INDIVIDUAL AIDS FOR THE AURALLY HANDICAPPED

CERTIFICATE

This is to certify that the independent Project entitled

"INDIVIDUAL AIDS FOR THE AURALLY HANDICAPPED"

is the bonafide work in part fulfilment for M.Sc, I year Speech and Hearing of the student with Reg. No.5

Director
All India Institute of Speech and Hearing
Mysore

CERTIFICATE

This is to certify that this independent Project has been prepared under my guidance.

GUIDE

DECLARATION

This independent Project is the result of my study undertaken under the guidance of Dr. (Miss) SHAILAJA NIKAM, Professor and Head, Department of Audiology, All India Institute of Speech and Hearing, Mysore-570 006, and has mot been submitted at any University for any other Diploma or Degree.

ACKNOWLEDGEMENT

I am grateful to Dr. (Miss) SHAILAJA NIKAM, Professor and Head, Department of Audiology, All India Institute of Speech and Hearing for her guidance and co-operation in completion of this Project.

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PRE-ELECTRIC AND NON-ELECTRIC HEARING AIDS

CHAPTER I

PRE-ELECTRIC AND NON-ELECTRIC HEARING AIDS

Hearing Aid literally means, a device capable of 'Aiding the hearing'. The main purpose of the hearing aid is to amplify the sound and enable it to reach the ear effectively. There are no records of earliest attempts through which man aided his hearing. But someone must have found out that hand cupped behind the pinna made the weaker sounds louder by deflecting the more sound energy into the external canal and at the same time cautioned the speaker to raise the volume of his voice.

We might speculate about the forms which first man made hearing aid took. These could be:

i. Hollowed out animal horn
ii. Broken sea-shell
iii. A large leaf rolled into a tube
iv. Length of a cane.

The earliest aids were termed as deaf aids, deaf instruments, speaking tubes and hearing tubes. It is difficult to determine about the firms which manufactured hearing aids. Only few pioneer firms can be indicated which were situated in England, France, United States of America and West Germany.

Ear Trumpets and Cornets

The ear trumpet certainly is one of the earliest hearing aids which were first fashioned from the horns of cows or rams suitably hollowed. Later wooden, metals and then shell horn followed the same. Contours as the animal horns had. Figure 1 shows an Ear Trumpet.

Several centuries before Christ, the Greeks bought a large variety of shells. These were hardened and painted and used as acoustical cornets (Berger, 1970).

These devices collect and direct the sound into the external auditory canal. The amplification is of primitive form. It required no other expenditure than buying of the instrument. Archigenes suggested that the deaf be trained by using an ear trumpet and employing loud sounds. (Goldstein, 1933)

A 17th century engraving is the oldest known graphical representation of a hearing device, which is a spherical tube, like a huge football of about two meters in length and at least 50 Cm, in diameter at its mid point. The central portion was a resonator. The instrument rested on a socket like axis. This permitted the tube to turn and orient one of the ends towards the mouth of the speaker. The user placed himself so that his ear was next to the opening at the other end of the tube.

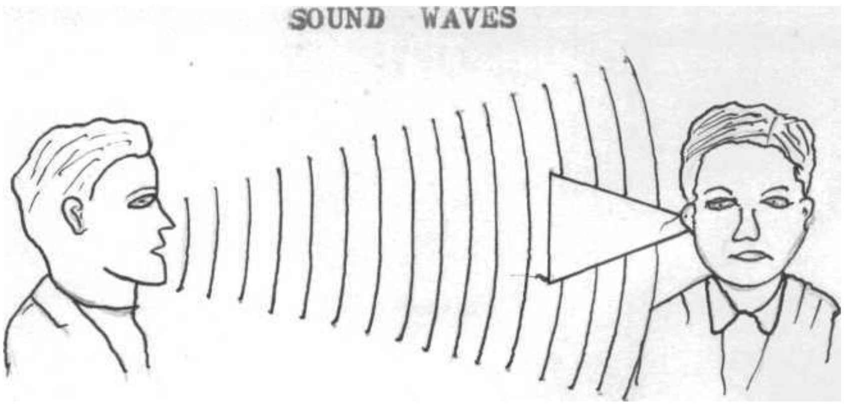

Fig.l: Ear Trumpet Collects Sound Energy, But has no Batteries or Amplifier.

Ear trumpets were varied in size with an intention to conceal or else make these serve several functions. These were made of vulcanite or imitations of tortoise shell. Other models were referred as banje-style since they somewhat resembled that musical instrument. The disc or bowel shape attachment (4-6" in diameter) at the speaker end was planned to increase amplification. One manufacturer called it 'Plantagenet' while other advertised as 'Green's Hearing Horn'. Length ranged from 14-17" open and 8-12" when collapsed. The various forms of ear trumpets are shown in Figures 1-1, 1-2 and 1-3 depicts the amplification of Bonjo styled Trumpet and Ear Trumpet (Non-Electrical) respectively.

The disadvantage of the ear trumpets and the cornets were that the user had to hold them with one hand while in use. This inconvenience was overcome in another model by mounting of scoop-like sound bells on a head band with a small ear-tip or circular ear cushion opening at the ear. These were called Auricles and other firm made it known to the public as Dupliphones.

The cornets seems to have taken their origin from sea shells, and the trumpets from animal horns. The speaking trumpets were usually designed to facilitate communication between naval ships or military units. Homer (Circa 850 B.C.) mentioned the large bronze trumpet that projected a voice over considerable distance.

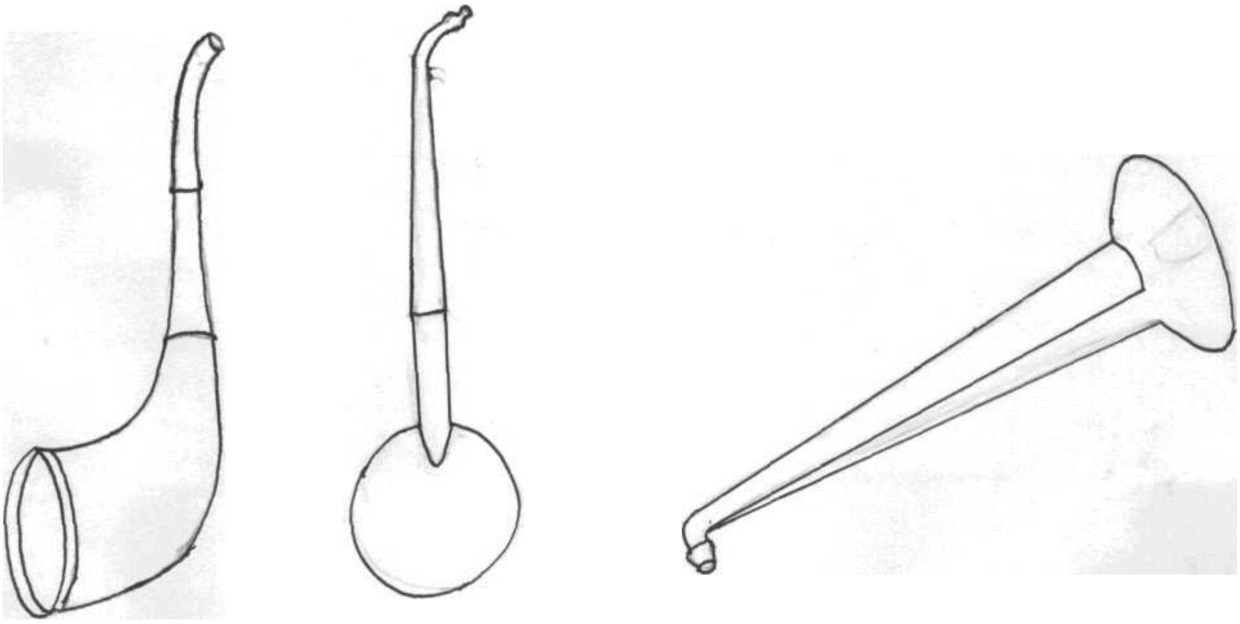

Fig.1-1: Shows Several Ear Trumpets

Fig.1-2: Amplification of Banjo Styled Trumpet.

