SU-8 mold is then stripped.

A second BCB insulation layer is then spin deposited on top of the first Copper coil structure and a via hole is opened to provide connectivity to the coil (Figure 6-10).

![2nd BCB layer](image)

Figure 6-10 Second BCB layer is deposited, patterned and etched to open the via hole.
A second gold seed layer is then e-beam evaporated for deposition of the via-interconnect channel copper layer. Then the via-interconnect channel copper layer is electrodeposited using another SU-8 photoresist mold structure. After the mold is stripped away, the gold layer is etched as before. A third layer of BCB is then spin-deposited and cured. The

![Diagram](image)

Figure 6-11: The copper via-interconnect channel is electrodeposited and is followed by deposition and etching of the third BCB layer.

The third BCB layer is patterned and etched using SF$_6$ plasma over the copper via-interconnect channel (Figure 6-11). The BCB layers are then through etched using SF$_6$ plasma up to the titanium seed layer over the substrate only in the regions where the pole structures have to be formed by electrodeposition of NiFe core material, i.e. in the center of the coils (Figure 6-12).

![Diagram](image)

Figure 6-12: The BCB layers are through etched in the central region for deposition of the NiFe core.
The NiFe core material is then electrodeposited in the opened holes (Figure 6-13). A very thin layer (50 nm) of amorphous silicon is then PECVD deposited over the pole faces (Figure 6-14). The flux-path through the core material is thus segmented at each layer by an amorphous silicon layer. These insulating layers will minimize eddy-current and hysteresis losses in the core. This process sequence completes the fabrication of a single module of a micro-electromagnetic actuator.

Subsequent modules are then deposited by repeating the appropriate process sequence until the desired number of turns is achieved. After depositing the final module, an enlarged pole faces are electrodeposited on the U-shaped core. (Figure 6-15)
Finally, the top surfaces of the two pole faces are polished to obtain a pole-to-pole planarity of less than 1 μm.

The fabrication simulation of a single module of the MEMS electromagnetic microactuator is carried out using IntelliSuite and no process incompatibility was reported. The complete process sequence for fabrication of a single module is given in Appendix V.

The advantage of this modular implementation is that an electromagnetic microactuator with a varying number of turns can be built by repeating a few process steps.
Chapter 7

CONCLUSIONS

In this thesis two microelectromechanical (MEMS) structures have been designed for use in a hearing instrument. The research work has concentrated on the design aspect and the process technology needed to create the desired microstructure devices. The process steps have been identified, especially in the case where a rapidly evolving fabrication technology can lead to an innovative geometry for the device.

A MEMS realization of a capacitive microphone has been designed that has a sensitivity of 52 mV/Pa and a frequency response extending up to 18 kHz. These performance figures are higher than those of the existing MEMS microphones. The microphone is designed using a low residual stress polysilicon germanium diaphragm of thickness 0.8 micron and an area of 2.5 x 2.5 mm², an airgap of thickness 3.0 micron and a silicon nitride backplate of thickness 1.0 micron with 1280 acoustical ports. This large number of acoustical ports will minimize the viscous losses at high frequencies due to compression of the air in the airgap. Each acoustical port has a dimension of 30 x 30 micron². The microphone is constructed using a combination of surface and bulk micromachining techniques. Due to the uniformity of residual stress along the direction of growth, poly silicon germanium diaphragm has an almost flat surface plane. Besides, low deposition temperature, conformal deposition, and low temperature doping properties of polysilicon germanium resulted in a uniformly thick diaphragm and easy CMOS integration. A Dickson type DC-DC voltage converter will be used to supply a bias voltage of 12 V DC. A buffer amplifier made of a simple source follower with high gain
and low input noise properties can be used to convert the current flow due to capacitance change into a usable ac voltage.

The MEMS realization of the electromagnetic microactuator employs a novel surface micromachined modular fabrication process. A magnetomotive force of 4.5 AT is realized by electrodeposition of 3200 copper coil turns and the associated supermalloy core segments in a modular fashion. Each module is constructed by electrodeposition of 200 turns of planar copper coil in a SU-8 negative photoresist mold, and electrodeposition of supermalloy (Ni 80%) core segment. The coil is 5 micron in width and 10 micron in height with an insulation spacing of 5 micron between adjacent coil turns. The supermalloy core has a dimension of 200 x 200 micron$^2$. BCB (Bisbenzo-cyclobutene), an electronic insulator material is used between each layer of planar copper coil and also to insulate the adjacent copper coil turns. A via-interconnect copper layer of a single turn embedded in the BCB insulation layer is used to provide connection between two successive modules. The magnetic flux path through the successive layers of the supermalloy core is segmented by depositing a 50 nm thin layer of amorphous silicon to restrict eddy current flow. An enlarged pole face of 600 x 600 micron$^2$ is used to realize a greater electromagnetic attraction force.

The advantage of this modular implementation is the magnetomotive force characteristics of the microactuator can be modified by changing the required number of modules. At a maximum coil current of 1.4 mA the microactuator consumes a power of 5.5 mW.

The design of a thin film micro-permanent magnet is optimized to use in conjunction with the microactuator for magnetic actuation of the round window of the cochlea. The
U-shaped thin film permanent magnet implant is designed using a Neodymium-Iron-Boron permanent magnet material that has a very high intrinsic coercivity ($iH_{c}$) and a very high maximum energy product $(BH)_{\text{max}}$. The equivalent magnetomotive force of the 1222 micron long permanent magnet implant is 21.5 AT. The electromagnetic microactuator and the permanent magnet implant together form a magnetic circuit that has a magnetomotive force of 26 AT. This amount of magnetomotive force is capable of producing a magnetic attraction force of $250 \times 10^{-6}$ Newton in an airgap of 500 micron length between the pole pieces of the microactuator and the permanent magnet. These three microstructures together with a VLSI signal process form an acousto-magnetic transduction system for use in a hearing instrument that resulted in better performance characteristics than a conventional hearing instrument.

Acoustical signal from the external world will be captured by the microphone implanted in the ear canal. The output signal from the microphone will undergo signal conditioning and processing in the VLSI signal processor. The output of the signal processor that will be an analog time varying signal, having same dynamic characteristics as the external acoustical signal will drive the electromagnetic microactuator to generate a time varying magnetic field. By this way an acousto-magnetic transduction can be realized to have a magnetic field that has a dynamic characteristic as an acoustical signal. This time-varying magnetic field will interact with the magnetic field produced by the micro permanent magnet that has been implanted on the round window of the cochlea and will force the micro permanent magnet into vibrational motion. Eventually the round window will be forced into motion as it is a thin membrane structure and is tacked to the micro permanent magnet implant. Thus it will be possible to create a pressure difference
between the oval and the round window of the cochlea using an artificial source. As the nature of this pressure difference will be very much similar to the pressure difference created during normal hearing process, the same way the cochlear fluid will flow to develop traveling waves in the basilar membrane. This eventually will excite the inner and the outer hair cells to send neuro-signal to the respective section of the brain.