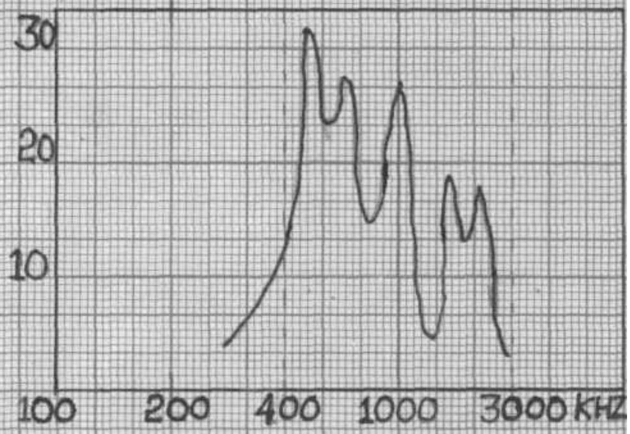

Fig.1-3: Ear Trumpet (Non-Electrical)

Kircher (1601-1680) a German scholar, mathematician and philosopher claims to have invented the speaking trumpet. He seems to have developed hearing trumpet in about 1649 while he was at Jesuit's College in Rome.

Lambert (1736) was probably the first man to put forth a theory as to how speaking trumpet works, but mistakenly based his explanation on light*

Speaking Tubes

The speaking tube consisted of funnel and tip ends which were made of metal or hard rubber, with the tube itself made of rubber or of spring wire covered with fabric. The speaking tube picked up the sound close to its source and directed the sound by limited pathway to the ear. The speaker would talk directly into the funnel end of the tube and the listener would hold the smaller tip of the tube at his ear. Here the advantage is similar to that of the speaker talking very close to the listener's ear. The amplification of speaking tube is illustrated in Figure 1-4.

Acoustic Cane

The Acoustic cane became popular around the first decade of 20th century* Its principle is same as that of speaking tube. The cane was not flexible but the deaf could use this as a cane and also when needed, for the purpose of

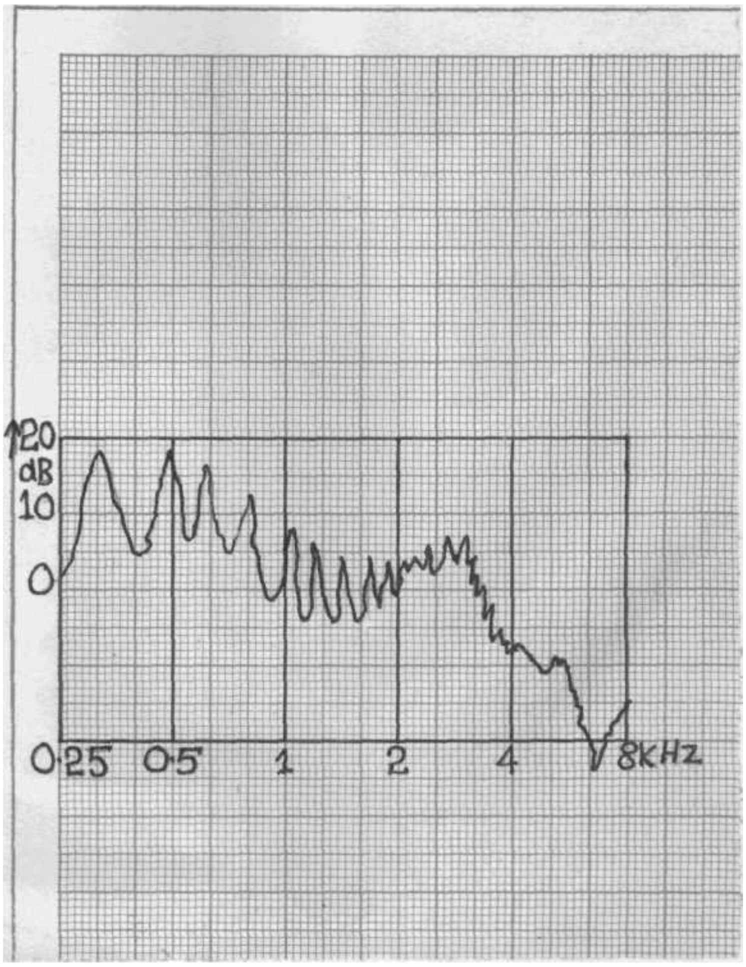

Fig.1-4: Amplificaton of Speaking Tube.

hearing. The curved handle was kept near the mouth of the speaker, while the other end was held at the ear of the deaf individual.

Pinna Inserts

These date back to about 1830. This was a sort of device which fit into the external canal. They have very limited use in mild hearing loss cases. It might be possible that some individuals found them useful in cases of collapsed ear canal.

Artificial Ear Drum

An artificial ear drum consisting of a small tube covered by a piece of pig's bladder was used by Marcus Banzer as early as 1640 (Goldstein, 1933). An early model of such device is shown in Figure 1-5.

The artificial ear drum was of use only to a person who had a large perforation of the ear drum and even then the help it afforded was doubtful. It required considerable skill to position the device and its aim was to serve as an artificial ear drum.

Devices similar to the artificial ear drum are still used today. An artificial ear drum, made out of a small piece of fish skin, connected to a bristle, can be brought into contact with the stapes through the perforation in the

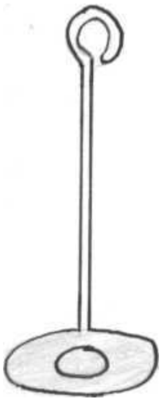

Fig.i-5: Artificial Eardrum

ear drum. Paper patches are also applied, but it is recommended they be inserted by the otolaryngologist under operating microscope. Some Indian Cigrarette Companies are manufacturing these papers.

A few firms in Europe still manufacture a limited number of non-electric aids. Some elderly individuals may find the non-electric devices easier to use than the latest miniature electronic hearing aids.

Bone Conduction Appliances

The concept of bone conduction hearing is old and was known at the time of Greek Physician and writer Claudius Galen (131-200 A.D.). Here the sound is applied directly or indirectly to the skull. Itard (?-l832) an otologist at the Paris Home for the Deaf-Mutes developed a red which the speaker held at one end between his teeth and the listener the other end between his teeth. Bulwar (1614-1684) also made use of bone conducted stimulation in the construction of auditory prosthesis.

Previously Haller (1757) reported his investigation regarding bone conduction in Elementa Physiologica Corporis Humani. It is reported that Beethoven (1770-1827), the music composer, after becoming deaf, used a stick with one end between his teeth and the other touching piano, to listen to his music (Berger, 1970).

Buchner (1770) related that deaf could hear by means of "long" thin slips of wood, as thick as the back of knife one end to be applied to the upper teeth of the speaker, and other to the upper teeth of deaf person. A conversation tube to be used similarily was invented by Painter (1828).

Dentaphone

The teeth provide more efficient bone conduction route for sound conduction than that via the skull particularly for low frequencies (Berger, 1970). Dentaphone utilizes the concept of bone conducted sound via teeth. Dentaphone consisted of a round flat case with a thin diaphragm in the shape of the flat cone at the front centre of the case. The user held the case in his hand and a small wooden piece in his mouth was gripped tightly with the teeth. The diaphragm picked up the sound and in its centre was a knot from a piece of silk covered wire which was held tightly so as to pass vibrations to the user's teeth, in this way via bone conduction to the inner ear. A drawing of this device is shown in Figure 1-6.

Osteophone

Osteophone was similar in construction to Dentaphone. Charles Herman patented it in 1879. The Osteophone consisted of a large diaphragm held at body, but with unattached rod curved so as to reach the teeth (Goldstein, 1933).

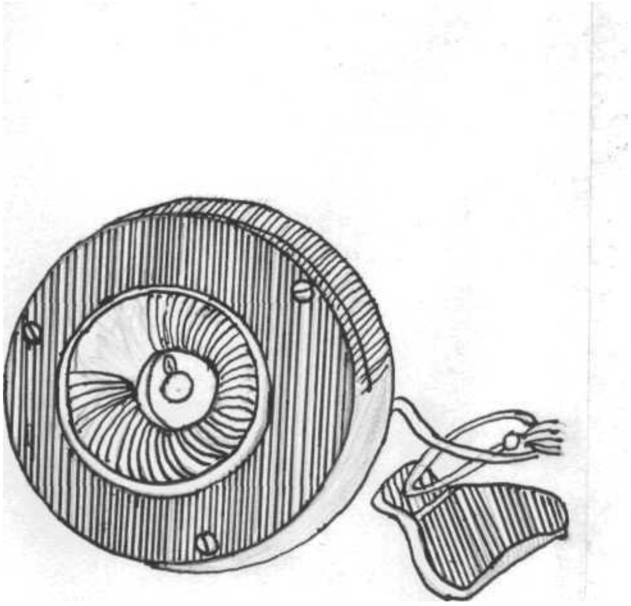

Fig.1-6: DENTAPHONE

The acoustic heard, acoustic bonnet and acoustic fan appeared late in 19th century. These inventions were designed to hide the hearing device. The acoustic fan was prepared from a sheet of vulconite with a handle and shape of fan. It is shown in Figure 1-7.

Audiophone

The hearing fan as invented by Rhodes (1879) may have been the first ane to be referred to as "audiophone" to describe a hearing aid. Later it became common to refer to the pre-electric hearing aids as audiophones.

Audiophone was made of composition material. The individual held the top edge of the fan against upper incisors as illustrated in Figure 1-8. The fan area collected the sound, then it travelled via the upper teeth to the inner ear by bone conduction. A series of cord which connected the upper edge of the fan blade with the handle, kept the large area of fan under tension and provided better vibration.

Approximate amplification characteristics which were limited around 500 Hz are shown in Figure 1-9.

Audinet

Another bone conduction fan, date unknown, also used the system of collected sound, via teeth and by bone

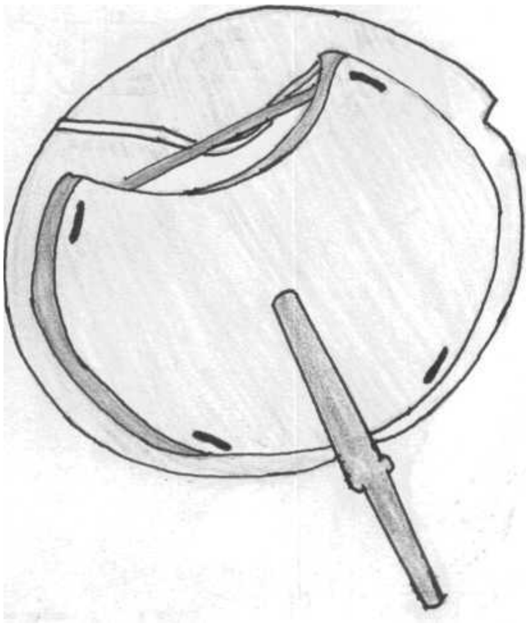

Fig.l-%: Acoustic Fan

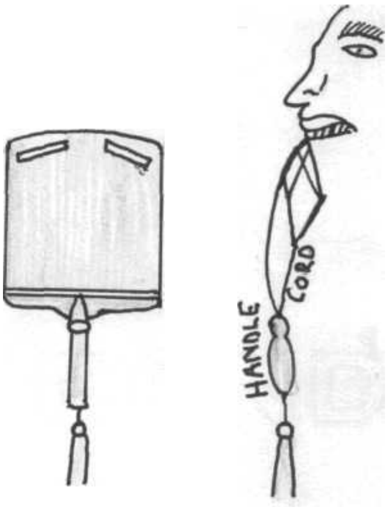

Fig.1-8: Rhode's Audiphone

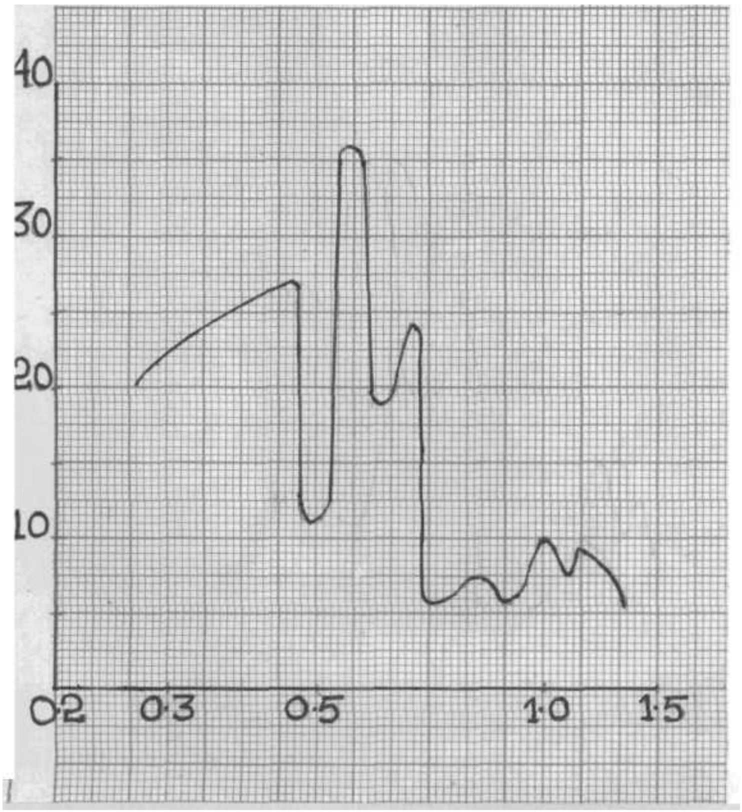

Fig.1-9: Amplification of
Rhode's Audiphone

conduction to inner ear. This fan seams to be modification of the Audiphone and was called Audinet. It was made up of two large thin discs of vulcanite, a somewhat smaller disc on the top and slightly separated from the largest disc. The handle consisted of solid vulconite rod. The process of sound collection and transmission in the Audinet was presumably more efficient than that of the Audiphone.

Other Non-electric Hearing Devices

Various other forms of acoustic fan design for air conduction were also available. The fan was built of a metal. It resembled the shape of a partial folded fan. Its handle portion was held behind the ear. The folded fan deflected and directed the sound into the ear. Its another version consisted of small trumpet hidden in the partially folded fan.

Several other instruments do not come in any of the above categories. These were especially made for the rich and royalty. A chair was constructed with a conoe-shaped back top which kept the head in a "prow" position and allowed the sound to be focussed towards ears.

Many members of the royal family had unique devices designed for their use. These consisted of special chairs or thrones which had trumpet bells built into the arm sets and with tubes leading to the ears, Duguet is considered to be the inventor of the acoustic chair in 1706.

The firm Rein & Son in 1819 made acoustic throne for King Goa VI of Portugal. Each arm rest had openings carved in the shape of lion heads. Courtriers knelt and spoke into the mouthpieces in the arms. The resonated sound was transmitted to an ear piece which was fitted on a tube. Here the flexibility of the user or speaker was restricted.

There was also an attempt by Vainwright in 1896 to hide the hearing aid by attaching a binaural ear trumpet to a pair of binoculars. The barrels of the binocular acted as a resonator to amplify a sound which was routed to ears. Sargent (1918) provided for a funnel-like sound catcher to be mounted on each eye-glass temple and from here the sound was directed towards the ear.

Ear Exercisers

The earliest ear excercisers were hand driven and later on it was possible to operate them on battery or alternating current. But such devices were regarded useless and people were discouraged from buying them.

Previously it was thought that deafness was caused by problems of inertia of the ossicles or a similar barrier to the sound transmission, hence the idea of ear exercisers came from the above belief (Berger, 1970).

In addition to the ear vibrators or massagers there was an assortment of oils, medicated ear pieces, electric belts, concentrated heat lamps and special devices that spultered sounds or flashed colour lights to 'restore hearing absolutely'.

CHAPTER 2

THE ELECTRIC HEARING AID

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THR ELECTRIC HEARING AID

The hearing trumpet and speaking tubes were certainly better than no hearing aid at all. But these were clumsy to handle and their amplifying capabilities were limited. The carbon hearing aid amplified the sound and it was a definite advance over the non-electric hearing instruments. In 1902, first commercially made hearing aid was introduced based on the invention of Alexander Graham Bell.

Bell (1872) devised a simple receiver transmitter and battery system for his mother (Newhart and Hartig, 1938). Bell who married Mabel Hubbard in 1877, attempted to make an amplification device for her, as she was deafened at the age of four years. He did not succeed around 1872-73, but his ideas were implemented in the development of a telephone. The earlier devices were named telephone aids or telemicrophone aids and later carbon aids. Jackson (1945) credited Dr. Ferdinand Alt with producing the first carbon device in 1900.

Hutchison (1902) first made the commercial carbon hearing aid. The earlier carbon hearing aids employed a microphone, an earphone, and were powered by battery. The carbon aids, gave amplification primarily near the low end and middle end of speech.

The Carbon Microphone

The microphone converts acoustical energy into - electrical energy. The change in electrical energy corresponds to the strength of the stimuli. This acoustical energy can be in the form of speech, music or noise. This electrical energy is converted back to sound waves with the help of receiver.