The MEMS structures have been designed and simulated using the IntelliSuite MEMS software package from the IntelliSense Corporation. This package offers a three-dimensional animation capability for determining the motions and fields associated with the microstructures. The simulations were checked with some structures with known theoretical properties and found to be very accurate. The devices have not yet been fabricated.

This type of microactuator coupled with a micro permanent magnet implant can be used as an enabling element in a variety of applications requiring a bionic interface for the artificial actuation of micro-vibrational bio-mechanisms like heart valves, body muscles, nerve conduits, etc.
REFERENCES


Appendix I

MATLAB IMPLEMENTATION OF ROWSOSKI
EXTERNAL AND MIDDLE-EAR MODEL

% J. ROSOWSKI EXTERNAL AND MIDDLE-EAR MODEL
% Ref: "Models of External and Middle-Ear Function"
% in "Auditory Computation", SHAR-6, Springer, 1993
% function [Zeq,Peq,PC,PS,PT,PEC,PEX] = rosowski(f,PPW)
% PPW - Plane wave pressure
% f - frequency values
% Outputs:
% PEX - external pressure (at pinna)
% PEC - pressure at ear canal
% PT - pressure at tympanic membrane
% PC - pressure at the base of the cochlea
% Peq - Thevenin pressure,
% Zeq - Thevenin impedance, according to the following model:
% NOTE: uses Bauer's 67 model for sound diffraction and scattering by the head.
% (Giguere, & Woodland, JASA (95)-1,1994, pp.332)
% F. Perdigao, IT-Coimbra, Dec.1996
function [Zeq,Peq,PC,PS,PT,PEC,PEX,ZiT] = roowski(f,PPW)

if nargin < 2, PPW=1; end

close all;

s=j*2*pi*f;

w=2*pi*f;

c=343; %--- propagation velocity of sound in air

r0=1.21; %--- air density

A0=4.4e-4; %--- concha horn area (wide end)

A1=A0/10; %--- concha horn area (narrow end)

IH=0.01; %--- horn length

IT=0.02; %--- tube length

as=0.1; %--- radius of head sphere

%************************************************************

% 1st part: The Equivalent Pressure Source,

% Diffraction and Scattering of Sound by the head

%************************************************************

%Simplified model of Bauer, 67

%(Giguere, & Woodland, JASA (95)-1,1994, pp.332)

%************************************************************

Zh=s*r0/(2*pi*as);

Rh=r0*c/(pi*(as^2));

Zr=s*0.7*r0/sqrt(pi*A0);

Rr=r0*c/A0;
GS=(Rh+2*Zh)/(Rh+Zh); %6dB gain at high frequencies.
ZR= zparal(Rh,Zh)+zparal(Rr,Zr); %Thevenin source impedance

%******************************************************************************

% 2nd part: Concha and External Canal,
%******************************************************************************

%--- HORN

alpha=0;
k=w/c-j*alpha;
a=0.5*log(Al/A0)/lH;
b=sqrt(k.^2 - a^2);
k1=-a-j*b;
k2=-a+j*b;
AH=exp(a*lH)*(k2.*exp(j*b*lH)-k1.*exp(-j*b*lH))./(2*j*b);
BH=exp(a*lH)*s*r0.*sin(b*lH)./(Al*b);
CH=-exp(-a*lH)*Al*(a^2+b.^2).*sin(b*lH)./(s*r0.*b);
DH=exp(-a*lH)*(k2.*exp(-j*b*lH)-k1.*exp(j*b*lH))./(2*j*b);

%TUBE
k=w/c-j*alpha;
z0=r0*c/Al;
AT=cos(k*lT);
BT=j*z0*sin(k*lT);
CT=j/z0*sin(k*lT);
DT=AT;
A=AH.*AT+BH.*CT;
B=AH.*BT+BH.*DT;
C=CH.*AT+DH.*CT;
D=CH.*BT+DH.*DT;

%****************************************************************************************

% 3rd part: Thevenin Equivalent at tympanic membrane
%****************************************************************************************

% GS*PPW = ZR*UEX + PEX;
%    = ZR*(C*PT+D*UT) + (A*PT+B*UT)
%    = PT*(ZR*C+A) + UT*(ZR*D+B)

****************************************************************************************

Pem = PPW.*GS./(ZR.*C+A);
Zem = (ZR.*D+B)./(ZR.*C+A);
semilogx(f,db(Pem))

%****************************************************************************************

% 4th part: Thevenin Equivalent at stapes
%****************************************************************************************

%Middle-Ear impedances
ATM=60e-6;
AFP=3.2e-6;
rMI=1.3;
CMC=3.9e-11;
CTC=4e-12;
RA=6e6;
LA=100; LT1=750; LT=6.6e3;
CT=3e-12; CT2=1.3e-11;
RT2=1.2e7;
RT=3e7;
CTS=1.1e-3;
RTS=4.3e-2;
RMI=7.2e-3;
LMI=7.9e-6;
%CMI=Inf;
LS=3e-6;
RJ=3.6;
CJ=4.9e-4;
CAL=9.4e-15;

%***************************************************

% COCHLEAR INPUT IMPEDANCE

%***************************************************

R0C=5e10;
RC=1e11;
LC=8e7;
CRW=1e-13;
ZC=zparal(RC,R0C+s*LC) +1.0/(s*CRW);
ZAL=1.0/(s*CAL) +1e10;
ZTC = zparal(1.0/(s*CTC), s*LA+RA+1.0/(s*CMC));
ZT = s*LT1 + zparal(1.0/(s*CT)+RT+s*LT, 1.0/(s*CT2)+RT2);
ZTS = 1.0/(s*CTS)+RTS;
%ZMI = RMI+s*LMI+1.0/(s*CMI);
ZMI = RMI+s*LMI; ZJ = RJ+1.0/(s*CJ);
% Reduction to transformer secondaries
%******************************************
% 1st transformer
%******************************************
Peq = ATM*Pem;
Zeq = (ATM^2)*(Zem + ZT + ZTC);
%******************************************
% 2nd transformer
%******************************************
Peq = rMI*Peq.*ZTS./ZTS+Zeq;
Zeq = (rMI^2)*(ZMI + zparal(Zeq,ZTS));
%******************************************
% 3rd transformer
%******************************************
Peq = (1/AFP)*Peq.*ZJ ./ZJ+Zeq;
Zeq = (1/(AFP^2))*(s*LS + zparal(Zeq,ZJ));
% Series Equivalent Impedance seen at cochlea
Zeq = Zeq + ZAL + 1.0/(s*CRW);
PC=Peq.*ZC./(ZC+Zeq+ZAL);
PS=Peq.*(ZC+ZAL)./(ZC+Zeq+ZAL);