The carbon microphone is dependent for its operation on the variation in the resistance of a small enclosure filled with carbon gronules. When the sound waves strike the diaphragm, these sound waves cause the carbon gronules in the cup to alternately be compressed or be decreased to a slight compression. A cross section diagram of carbon microphone is shown in Figure 2. If battery is connected to one side of the microphone and otherside to the receiver, the small changes in resistance will end up amplified in the earphone.

The battery voltage mostly depends upon the size of the battery which limits the amplifying characteristics of particular circuit. The range and magnitude of the frequency response of the carbon microphone is also determined by the ability of carbon granules to change their density.

The peak resonance which occurs near 2000 Hz is associated with the fundamental resonant frequency of the

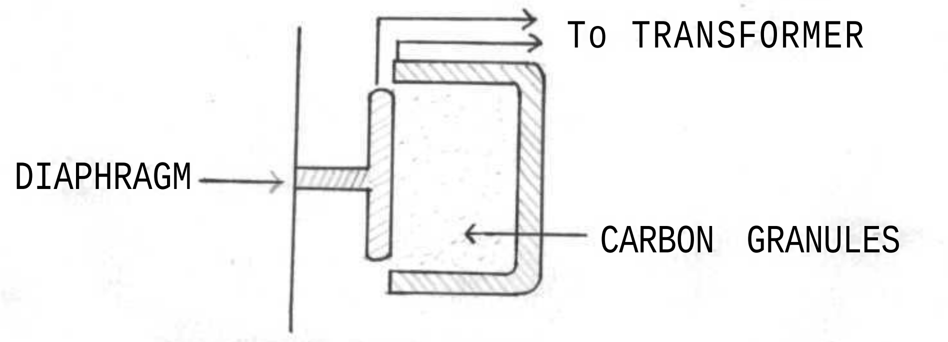

Fig. 2: Carbon Micrphone

diaphragm. The resonance of the microphone causes the microphone to be non-linear and is usually a disadvantage. The use of stretched diaphragm makes it possible to extend the region of relatively uniform response to about 8000 Hz, but only at the expense of decreased sensitivity.

The other undesirable characteristic of the microphone iS that once the diaphragm has been set in motion by sound energy, it tends to stay in motion. This produces distortion which can be overcome by adding soft material to the diaphragm which will dampen this motion.

The carbon microphones are not sensitive to all the frequencies. This adversely affects the quality of the voice reproduced. The amplification can also be increased by employing a large number of microphones, but this makes the hearing aid too bulky.

Increase in amplification can also be sought by increasing the battery voltage. But high voltage leads to more noise and also shortens the life of carbon materials due to burning. For best results, the carbon needed renewing about once a year and even more frequently if used everyday or exposed to a lot of dust.

The Magnetic Earphone

The earphone changes the amplified electrical signal into acoustical energy. In the magnetic earphone, soft iron diaphragm is held by the external case enclosure so that it allows for a small air gap between the diaphragm and the soft iron pole pieces which join the permanent magnet. When the signal current varies, it alternatively pulls the diaphragm towards pole pieces or allow the diaphragm to return to its normal position.

As the current flows through the coils on the pole pieces of the magnet, it develops electromagnetic force. The incoming current varies with the signal, it sets inward and outward motion in the diaphragm. This in turn generates audible sound waves. A diagram of simple magnetic earphone is shown in Figure 2-1.

Bone Conduction Vibrator

The sound can by pass the external and middle ear, in bone conduction. Here the diaphragm is connected to a metal or plastic plate, or it moves a small plunger.

In 1920, large bone conduction vibrators were introduced with some audiometers and a few table model hearing aids. Leiber (1932) developed a wearable bone conductor vibrator for the carbon hearing aid, and it was marketed by Sonotone. Leiber also invented other popular bone conduction vibrators. Since the bottom part of the vibrator is placed in contact with the head, the energy

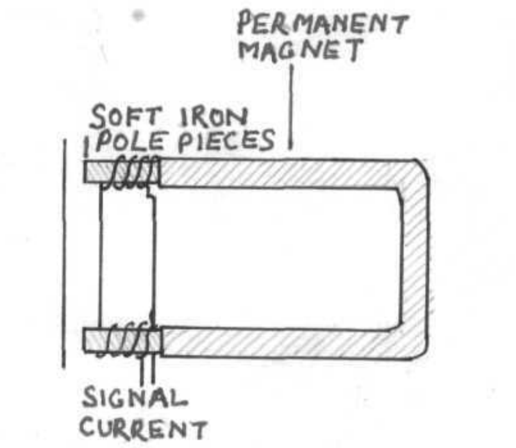

Fig.2-1: Magnetic Earphone

from the vibrator is transferred to the skull, and on to the inner ear (Leiber, 1932).

The bone conduction vibrator is usually more bulky and tends to be a poor reproducer of high frequency sounds, (Davis and Silverman, i960). It also requires more energy than the air-conduction receiver. It is particularly useful in those cases where the external ear canal is absent or deformed, where theSe is large air bone gap or chronic discharge (Berger and Millin, 1970).

Fiske (1880) patented one bone conduction device. This dental attachment had a hinged flat mouth piece, which was attached to a diaphragm device. The attachment was to be connected directly to the telephone, which enabled the user to hear by bone conduction. Schvenche patented another design in 1939. These designs called for a bone conduction receiver in smoking pipe, which was to be powered by a carbon hearing aid.

In most respects, carbon hearing aid was a great improvement over the non-electric instruments. Problems consisted of distortion, resulting from resonance in the microphone and receiver, and difficulties due to noise inherent in the microphone. Later, however there were improvements in the carbon aid which included an amplifier and allowed for some indirect amplification of the signal.

Boosters were introduced about 1924, first in Germany, the design was similar to that of carbon microphone.

The carbon hearing aid proved to be quite practical and was not easily damaged. Though it did not provide sufficient gain, the instrument performed well and the internal noise was bearable. A typical acoustic response of carbon aid is shown in the Figure 2-2.

Damp weather and body movements also interfered with the efficiency of microphone. Battery connections of the carbon aid were to be kept clean and tight, using batteries alternatively to give them time to recuperate. It was also necessary to have the microphone clean and dry and moisture from the earphone was to be avoided.

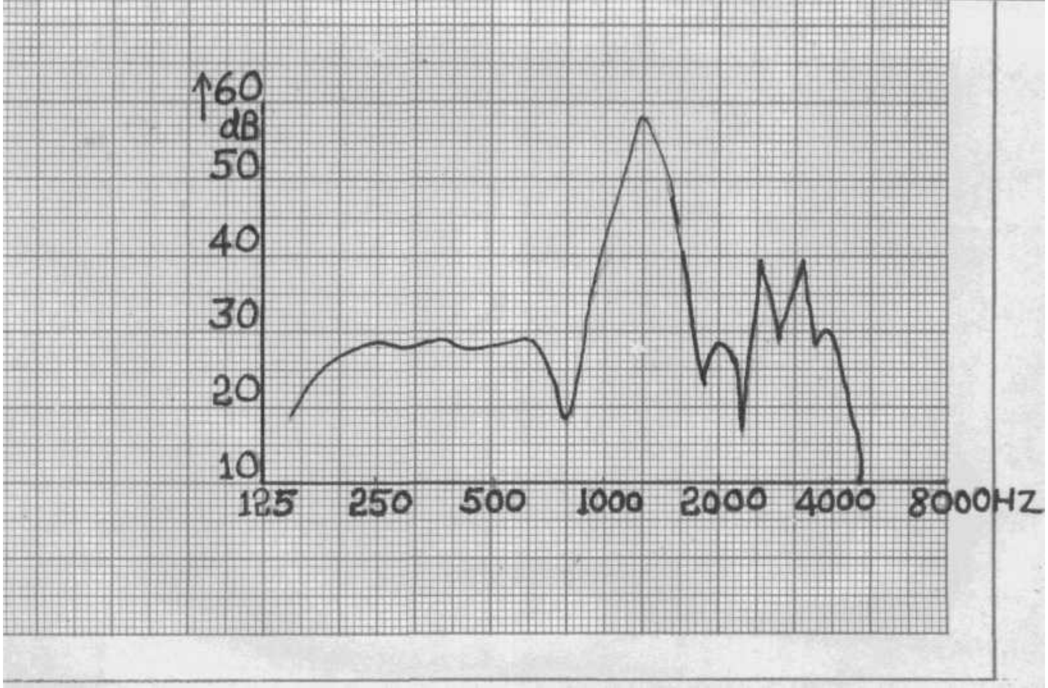

Fig.2-2: Carbon Aid Response.

CHAPTER 3

THE ELECTRONIC HEARING AID

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THE ELECTRONIC HEARING AID

Fleming (1904) discovered the principle of diode. Diode is the simplest combination of elements which constitute an electron tube. It consists of a cathode which serves as an emitter of electrons, and a plate or anode serving as a collector of electrons. Both electrodes are placed in a highly evacuated envelope of glass or metal, to avoid the contamination of cathode emitting surface and to allow free motion of electrons without the collision with air particles. Any air still remaining inside is cancelled by placing a geter compound in the envelope.

The diode can only conduct current from cathode to plate, and this property makes the diode a useful device. For the operation of diode two supply voltage are needed, one is applied to heater or filament circuit and the other to the plate cathode circuit.

Triode Vacuum Tube

In i906, De Forest added a third element, the control grid between cathode and plate of a diode, as it involved three elements it was called triode. He called it "audion". The control grid enables the tube to amplify by controlling the flow of electrons.

Tetrodes and Pentodes

The development of the tetrode took place due to undesirable grid-to-plate capacitance in triodes. The four element vacaum tube, containing two grids Is called tetrode. A vacuum tube in which three of five elements are grids is called a pentode which was developed in 1931. These additional grids, called the suppression grid and screen grid are placed between the control grid and the plate of vacuum tube. Hearing aids usually employed three or four pentodes and sometimes one or two diodes. Amplification ability of vacuum tubes in a circuit is cumulative. The actual performance in a vacuum tube hearing aid was much higher than this (Schachtel, 1948).

OTHER CIRCUIT COMPONENTS

The Microphone

The crystal microphone was most commonly used with hearing aids (Berger, 1970). It makes use of piezo electric crystal or dielectrics as the transducing elements. Rochelle salt, Ammonium dihydrogen phosphate (A.D.P.) and the lithium sulphate crystals are used today. These crystals or dielectrics become electrical polarized and produce voltages linearly related to mechanical forces, upon being distorted by the action of incident sound waves. Rochelle salt is widely employed in the design of crystal microphones as it has the strongest piezo-electric effect. Brother Curie (1880) discovered the direct piezo-electric effect.

The crystal microphone is housed in a thin metal cup. Tinfoil electrodes are on each side of the crystal and from the electrodes, wire leads to the amplifier portion of the hearing aid. The physical size of an crystal governs the range of frequencies to which it will respond. The ceranic microphone has piezo-electric characteristics. It has advantage over the crystal and magnetic microphone that includes having excellent low frequency characteristics and may operate in high temperature and humidity.

Receiver

The function of the receiver is to accept amplified power from the hearing aid. The receiver for vacuum tube hearing aid was either of the air conduction or bone conduction type. The air conduction receiver was based on piezo-electric effect, but utilized the reverse operation property of the crystal i.e., when electrical impulses are passed through it, it will vibrate mechanically.

The bone conduction receiver for the vacuum tube aid was of magnetic type. The vacuum tube hearing aids were designed both for air as well as bone conduction receivers, due to some technical problems, they had a crystal magnetic switch, and separate cord connections for selection of the receiver to be used (Berger, 1970).

Volume Control

It became more easy to control the gain of vacuum Tube Hearing Aid than was possible for the carbon aid. The volume control is a variable voltage divider which permits selection of greater or lesser output of a preceding stage to be applied to the next stage in amplifier. Gain is also less correctly called volume. There are three types of volume control which are provided in vacuum tube and transistor circuits.

1. A Potentiometer

The volume control was usually provided in the form of external dial, which made it possible for the user to have finer adjustments. The volume control adjusts the difference between the size of electrical signal going to the amplifier and electrical signal coming out of the amplifier (Viconeth, 1967).

2. Tone Control

Tone control may be considered as a "frequency selective volume control". It selects the frequency range to be boosted. There are three positions for the adjustment of this type of control in hearing aids, for high, middle or low frequency emphasis. In the electronic design of this circuit, high frequency emphasis is actually an attenuation of low frequency and vice-versa. Most of the Indian hearing aids are provided with three tone control positions, high, normal and low. But one manufacturer has introduced, normal, $high_1$.$high_2$ and low tone control in their various models.

3. Automatic volume control (Compression. AVC or AGC Automatic Gain Control)

In AVC the loudest sound triggers on additional circuit, which compresses the loudest sounds but permits the weaker signals to progress through on amplifier to the . listener. Here the part of output is fed back to input through the negative feed back mechanism which varies constantly with output. The AVC circuit keeps the output level constant. Hearing aids with this provision are usually more expensive than models without it (Fisch, 1959). One manufacturer in London (The Multitone Co.) was the one of the pioneers in AVC circuitry. They introduced AVC in a table model in 1936 and in a wearable hearing aid in 1948 (Berger, 1970). The first known AVC circuit in an American made hearing aid was the Model K instrument of Vacolite in 1949 (Berger, 1970).

The Vacuum Tube Hearing Aid

The credit for the first vacuum hearing aid goes to Globe Ear-phone Company, for Vactuphone invented by Hanson in 1920. It was commercially distributed in October 1921. The earliest vacuum tube hearing aids were large, bulky and not portable. Most people did not purchase them because of size and price.

Amplivox or Multitone, both of England, were the first one to produce the first wearable hearing aid. These first wearable hearing aids were called Multi-pak because the hearing aid consisted of two or more parts or paks. Multi-pak aids were also named two piece aids, as the model had an amplifier-microphone pak and separate battery pak. Wengel (1937) also produced wearable vacuum tube in United States (Berger, 1970).

The final development for the vacuum tube hearing aid was mono-pak, which took place in 1946. It consisted of one piece vacuum tube which did away with the separate battery pak. The reduction in vacuum tube size brought a similar reduction in battery voltage requirement and thus in battery size.

CHAPTER 4

THE TRANSISTORIZED HEARING AIDS

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THE TRANSISTORIZED HEARING AIDS

In recent years, the transistor, a new type of electronic device has almost replaced the bulky vacuum tubes. Electron tubes utilize the flow of free electrons through a vacuum, but the transistor relies for its operation on the movement of charge carriers, through a solid substance, a semiconductor.

Bardeen, Brattain and Shochley (1947) developed transistor at Bell Telephone Laboratories, It is a current controlling device. The first transistors were of point-contact type. The base crystal was of N type semiconductor material. The emitter and collector were small areas of P type material around the base crystal. The three element rod was housed in a plastic case, with wires of three elements extending outside.

Junction Transistors

The junction transistor may be of NPN or PNP type. They are in many respects analogus to vacuum tubes. There are three elements in transistor, the collector, base and emitter which corresponds to plate, grid and cathode in electron tube. The first junction transistors were made of Germanium.

Transistors are much smaller than tubes, have no filament and hence do not need power for heating and may be operated in any position. They start to work instantaneously and do not need time to warm up. They are mechanically rugged and have practically unlimited life. The transistor stands well under vibrations and other mechanical hazards. The transistor require the power between 0.3 to 1.3 volts and can be operated with pen-torch cells. This brings a great reduction in size and weight of a hearing aid. A junction transistor can function as an oscillator or amplifier.

Transistor Hearing Aid Circuits

The following Figure 2-3 shows a transistor circuit.

The holes in the P-region, known as emitter are repelled by the positive battery terminal toward the P-N or emitter junction. Under the influence of the electric field, the holes overcome the barrier and cross into base region. The remaining holes penetrate through the base and flow to the collector. Current conduction in P-N-P transistor takes place by hole conduction from emitter to collector, while electrons carry the current in external circuit. A large amount of power in the collector circuit may be controlled by a small amount of power in emitter current. The input power to emitter circuit is quite small. The power gain in a transistor thus reaches quite high value.