--- ZiT - Impedance at Tympanus

ZiT = (ZAL+ZC)*(AFP^2);
ZiT = zparal(s*LS + ZiT, ZJ);
ZiT = ZiT/(rMI^2);
ZiT = zparal(ZMI+ZiT,ZTS);
ZiT = ZiT/(ATM^2);
ZiT = ZT+ZiT+ZTC;

PT/PPW

PT = PPW.*GS./(ZR.*(C+D./ZiT)+A+B./ZiT);

PEC and PEX

PEC = PT.*(AT+BT./ZiT);
PEX = PT.*(A+B./ZiT);
ZiEX = (A+B./ZiT)./(C+D./ZiT);
ZiEC = (AT+BT./ZiT)./(CT+DT./ZiT);
ZoeX=(B+D.*ZR)./(A+C.*ZR);
semilogx(f,db2(PC),f,db2(PS),f,db2(PT),f,db2(PEC),f,db2(PEX))
figure

plot(f,abs(PC),f,abs(PS),f,abs(PT),f,abs(PEC),f,abs(PFX));

return

%******************************************************************************

% Function zparal

%--Z1//Z2:

%--Z = (Z1.*Z2)./(Z1+Z2);

function Z = zparal(Z1,Z2)

Z = (Z1.*Z2)./(Z1+Z2);
Appendix II

MATLAB CODE TO CALCULATE
MICROPHONE FREQUENCY RESPONSE

% Polysilicon germanium microphone frequency response
% Equations (5-1) to (5-19)
% Reference:
% P.C. Hsu, C.H. Mastrangelo and K. D. Wise. “A High Density Polysilicon Diaphragm
Condenser %Microphone”, Conference Record IEEE 11th International Workshop On

***************************************************************************
close all
clear all

% INPUT
% all units are in MKS system
E=1.55e11; % Young's Modulus
t=.8e-6; % Poly SiGe diaphragm thickness
v=.28; % Poisson's ratio
D=E*t^3/(12*(1-v^2)); % Rigidity
sigmar=20e6; % Residual stress;
rho=3200*t; % Area density of poly SiGe diaphragm
rhol=1/rho;
\( \text{rho}_0 = 1.21; \) % Air density

\( a = 2.5 \times 10^{-3}; \) % Diaphragm side length and width

\( d = 3 \times 10^{-6}; \) % Average air-gap distance

\( c = 343; \) % Velocity of sound in air

\( \text{bpt} = 1 \times 10^{-6}; \) % Backplate thickness

\( f = 1.5 \times 10^5; \) % Frequency

\( \varepsilon_0 = 8.854 \times 10^{-12}; \) % Air permittivity

\( V_b = 12; \) % Bias voltage

% CALCULATION

\( w = 2\pi f; \)

\( T = (\sigma \text{mar} \ast t); \)

\( TT = (D \ast \pi^2) / a^4 + (T / (2 \ast a^2)); \)

\( \text{fres} = \sqrt{\rho_1 \ast (TT)} \) % First resonant frequency

\( R_r = (\rho_0 \ast a^4 \ast w^2 / (2 \ast \pi \ast c)); \) % Air radiative resistance

\( M_r = 8 \ast \rho_0 \ast a^3 / (3 \ast \pi \ast \sqrt{\pi}); \) % Air acoustical mass

\( C_m = (32 \ast a^2 / ((\pi^6) \ast (2 \ast \pi^2 \ast D + a^2 \ast T))); \) % Diaphragm compliance

\( \text{Mm} = (\pi^4 \ast \rho_0 \ast (2 \ast \pi^2 \ast D + a^2 \ast T) / (64 \ast T)); \)

\( r = 30 \times 10^{-6}; \) % Backplate hole radius

\( A_A = 256 \times 5; \) % Number of holes

\( A_A = A_A \ast r^2; \) % Area covered by holes

\( \alpha = A_A \ast r^2; \) % Surface fraction occupied by holes

\( n = (A_A / a^2); \) % Number of holes per unit area

133
eta=17.1e-6; % Air viscosity coefficient

% Airgap viscosity loss:
Rg=(12*eta*a^2*((alpha/2)-(alpha^2/8)-(log(alpha)/4)-3/8))/(n*d^3*pi);

Ca=d/(rho0*c^2*alpha^2*a^2); % Compliance of Rg
Rh=8*bpt*eta*a^2/(pi*n*r^4); % Viscosity loss of the backplate holes

% OUTPUT

% Equivalent impedance:
Zt=Rr+(j*w*(Mr+Mm)+1./(j*w*Cm))+(Rg+Rh)./(1+j*w*(Rg+Rh)*Ca);

S=Vb*a^2./(j*(w*d.*Zt)); % Sensitivity
SdB=20*log10(abs(S)); % Sensitivity in dB

semilogx(f,SdB)
title(['Frequency Response of Capacitive Microphone']);
xlabel(['Frequency in Hz']);
ylabel(['Sensitivity (dB, reference 1V/Pa)']);

grid

figure

semilogx(f,abs(S)); % Sensitivity in absolute value

title(['Frequency Response of Capacitive Microphone']);
xlabel(['Frequency in Hz']);
ylabel(['Sensitivity (mV/P)']);

grid

cmic=epsilon0*a^2/d%; % Microphone capacitance

Vp=1.9275*sqrt((T*d^3)/(epsilon0*a^2)) % Pull-in voltage
Appendix III

INTELLISUITE SIMULATION OF THE
MICROPHONE FABRICATION PROCESS

Number of steps in Process: 64
1. Definition Si Czochralski 100
   Operation on both sides.
   t_film: 250000 nm
   diameter: 101.6 mm
   flat_dir: 100 Vector
   dope_conc: 1e+15 /cm3
   resist: 2.54 Ohm-cm

2. Etch Si Clean Pirahna
   Operation on both sides.
   H2SO4_conc: 75 %
   H2O2_conc: 25 %
   time_etch: 10 min

3. Deposition Si3N4 LPCVD SIH2CL2_NH3
   Operation on top side.
   T_dep: 850 deg_C
   P_dep: 0.026 Pa
   SiH2Cl2_NH3_R: 3.88 ratio
   Results:
   t_film: 1000 nm

4. Deposition Ti Sputter Ar-Ambient
   Operation on top side.
   Rf_Pow: 250 W
   Rf_Freq: 13.56 MHz
   P_dep: 0.1 Pa
   time_dep: 10 min
   Results:
   t_film: 30 nm