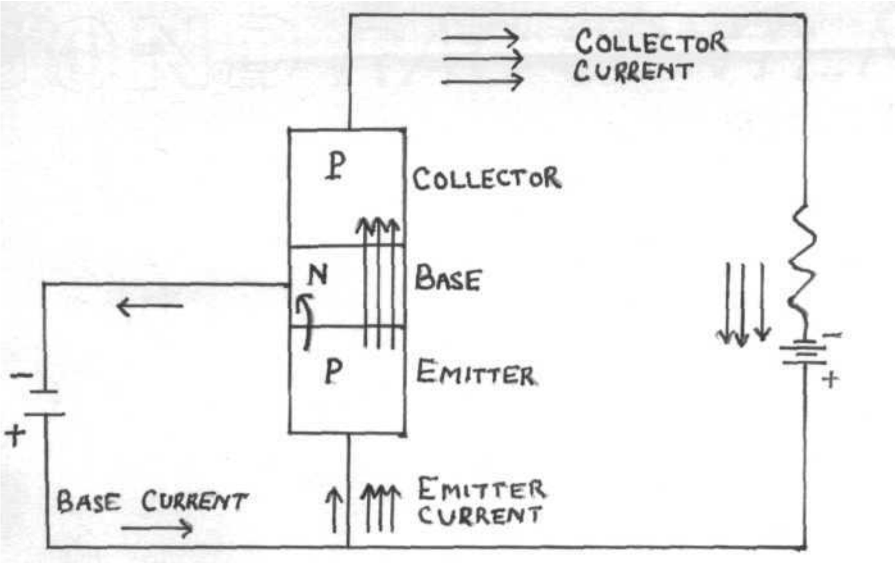

Fig.2-3: Voltage and Currents on a Transistor

The use of microphone and telephone circuit together allows the person to hear through telephone as well as microphone. Most of the Indian hearing aids provide for separate microphone and telephone switch, but one manufacturer has combined these functions into one switch in his product.

A single stage of voltage or power amplification contains a transistor and its related circuitry and components. Three or more stages of amplification are used in hearing aids as one stage is not usually sufficient. Extra transistors are necessary to provide an amplifier with suitable accuracy, stability and flexibility (Victoreen, i960). The design of the circuit plays an important role for the effective use of transistors rather than their number. To obtain a desired output, several stages must be connected so that amplified output of one stage is properly fed to the next stage and amplified* The connection of one stage to other is known as coupling. Several methods are used to couple amplifiers, the three most common are:

1. resistance - capacitance coupling often referred as resistance coupling or RC.
2. impedance coupling, and
3. transformed coupling or TC.

This is illustrated in Figures, 2-4, 2-5 and 2-6.

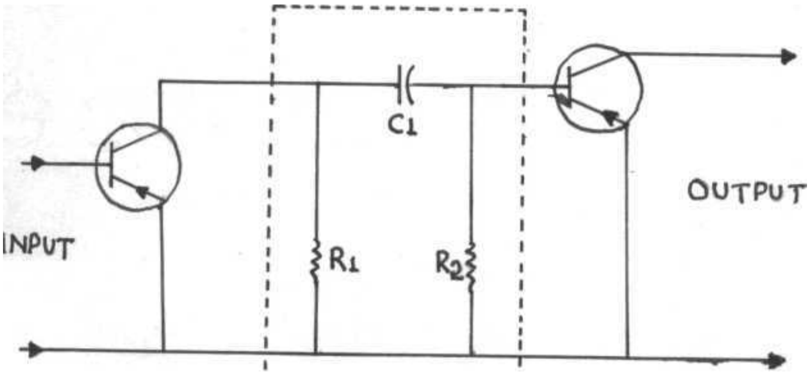

Fig.2-4: Resistance Coupling.

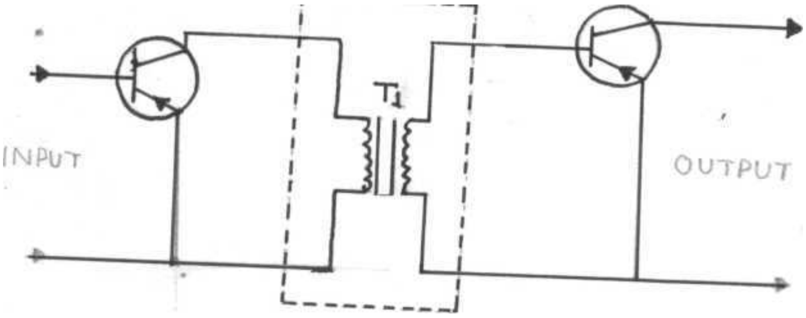

Fig. 2-5: Transformer Coupling.

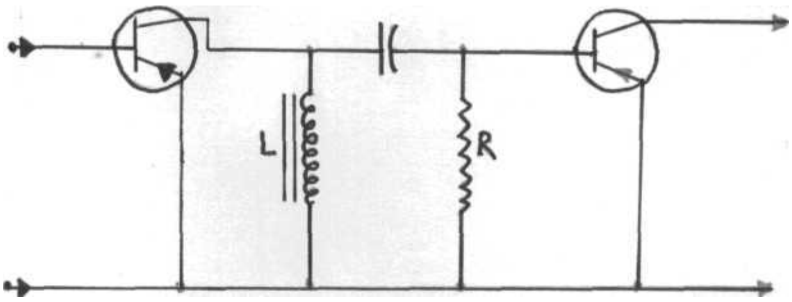

Fig. 2-6 : Impedance Coupling.

Feed back: The persona working with or using hearing aids are quite familiar with feed back, when the microphone of hearing aid is kept too near the receiver, a whistle like sound is heard. Feedback need not be only annoying, it also serves many useful purposes. Feedback may be said to be in operation in a system in which a specific output effect results from a specific input signal, where the output effect in turn affects the signal (Wanink, i968). The feedback may be categorized as positive feedback and negative feedback. Positive feedback is sometimes referred to as regeneration or in phase feedback and helps to add amplification factor. The negative feedback may be referred as degeneration or out of phase or inverse feedback and reduces the amplification factor and distortion. The inverse feedback circuit eliminates or at least minimized undesirable features like harmonic distortion which occur in amplifying circuit.

There is another circuit which concerns the output stage. When the two transistors are arranged so, that as the alternating signal arrives, one of the transistor is pushing it and another is pulling it, such circuit is known as push-pull circuit, which is shown in Figure 3-7. This circuit design allows about twice the power to be produced as for a single output stage, with an increase of about 6 dB in output pressure. This circuit needs an extra transistor and more current especially at high volume control settings.

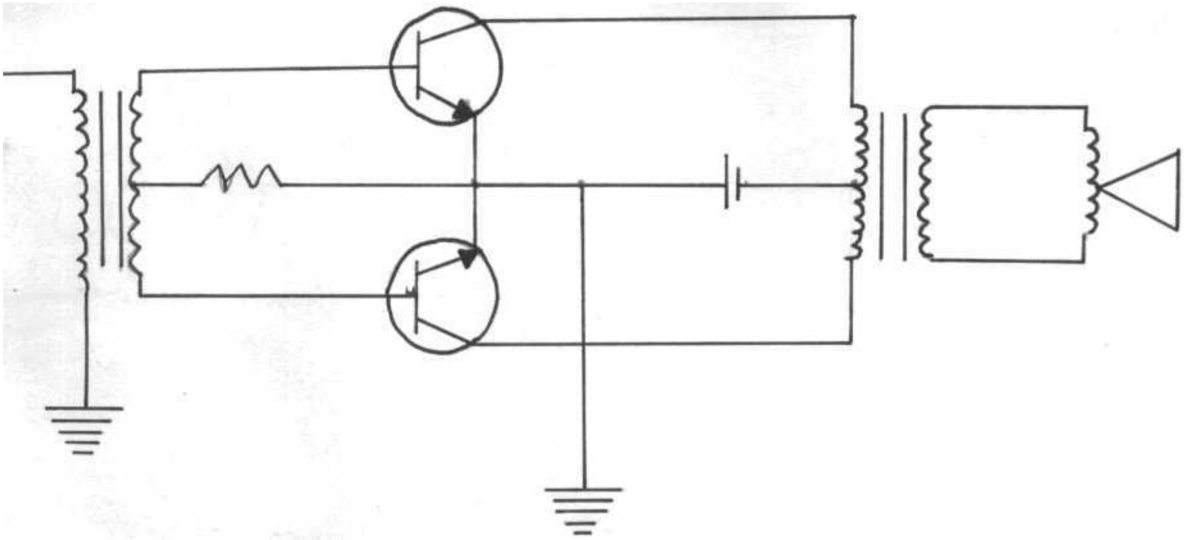

Fig. 2-7: Push-Pull Circuit.

The push-pull circuit has inherently low distortion and provides more power to the receiver.

This electronic Circuit is found in various models of two Indian hearing aid manufacturers.

Out Put Limitation

A person with normal hearing reports that sound is becoming uncomfortable, when the sound pressure level at ear reached 120 dB SPL. Sound becomes painful at still higher levels. Tolerance to loud sounds is reduced, in various ear diseases. Hence we should limit the out put of hearing aid in some way. An out put not greater than 130 dB is recommended, particularly in fitting children with hearing aids.

Two methods of limiting out put are used - Symmetrical Peak Clipping and Unsymmetrical Peak Clipping. Symmetrical Peak Clipping is usually accomplished through the use of push-pull circuit and here only the add harmonics are present as distortion products of the out put signal (Krebs, 1964). A hearing aid with a single transistor out put will usually produce both odd and even harmonics. This is unsymmetrical peak clipping and is an undesirable form of limiting, as it adversely affects the intelligibility of speech. The other advantage of limiting output is battery savings (Berger, 1970).

Automatic Volume Control

In AVC circuit part of the loudest sounds are fed back into an earlier stage of amplifying system, thus preventing the loudest sounds from reaching the ear. The AVC feed back is a working voltage. It incrases the grid or base bias which reduces the amplification factor of the system. The disadvantage of the AVC system is that it takes a fraction of second for the signal to be fed back, which causes a silent period in sound stimuli. Other problems consist of selecting the limiting level, the attack time and recovery time.

The main function of AVC is to limit the actual sound pressure level. This is accomplished by employing a feedback network which is required to measure the output level of the hearing aid and adjust this output until it is at the desired level. AVC system begins to work to a lesser degree even before this limiting level is reached but it takes some time to react. As soon as a loud sound enters the instrument having AVC circuit, the output will initially and momentarily rise to a level above the limiting level. Then the AVC circuit works and brings the output to a limiting level automatically. The time required for the AVC circuit to bring the gain to a limiting level is its attach time.

The circuit automatically returns to its normal gain function, when the loud sound which triggered the circuit is no longer present. The brief period required by the circuit to switch from operation to no operation stage is the recovery or release time. This time can be a critical factor in the disruption of speech flow or understanding. This problem is avoided by keeping the recovery time from 0.05 to 0.15 m.sec. so that soft sounds are not missed out.

At present there is considerable confusion and there are no standard description of AVC circuits. The AVC circuit in particular hearing aid may have a large or small compression range and several attack and release times, care should be taken to check these parameters while prescribing the aid. The AVC may be considered for the person who will use the hearing aid in considerable and fluctuating noise levels. AVC improves speech to noise ration without it noise might be amplified more than speech.

Curvilinear AVC

Amplification is linear in AVC circuit up to the point, where the AVC circuit goes into operation. A new circuit called Adjustable AVC (Curvilinear or Lograthimic AVC) has been introduced, which utilizes combination of AVC and peak clipping. It is an effort to overcome the problems of attacK and release time in linear AVC. The curvilinear AVC quickly adjusts for the increase and decrease in gain brought about by sudden loud sounds and yet does not cause the distortion of peak clipping or the attach release problem of linear AVC. There is no specific limiting level in adjustable AVC and some degree of limiting action takes place even at fairly low put intervals. As the curvilinear AVC circuit is linear only at the lowest sound pressures it does not follow the rule that input plus gain equals output.

There are no indications that curvilinear circuit is necessarily better than no AVC or linear AVC. The choice of these will depend upon the environment in which the hearing aid will be routinely worn. In few aids, there are provisions to cut-off the circuit of AVC or curvilinear, if one desires.

CHAPTER 5

HEARING AIDS WITH DIRECTIONAL MICROPHONES

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HEARING AIDS WITH DIRECTIONAL MICROPHONES

It is well known fact that our ability to understand speech is seriously impaired in the presence of competing noise. Hearing aids with omnidirectional microphones provide equal amplification for sounds arriving from any direction. In a noisy area hearing aid will amplify the speech along with the background noise, to avoid this directional microphones are used in the hearing aid. Willco (1969) developed the first hearing aid with directional microphone.

A directional microphone hearing aid (DMHA) is one whose microphone responds better to sounds from certain angles of incidence. DMHA helps in separating the primary message from the competing background noise through the directional sensitivity of the microphone response resulting in more intense speech from the front of the listener than that coming behind the listener.

To make a IMHA with a maximum sensitivity in forward direction, an additional microphone inlet pointing directly backwards is made, which gives a directional effect to sounds coming from front direction of the hearing aid and also almost eliminates sounds coming from the rear end of the hearing aid.

Figure 2.8 depicts a conventional type of microphone and a microphone with the additional rear inlet. In the conventional type microphone the cavity below the membrane is completely closed while the other microphone is vented through rear inlet. The sensitivity of any type of microphone is proportional to the working pressure on the membrane, defined as the difference between the sound pressures in the two cavities on the both sides of the membrane. When there is difference between the forward cavity pressure (Pf) and the rear cavity pressure (pr) it will result in increased gain with frequency as long as the microphone size can be considered small in comparison to wave length. If Pf=Pr there will be very low output from the microphone.

Figure 2-9 gives directional diagrams for various types of microphones.

1. The circle indicates the response of a non-directional microphone.

2. The 8 shaped curve is of microphone with two sound inlets pointing maximum sensitivity for front and back sounds.

3. The apple shaped curve is a compromise between the two with a small rear inlet with maximal sensitivity only for forward and not rear sounds. With an proper obstruction to rear opening only to a certain degree a phase-shift is

Fig. 2-8: PF (Pressure in the Forward Inlet)

PR (Pressure in the Rear Inlet)

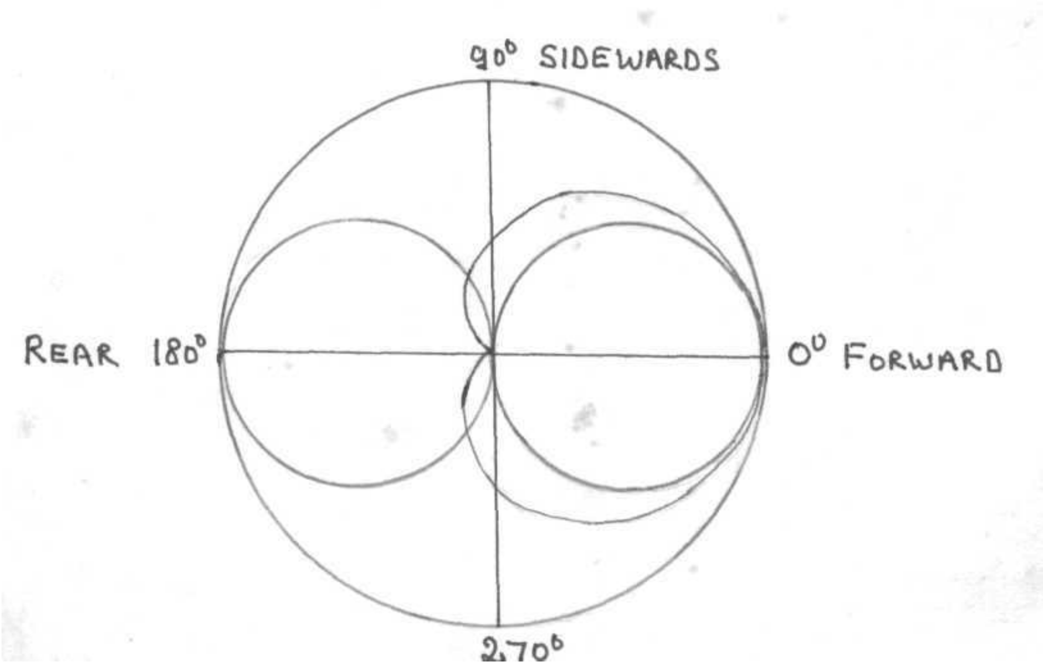

Fig. 2-9: Directional Characteristics for Various Types of Microphones

introduced between the outside sound and the sound just inside the rear inlet. With a phase shift of proper magnitude, almost all sounds coming from the rear can be eliminated.

The Phase Shift or Time Delay

Mostly ½" microphones are employed in the hearing aids. A time delay of 40 u sec (the time required to travel a distance of 13 mm or ½") can eliminate almost all sounds coming through the rear. In Figure 3 it can be seen when a damping element in the rear inlet opening is introduced, this together with the rear equity will form an acoustical delay element and with the correct dimensioning, this element will be able to delay the sound approximately 40 sec.

In Figure 3-1, the working pressure on the membrane from a rear sound is almost eliminated, while it is highest, when the sound comes from a forward direction and the effect is half when the sound comes from one side. This is the directional effect which is being aimed at, and it works in the frequency range from about 200 Hz upto 2,000 HZ, and the difference will be 15 to 20 dB between sensitivity to forward and rear sounds.