5. Deposition Au E-Beam E-Beam
   Operation on top side.
   Rf_Pow: 100 W
   Rf_Freq: 13.56 MHz
T_{dep}: 100 \text{ deg C} \\
P_{vac}: 10 \text{ uTorr} \\
time_{dep}: 10 \text{ min} \\

Results: \\
t_{film}: 30 \text{ nm} \\

6. Deposition SiO2 Thermal Dry \\
   Operation on both sides. \\
   T_{dep}: 1100 \text{ deg C} \\
   P_{dep}: 101325 \text{ Pa} \\
   time_{dep}: 60 \text{ min} \\
   O_{2pp}: 1 \text{ part.pr} \\

Results: \\
t_{film}: 3000 \text{ nm} \\
t_{etch}: 3000 \text{ nm} \\
t_{after}: 0 \text{ nm} \\

7. Deposition PR-S1800 Spin S1805 \\
   Operation on top side. \\
   Speed: 4000 \text{ rpm} \\
   time_{spin}: 30 \text{ sec} \\
   T_{soft}: 115 \text{ deg C} \\
   time_{soft}: 60 \text{ sec} \\
   \lambda: 436 \text{ nm} \\

Results: \\
t_{film}: 500 \text{ nm} \\

8. Definition UV Contact Suss \\
   Operation on top side. \\
   mask_no: 501 # \\
   Power: 250 \text{ W} \\
   \lambda: 436 \text{ nm} \\
   time_{exp}: 10 \text{ sec} \\

Results: \\
dose: 52 \text{ J} \\

9. Etch PR-S1800 Wet 1112A \\
   Operation on both sides. \\
   T_{etch}: 20 \text{ deg C} \\
   time_{etch}: 5 \text{ min} \\

Results: \\
t_{etch}: 990000 \text{ nm} \\

10. Etch SiO2 Wet BHF \\
    Operation on both sides. \\
    HF_{conc}: 80 \%
H2Opp: 1 part.pr
time_etch: 8 min

Results:
t_etch: 4000 nm

11. Etch PR-S1800 Wet 1165
Operation on both sides.
T_etch: 20 deg_C
time_etch: 5 min

Results:
t_etch: 990000 nm

12. Deposition Si3N4 PECVD Ar
Operation on top side.
T_dep: 275 deg_C
P_dep: 127 Pa
Rf.Pow: 250 W
Rf_FREQ: 13.56 MHz
tot_fl: 1500 sccm
Arpp: 0.959 part.pr
NH3pp: 0.024 part.pr
SiH4pp: 0.017 part.pr

Results:
tFilm: 300 nm

13. Deposition PR-S1800 Spin S1805
Operation on top side.
Speed: 4000 rpm
time_spin: 30 sec
T_soft: 115 deg_C
time_soft: 60 sec
lambda: 436 nm

Results:
t_film: 500 nm

14. Definition UV Contact Suss
Operation on top side.
mask_no: 502 #
Power: 250 W
lambda: 436 nm
time_exp: 10 sec

Results:
dose: 52 J

15. Etch PR-S1800 Wet 1112A
Operation on both sides.
T_etch: 20 deg_C
  time_etch: 5 min

Results:
  t_etch: 990000 nm

16. Etch Si3N4 Wet Wet
    Operation on top side.
    H2PO4_conc: 75 %
    H2O2_conc: 15 %
    time_etch: 180 min

Results:
  t_etch: 300 nm

17. Etch PR-S1800 Wet 1165
    Operation on both sides.
    T_etch: 20 deg_C
    time_etch: 5 min

Results:
  t_etch: 990000 nm

18. Etch Si3N4 CLEAN RCA
    Operation on top side.
    time_etch: 30 min
    T_etch: 80 deg_C

19. Deposition PolySi LPCVD SiH4
    Operation on both sides.
    T_depl: 630 deg_C
    P_depl: 53 Pa
    time_depl: 15 min
    T_anne: 1100 deg_C
    time_anne: 0 min

Results:
  t_film: 40 nm

20. Deposition PolySiGe LPCVD Standard
    Operation on top side.
    T_depl: 475 deg_C
    P_depl: 0.33 Pa

Results:
  t_film: 760 nm

21. Deposition P Implant P
    Operation on top side.
    ion_dose: 1e+14 /cm3
    ion_energy: 8.01025 fJ
Results:
  depth_in: 800 nm
  depth_jp: 800 nm

22. Deposition P Anneal Generic
   Operation on top side.
   T_anne: 1050 deg_C
   time_an: 60 min
   N2pp: 1 part.pr
   P_dep: 103325 Pa

Results:
  depth_in: 800 nm

23. Deposition PR-S1800 Spin S1805
   Operation on top side.
   Speed: 4000 rpm
   time_spin: 30 sec
   T_soft: 115 deg_C
   time_soft: 60 sec
   lambda: 436 nm

Results:
  t_film: 500 nm

24. Definition UV Contact Suss
   Operation on top side.
   mask_no: 503 #
   Power: 250 W
   lambda: 436 nm
   time_exp: 10 sec

Results:
  dose: 52 J

25. Etch PR-S1800 Wet 1112A
   Operation on both sides.
   T_etch: 20 deg_C
   time_etch: 5 min

Results:
  t_etch: 990000 nm

26. Etch PolySiGe Dry SF6-Plasma
   Operation on top side.
   P_etch: 19.998 Pa
   Power: 375 W
   time_etch: 5 min

Results:
  t_etch: 760 nm
27. Etch PolySi Dry SF6-Plasma
   Operation on both sides.
   time_etch: 15 min
   P_base: 0.0001 Pa
Results:
   t_etch: 40 nm

28. Etch PR-S1800 Wet 1165
   Operation on both sides.
   T_etch: 20 deg_C
   time_etch: 5 min
Results:
   t_etch: 990000 nm

29. Deposition SiO2 Thermal Dry
   Operation on both sides.
   T_dep: 1100 deg_C
   P_dep: 101325 Pa
   time_dep: 60 min
   O2pp: 1 part.pr
Results:
   t_film: 108 nm
   t_etch: 108 nm
   t_after: 0 nm

30. Deposition PR-S1800 Spin S1805
   Operation on top side.
   Speed: 4000 rpm
   time_spin: 30 sec
   T_soft: 115 deg_C
   time_soft: 60 sec
   lambda: 436 nm
Results:
   t_film: 500 nm

31. Definition UV Contact Suss
   Operation on top side.
   mask_no: 504 #
   Power: 250 W
   lambda: 436 nm
   time_exp: 10 sec
Results:
   dose: 52 J

32. Etch PR-S1800 Wet 1112A
Operation on both sides.
T_etch: 20 deg_C
time_etch: 5 min

Results:
t_etch: 990000 nm

33. Etch SiO2 Wet BHF
Operation on both sides.
HF_conc: 80 %
H2O2_pp: 1 part.pr
time_etch: 8 min

Results:
t_etch: 108 nm

34. Etch Si3N4 Wet Wet
Operation on top side.
H2PO4_conc: 75 %
H2O2_conc: 15 %
time_etch: 180 min

Results:
t_etch: 990000 nm

35. Etch Si3N4 CLEAN RCA
Operation on top side.
time_etch: 30 min
T_etch: 80 deg_C

36. Deposition SiO2 Thermal Dry
Operation on both sides.
T_dep: 1100 deg_C
P_dep: 101325 Pa
time_dep: 60 min
O2_pp: 1 part.pr

Results:
t_film: 500 nm
t_etch: 500 nm
t_after: 0 nm

37. Deposition PR-S1800 Spin S1805
Operation on bottom side.
Speed: 4000 rpm
time_spin: 30 sec
T_soft: 115 deg_C
time_soft: 60 sec
lambda: 436 nm

Results:
t_film: 500 nm

38. Definition UV Contact Suss
   Operation on bottom side.
   mask_no: 505 #
   Power: 250 W
   lambda: 436 nm
   time_exp: 10 sec
Results:
   dose: 52 J

39. Etch PR-S1800 Wet 1112A
   Operation on both sides.
   T_etch: 20 deg_C
   time_etch: 5 min
Results:
   t_etch: 990000 nm

40. Etch SiO2 Wet BHF
   Operation on both sides.
   HF_conc: 80 %
   H2Opr: 1 part.pr
   time_etch: 8 min
Results:
   t_etch: 500 nm

41. Etch PR-S1800 Wet 1165
   Operation on both sides.
   T_etch: 20 deg_C
   time_etch: 5 min
Results:
   t_etch: 990000 nm

42. Etch Si Dry SF6_O2
   Operation on bottom side.
   Rf_Freq: 13.56 MHz
   Rf_Pow: 1000 W
   P_etch: 0.133 Pa
   time_etch: 12.5 min
   T_etch: -110 deg_C
Results:
   t_etch: 250000 nm

43. Etch SiO2 Wet BHF
   Operation on both sides.
   HF_conc: 80 %
H₂O₂pp: 1 part.pr

time_etch: 8 min

Results:

t_etch: 500 nm

44. Etch Si Clean RCA
   Operation on both sides.
   time_etch: 30 min
   T_etch: 80 deg_C

45. Deposition SiO₂ Thermal Dry
   Operation on both sides.
   T_dep: 1100 deg_C
   P_dep: 101325 Pa
   time_dep: 60 min
   O₂pp: 1 part.pr

Results:

t_film: 108 nm

t_etch: 108 nm

t_after: 0 nm

46. Deposition PR-S1800 Spin S1805
   Operation on bottom side.
   Speed: 4000 rpm
   time_spin: 30 sec
   T_soft: 115 deg_C
   time_soft: 60 sec
   lambda: 436 nm

Results:

t_film: 500 nm

47. Definition UV Contact Suss
   Operation on bottom side.
   mask_no: 506 #
   Power: 220 W
   lambda: 436 nm
   time_exp: 10 sec

Results:

dose: 52 J

48. Etch PR-S1800 Wet 1112A
   Operation on both sides.
   T_etch: 20 deg_C
   time_etch: 5 min

Results:

t_etch: 990000 nm
49. Etch SiO2 Wet BHF
   Operation on both sides.
   HF_conc: 80 %
   H2Opp: 1 part.pr
   time_etch: 8 min
Results:
   t_etch: 500 nm

50. Etch Si3N4 RIE RIE
   Operation on bottom side.
   time_etch: 5 min
Results:
   t_etch: 990000 nm

51. Etch Ti RIE SF6/CHF4_O2
   Operation on bottom side.
   P_etch: 0.00133 Pa
   time_etch: 5 min
   Rf_Freq: 13.56 MHz
   Rf_Pow: 750 W
Results:
   t_etch: 30 nm

52. Etch Au RIE CL2-Plasma
   Operation on bottom side.
   P_etch: 0.1333 Pa
   T_etch: 110 deg.C
   Rf_Freq: 13.56 MHz
   Rf_Pow: 1500 W
Results:
   t_film: 30 nm

53. Etch PR-S1800 Wet 1165
   Operation on both sides.
   T_etch: 20 deg.C
   time_etch: 5 min
Results:
   t_etch: 990000 nm

54. Etch SiO2 Wet BHF
   Operation on both sides.
   HF_conc: 80 %
   H2Opp: 1 part.pr
   time_etch: 8 min
Results:
t_etch: 500 nm

55. Etch Si Clean RCA
   Operation on both sides.
   time_etch: 30 min
   T_etch: 80 deg_C

56. Deposition Cr Evaporate Evaporation
   Operation on top side.
   P_dep: 0.1 Pa
   time_dep: 10 min
Results:
   t_film: 50 nm

57. Deposition Au E-Beam E-Beam
   Operation on top side.
   Rf_Pow: 100 W
   Rf_Freq: 13.56 MHz
   T_dep: 100 deg_C
   P_vac: 10 uTorr
   time_dep: 10 min
Results:
   t_film: 400 nm

58. Deposition PR-S1800 Spin S1805
   Operation on top side.
   Speed: 4000 rpm
   time.spin: 30 sec
   T_soft: 115 deg_C
   time_soft: 60 sec
   lambda: 436 nm
Results:
   t_film: 500 nm

59. Definition UV Contact Suss
   Operation on top side.
   mask_no: 508 #
   Power: 250 W
   lambda: 436 nm
   time_exp: 10 sec
Results:
   dose: 52 J

60. Etch PR-S1800 Wet 1112A
   Operation on both sides.
   T_etch: 20 deg_C
time_etch: 5 min

Results:

t_etch: 990000 nm

61. Etch Au Wet PAN
    Operation on top side.
    H2PO4_conc: 75 %
    C2H4O2_conc: 15 %
    HNO3_conc: 5 %
    time_etch: 5 min

Results:

t_etch: 400 nm

62. Etch Cr Wet PAN
    Operation on top side.
    H2PO4_conc: 75 %
    C2H4O2_conc: 15 %
    HNO3_conc: 5 %
    time_etch: 1 min

Results:

t_etch: 990000 nm

63. Etch PR-S1800 Wet 1165
    Operation on both sides.
    T_etch: 20 deg_C
    time_etch: 5 min

Results:

t_etch: 990000 nm

64. Etch SiO2 Wet Sacrifice
    Operation on bottom side.
    time_etch: 60 min
    T_etch: 72 deg_C

Results:

t_etch: 1e+06 nm
Appendix IV

FEA_JOB.MESSAGE FILE FOR THE ELECTROMECHANICAL SIMULATION OF THE MICROPHONE

(Note: This message is automatically generated after the simulation run. The format is kept unchanged)

1 ABAQUS VERSION 5.8-1 DATE 02-APR-2000 TIME 19:07:54 PAGE 1

Created From IntelliCad
LOAD CASE 1
STEP 13 INCREMENT 1 STEP TIME 0.

STEP 13 STATIC ANALYSIS

LOAD CASE 1

AUTOMATIC TIME CONTROL WITH -
A SUGGESTED INITIAL TIME INCREMENT OF
AND A TOTAL TIME PERIOD OF
THE MINIMUM TIME INCREMENT ALLOWED IS
THE MAXIMUM TIME INCREMENT ALLOWED IS

CONVERGENCE TOLERANCE PARAMETERS FOR FORCE
CRITERION FOR RESIDUAL FORCE FOR A NONLINEAR PROBLEM
CRITERION FOR DISP. CORRECTION IN A NONLINEAR PROBLEM
INIT. VALUE OF TIME AVG. FORCE IS TIME AVG. FORCE IN PREVIOUS STEP
AVERAGE FORCE IS TIME AVERAGE FORCE
ALTERNATE CRIT. FOR RESIDUAL FORCE FOR A NONLINEAR PROBLEM
CRITERION FOR ZERO FORCE RELATIVE TO TIME AVG. FORCE
CRITERION FOR RESIDUAL FORCE WHEN THERE IS ZERO FLUX
CRITERION FOR DISP. CORRECTION WHEN THERE IS ZERO FLUX
CRITERION FOR RESIDUAL FORCE FOR A LINEAR INCREMENT
FIELD CONVERSION RATIO

VOLUMETRIC STRAIN COMPATIBILITY TOLERANCE FOR HYBRID SOLIDS
AXIAL STRAIN COMPATIBILITY TOLERANCE FOR HYBRID BEAMS
TRANS. SHEAR STRAIN COMPATIBILITY TOLERANCE FOR HYBRID BEAMS
SOFT CONTACT CONSTRAINT COMPATIBILITY TOLERANCE FOR P>P0
SOFT CONTACT CONSTRAINT COMPATIBILITY TOLERANCE FOR P=0.0
DISPLACEMENT COMPATIBILITY TOLERANCE FOR DCoup ELEMENTS
ROTATION COMPATIBILITY TOLERANCE FOR DCoup ELEMENTS

TIME INCREMENTATION CONTROL PARAMETERS:
FIRST EQUILIBRIUM ITERATION FOR CONSECUTIVE DIVERGENCE CHECK
EQUILIBRIUM ITERATION AT WHICH LOG. CONVERGENCE RATE CHECK BEGINS 8
EQUILIBRIUM ITERATION AFTER WHICH ALTERNATE RESIDUAL IS USED 9
MAXIMUM EQUILIBRIUM ITERATIONS ALLOWED 16
EQUILIBRIUM ITERATION COUNT FOR CUT-BACK IN NEXT INCREMENT 10
MAXIMUM EQUILIB. ITERS IN TWO INCREMENTS FOR TIME INCREMENT INCREASE 4
MAXIMUM ITERATIONS FOR SEVERE DISCONTINUITIES 12
MAXIMUM CUT-BACKS ALLOWED IN AN INCREMENT 5
MAXIMUM DISCON. ITERS IN TWO INCREMENTS FOR TIME INCREMENT INCREASE 6
CUT-BACK FACTOR AFTER DIVERGENCE 0.250
CUT-BACK FACTOR FOR TOO SLOW CONVERGENCE 0.500
CUT-BACK FACTOR AFTER TOO MANY EQUILIBRIUM ITERATIONS 0.7500
CUT-BACK FACTOR AFTER TOO MANY SEVERE DISCONTINUITY ITERATIONS 0.2500
CUT-BACK FACTOR AFTER PROBLEMS IN ELEMENT ASSEMBLY 0.2500
INCREASE FACTOR AFTER TWO INCREMENTS THAT CONVERGE QUICKLY 1.500
MAX. TIME INCREMENT INCREASE FACTOR ALLOWED 1.500
MAX. TIME INCREMENT INCREASE FACTOR ALLOWED (DYNAMICS) 1.250
MAX. TIME INCREMENT INCREASE FACTOR ALLOWED (DIFFUSION) 2.000
MINIMUM TIME INCREMENT RATIO FOR EXTRAPOLATION TO OCCUR 0.1000
MAX. RATIO OF TIME INCREMENT TO STABILITY LIMIT 1.000
FRACTION OF STABILITY LIMIT FOR NEW TIME INCREMENT 0.9500

PRINT OF INCREMENT NUMBER, TIME, ETC., EVERY 1 INCREMENTS

RESTART FILE WILL BE WRITTEN EVERY 999 INCREMENTS

ONLY THE LATEST INCREMENT WILL BE RETAINED

THE MAXIMUM NUMBER OF INCREMENTS IN THIS STEP IS 200

LARGE DISPLACEMENT THEORY WILL BE USED

EXTRAPOLATION WILL BE USED

CHARACTERISTIC ELEMENT LENGTH 75.2

PRINT OF INCREMENT NUMBER, TIME, ETC., TO THE MESSAGE FILE EVERY 1 INCREMENTS

FILE OUTPUT WILL BE WRITTEN IN CARD IMAGE FORMAT

INCREMENT 1 STARTS. ATTEMPT NUMBER 1, TIME INCREMENT 1.00

EQUILIBRIUM ITERATION 1

LARGEST RESIDUAL FORCE 4.803E-03 AT NODE 7483 DOF 3
LARGEST INCREMENT OF DISP. -2.412E-03 AT NODE 4837 DOF 3
LARGEST CORRECTION TO DISP. -2.412E-03 AT NODE 4837 DOF 3
DISP. CORRECTION TOO LARGE COMPARED TO DISP. INCREMENT

EQUILIBRIUM ITERATION 2

LARGEST RESIDUAL FORCE 1.338E-08 AT NODE 7468 DOF 3
LARGEST INCREMENT OF DISP. -2.412E-03 AT NODE 4837 DOF 3
LARGEST CORRECTION TO DISP. 7.777E-08 AT NODE 4821 DOF 3

148
THE FORCE EQUILIBRIUM EQUATIONS HAVE CONVERGED

ITERATION SUMMARY FOR THE INCREMENT: 2 TOTAL ITERATIONS, OF WHICH 0 ARE SEVERE DISCONTINUITY ITERATIONS AND 2 ARE EQUILIBRIUM ITERATIONS.

TIME INCREMENT COMPLETED 1.00, FRACTION OF STEP COMPLETED 1.00
STEP TIME COMPLETED 1.00, TOTAL TIME COMPLETED 13.0

RESTART INFORMATION WRITTEN IN STEP 13 AFTER INCREMENT 1

THE ANALYSIS HAS BEEN COMPLETED

ANALYSIS SUMMARY:
TOTAL OF 13 INCREMENTS
0 CUTBACKS IN AUTOMATIC INCREMENTATION
25 ITERATIONS
25-passes through the equation solver of which 25 involve matrix decomposition, including
0 decomposition(s) of the mass matrix
0 additional residual evaluations for line searches
0 additional operator evaluations for line searches
0 warning messages during user input processing
0 warning messages during analysis
0 analysis warnings are numerical problem messages
0 analysis warnings are negative eigenvalue messages
0 error messages

THE SPARSE SOLVER HAS BEEN USED FOR THIS ANALYSIS.

JOB TIME SUMMARY
USER TIME (SEC) = 151.57
SYSTEM TIME (SEC) = 7.6900
TOTAL CPU TIME (SEC) = 159.26
WALLCLOCK TIME (SEC) = 310
Appendix V

INTELLISUITE SIMULATION OF FABRICATION OF THE FIRST MODULE OF THE ELECTROMAGNETIC MICROACTUATOR

Number of steps in Process: 51
1. Definition FeSi Si3pc 100
   Operation on top side.
   t_film: 50000 nm
   flat_dir: 100 Vector
diameter: 101.6 mm
   resist: 4.8e-05 Ohm-cm

2. Etch FeSi Clean Piranha
   Operation on top side.
   H2SO4_conc: 75 %
   H2O2_conc: 25 %
time_etch: 10 min

3. Deposition Ti Sputter Ar-Ambient
   Operation on top side.
   Rf_Pow: 250 W
   Rf_Freq: 13.56 MHz
   P_dep: 0.1 Pa
time_dep: 10 min
Results:
   t_film: 100 nm

4. Deposition BCB Spin Standard
   Operation on top side.
   Speed: 4000 rpm
time_spin: 30 sec
   T_soft: 150 deg C
time_soft: 60 sec
   lambda: 436 nm
   T_hard: 230 deg C
Results:
   t_film: 10000 nm

5. Deposition PR-S1800 Spin S1805
   Operation on top side.
   Speed: 4000 rpm
time_spin: 30 sec
   T_soft: 115 deg C
time_soft: 60 sec
   lambda: 436 nm
Results:
  t_film: 500 nm

6. Definition UV Contact Suss
   Operation on top side.
   mask_no: 10 #
   Power: 250 W
   lambda: 436 nm
   time_exp: 10 sec
Results:
   dose: 52 J

7. Etch PR-S1800 Wet 1112A
   Operation on both sides.
   T_etch: 20 deg_C
   time_etch: 5 min
Results:
   t_etch: 990000 nm

8. Etch BCB Plasma 02
   Operation on top side.
   Rf_Pow: 250 W
   Rf_Freq: 13.59 MHz
   time_etch: 10 min
   T_etch: 50 deg_C
   P_etch: 30 Pa
Results:
   t_etch: 10000 nm

9. Etch PR-S1800 Wet 1165
   Operation on both sides.
   T_etch: 20 deg_C
   time_etch: 5 min
Results:
   t_etch: 500 nm

10. Deposition Au E-Beam E-Beam
    Operation on top side.
    Rf_Pow: 100 W
    Rf_Freq: 13.56 MHz
    T_dep: 100 deg_C
    P_vac: 10 uTorr
    time_dep: 10 min
Results:
    t_film: 75 nm

11. Deposition SU-8 SPIN SPIN
    Operation on top side.
    T_soft: 75 deg_C
    time_hard: 15 sec
    Speed: 1100 rpm
    T_hard: 95 deg_C
    time_soft: 3 sec
Results:
    t_film: 10000 nm

12. Definition UV Contact Suss
Operation on top side.
  mask_no: 11 #
  Power: 250 W
  lambda: 436 nm
  time_exp: 10 sec
Results:
  dose: 52 J

13. Etch SU-8 PGMEA etch
   Operation on top side.
   time_etch: 6 min
Results:
  t_etch: 990000 nm

14. Deposition Cu Electroplat General
   Operation on top side.
   area: 1 cm²
   curr_dens: 0.01 amp/cm²
   efficiency: 92 %
   time_dep: 10000 min
   T_dep: 60 deg_C
Results:
  t_film: 10000 nm

15. Etch Cu Dry SF6-Plasma
   Operation on top side.
   P_base: 0.0001 Pa
   time_etch: 15 min
Results:
  t_etch: 10000 nm

16. Etch SU-8 PGMEA etch
   Operation on top side.
   time_etch: 6 min
Results:
  t_etch: 10000 nm

17. Etch Au Wet PAN
   Operation on top side.
   H2PO4_conc: 75 %
   C2H4O2_conc: 15 %
   HNO3_conc: 5 %
   time_etch: 20 min
Results:
  t_etch: 990000 nm

18. Deposition BCB Spin Standard
   Operation on top side.
   Speed: 4000 rpm
   time_spin: 30 sec
   T_soft: 150 deg_C
   time_soft: 60 sec
   lambda: 436 nm
   T_hard: 230 deg_C
Results:
  t_film: 10000 nm
19. Deposition PR-S1800 Spin S1805
   Operation on top side.
   Speed: 4000 rpm
   time_spin: 30 sec
   T_soft: 115 deg_C
   time_soft: 60 sec
   lambda: 436 nm
Results:
   t_film: 500 nm

20. Definition UV Contact Suss
   Operation on top side.
   mask_no: 12 #
   Power: 250 W
   lambda: 436 nm
   time_exp: 10 sec
Results:
   dose: 52 J

21. Etch PR-S1800 Wet 1112A
   Operation on both sides.
   T_etch: 20 deg_C
   time_etch: 5 min
Results:
   t_etch: 990000 nm

22. Etch BCB Plasma 02
   Operation on top side.
   Rf_Pow: 250 W
   Rf_Freq: 13.59 MHz
   time_etch: 10 min
   T_etch: 50 deg_C
   P_etch: 30 Pa
Results:
   t_etch: 10000 nm

23. Etch PR-S1800 Wet 1165
   Operation on both sides.
   T_etch: 20 deg_C
   time_etch: 5 min
Results:
   t_etch: 500 nm

24. Deposition BCB Spin Standard
   Operation on top side.
   Speed: 4000 rpm
   time_spin: 30 sec
   T_soft: 150 deg_C
   time_soft: 60 sec
   lambda: 436 nm
   T_hard: 230 deg_C
Results:
   t_film: 10000 nm

25. Deposition PR-S1800 Spin S1805
   Operation on top side.
   Speed: 4000 rpm
time_spin: 30 sec
t_soft: 115 deg_C
time_soft: 60 sec
lambda: 436 nm

Results:
t_film: 500 nm

26. Definition UV Contact Suss
Operation on top side.
mask_no: 14 #
Power: 250 W
lambda: 436 nm
time_exp: 10 sec

Results:
dose: 52 J

27. Etch PR-S1800 Wet 1112A
Operation on both sides.
T_etch: 20 deg_C
time_etch: 5 min

Results:
t_etch: 990000 nm

28. Etch BCB Plasma 02
Operation on top side.
Rf_Pow: 250 W
Rf_Freq: 13.59 MHz
time_etch: 10 min
T_etch: 50 deg_C
P_etch: 30 Pa

Results:
t_etch: 990000 nm

29. Etch PR-S1800 Wet 1165
Operation on both sides.
T_etch: 20 deg_C
time_etch: 5 min

Results:
t_etch: 500 nm

30. Deposition Au E-Beam E-Beam
Operation on top side.
Rf_Pow: 100 W
Rf_Freq: 13.56 MHz
T_dep: 100 deg_C
P_vac: 10 uTorr
time_dep: 10 min

Results:
t_film: 75 nm

31. Deposition PR-S1800 Spin S1805
Operation on top side.
Speed: 4000 rpm
time_spin: 30 sec
T_soft: 115 deg_C
time_soft: 60 sec
lambda: 436 nm
Results:
  t_film: 500 nm

32. Definition UV Contact Suss
   Operation on top side.
   mask_no: 15 #
   Power: 250 W
   lambda: 436 nm
   time_exp: 10 sec
Results:
  dose: 52 J

33. Etch PR-S1800 Wet 1112A
   Operation on both sides.
   T_etch: 20 deg_C
   time_etch: 5 min
Results:
  t_etch: 990000 nm

34. Etch Au Wet PAN
   Operation on top side.
   H2PO4_conc: 75 %
   C2H4O2_cnc: 15 %
   HNO3_conc: 5 %
   time_etch: 20 min
Results:
  t_etch: 75 nm

35. Etch PR-S1800 Wet 1165
   Operation on both sides.
   T_etch: 20 deg_C
   time_etch: 5 min
Results:
  t_etch: 500 nm

36. Deposition Cu Electroplat General
   Operation on top side.
   area: 1 cm2
   curr_dens: 0.01 amp/cm2
   efficiency: 92 %
   time_dep: 5000 min
   T_dep: 60 deg_C
Results:
  t_film: 10000 nm

37. Etch Cu Dry SF6-Plasma
   Operation on top side.
   P_base: 0.0001 Pa
   time_etch: 15 min
Results:
  t_etch: 10075 nm

38. Deposition PR-S1800 Spin S1805
   Operation on top side.
   Speed: 4000 rpm
   time_spin: 30 sec
   T_soft: 115 deg_C
time_soft: 60 sec  
lambda: 436 nm  

Results:  
t_film: 500 nm

39. Definition UV Contact Suss  
Operation on top side.  
mask_no: 16 #  
Power: 250 W  
lambda: 436 nm  
time_exp: 10 sec  

Results:  
dose: 52 J

40. Etch PR-S1800 Wet 1112A  
Operation on both sides.  
T_etch: 20 deg_C  
time_etch: 5 min  

Results:  
t_etch: 990000 nm

41. Etch BCB Plasma 02  
Operation on top side.  
Rf_Pow: 250 W  
Rf_Freq: 13.59 MHz  
time_etch: 10 min  
T_etch: 50 deg_C  
P_etch: 30 Pa  

Results:  
t_etch: 990000 nm

42. Etch BCB Plasma 02  
Operation on top side.  
Rf_Pow: 250 W  
Rf_Freq: 13.59 MHz  
time_etch: 10 min  
T_etch: 50 deg_C  
P_etch: 30 Pa  

Results:  
t_etch: 990000 nm

43. Etch BCB Plasma 02  
Operation on top side.  
Rf_Pow: 250 W  
Rf_Freq: 13.59 MHz  
time_etch: 10 min  
T_etch: 50 deg_C  
P_etch: 30 Pa  

Results:  
t_etch: 990000 nm

44. Deposition NiFe Electroplat Standard  
Operation on top side.  
T_depl: 60 deg_C  
time_depl: 10000 min  
efficiency: 92 %  
curr_dens: 0.01 amp/cm2
area: 1 cm²

Results:

\[ t_{\text{film}}: 40000 \text{ nm} \]

45. Etch PR-S1800 Wet 1165
Operation on both sides.
\[ T_{\text{etch}}: 20 \text{ deg_C} \]
\[ \text{time}_\text{etch}: 5 \text{ min} \]

Results:

\[ t_{\text{etch}}: 500 \text{ nm} \]

46. Deposition alpha_Si PECVD Standard
Operation on top side.
\[ T_{\text{dep}}: 250 \text{ deg_C} \]
\[ P_{\text{dep}}: 9.999 \text{ Pa} \]
\[ \text{time}_\text{dep}: 3 \text{ min} \]

Results:

\[ t_{\text{film}}: 50 \text{ nm} \]

47. Deposition PR-S1800 Spin S1805
Operation on top side.
\[ \text{Speed}: 4000 \text{ rpm} \]
\[ \text{time}_\text{spin}: 30 \text{ sec} \]
\[ T_{\text{soft}}: 115 \text{ deg_C} \]
\[ \text{time}_\text{soft}: 60 \text{ sec} \]
\[ \lambda: 436 \text{ nm} \]

Results:

\[ t_{\text{film}}: 500 \text{ nm} \]

48. Definition UV Contact Suss
Operation on top side.
\[ \text{mask_no}: 18 \# \]
\[ \text{Power}: 250 \text{ W} \]
\[ \lambda: 436 \text{ nm} \]
\[ \text{time}_\text{exp}: 10 \text{ sec} \]

Results:

\[ \text{dose}: 52 \text{ J} \]

49. Etch PR-S1800 Wet 1112A
Operation on both sides.
\[ T_{\text{etch}}: 20 \text{ deg_C} \]
\[ \text{time}_\text{etch}: 5 \text{ min} \]

Results:

\[ t_{\text{etch}}: 990000 \text{ nm} \]

50. Etch alpha_Si RIE SF6/02
Operation on top side.
\[ \text{time}_\text{etch}: 3 \text{ min} \]

Results:

\[ t_{\text{etch}}: 50 \text{ nm} \]

51. Etch PR-S1800 Wet 1165
Operation on both sides.
\[ T_{\text{etch}}: 20 \text{ deg_C} \]
\[ \text{time}_\text{etch}: 5 \text{ min} \]

Results:

\[ t_{\text{etch}}: 500 \text{ nm} \]
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