To design a directional microphone a very precisely made acoustical delay element having an accurate acoustical resistance corresponding to a delay equivalent to one full

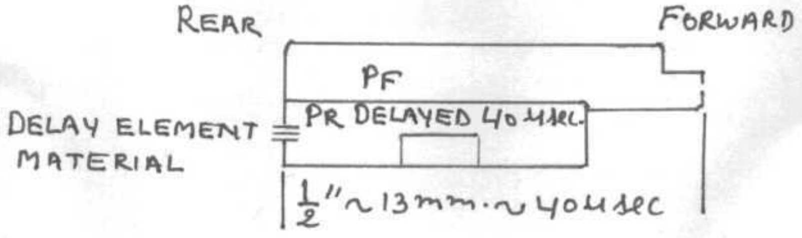
REAR
FORWARD
PF
PR DELAYED 40 μsec.
DELAY ELEMENT MATERIAL
$\frac{1}{2}$" ~ 13mm. ~ 40 μsec

Fig. 3.

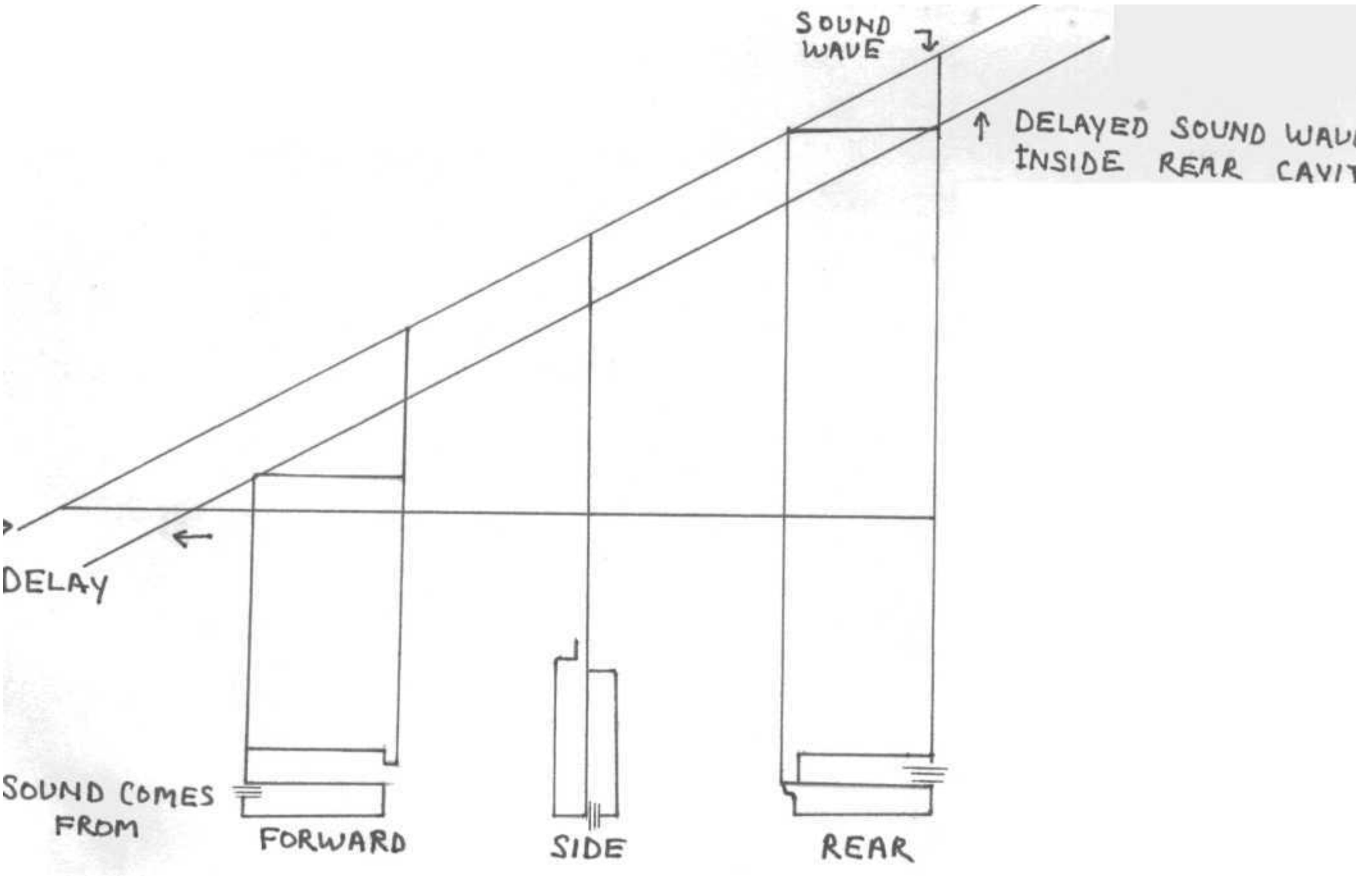

Fig. 3-1 : Directional Micrphone

microphone length needs to be introduced in rear inlet opening. A delay element having too short or long delay will disturb the directional effect. Precautions should be taken that delay element is not covered with dirt or grease as it is likely to cancel the directional characteristics of a microphone.

Frank and Gooden (1974) used normal subjects to study if SD scores obtained using DMHA are influenced by the location and level of the interfering noise source. Results showed that SD improved when the background noise was located at 180° or at 0° to 180° compared to 0° azimuth. The discrimination will be markedly increased when noise is louder or as loud as the speech signal. Authors recommend that EMHA should be used in a noisy environment, while conventional microphone hearing aid in less noisy area.

According to Griffind and Preves (1976) a directional microphone ean be used in body level and in the ear (ITE) hearing aids. They studied directional sensitivity with ITE hearing aid using cardoid and using supercardiod directional microphone in comparison with an ITE and CMHA. It was found that ITE hearing aid with non-directional microphone has some amount directionality because of head diffraction. So, in addition to this if a directional microphone is used there will be:

(a) Extra amount of directionality.

(b) ITE, CMHA has a good SN ratio because microphone inlet is situated in concha of ear and using a IMHA even greater S/N ratio is obtained.

(c) With ITE aid using directional microphone, head diffraction is combined with the rejection of undesirable signals in the rear half by the directional sensitivity and so gives gpod SD (Speech discrimination).

(d) Cardoid and supercardoid directional microphone may offer maximal efficiency when used in an ITE aid located in the concha.

Janet Enger (1978) designed a study to find out whether aided SD scores would be influenced by varying the placement of the IMHA in the sound field testing arrangement. It waa found that orienting the microphone at 0^o of incidence to the primary signal improves SD between 5% and 30%.

Advantages and Disadvantage of Aid with Directional Microphone

1. A DMHA reduces unwanted background noise in comparison with an actual speech signal.

2. It secures improvement in speech discrimination as unwanted background noise can be decreased provided that the person speaking is positioned within the forward directed open axis of the microphone.

3. It improves the users localization.

4. These can also be utilized for increasing the S/N ratio and therefore add new dimension to the speech in noise procedure for hearing aid assessment.

5. Some subjects wearing directional microphone hearing aid noticed that they could not detect sounds behind for example, door and telephone bells.

6. Patients may feel tuned out and unable to react to warning signals.

7. Rubbing noise is also considered as a negative factor.

CHAPTER 6

CROS AND CURRENT VERSIONS OF CROS

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CROS AND CURRENT VERSIONS OF CROS

Hariord and Barry (1965) evolved the principle of Contralateral Routing of Signals (CROS) and reported their study on the effects of placing a microphone on the dead ear side and transmitting sound electrically to the good ear through the use of open ear mould.

Pre-requisites for the use of CROS

1. A patient should be well motivated, as it seems to play a role in the consideration of a CROS hearing aid.

2. Daily activities of living and status of hearing in the better ear of the patient is another important consideration. The greater the demands the greater benefit he should gain from CROS. A person with some hearing loss in good ear, but not enough to warrant the use of amplification directly to that ear can be expected to react more favourably to the use of CROS amplification.

Limitations of CROS

The limitations reported are:

1. The distortion of the sound being fed to the good ear by the hearing aid.

2. Inability to locate an optimum volume selecting for sounds in the distance and sounds at close range, making the close sounds too loud.

3. Person will not be able to enjoy the music properly while wearing the hearing aid.

4. Difficulty in understanding in noisy places and groups.

After the introduction of CROS type hearing aids several versions of CROS have emerged. These versions of CROS either eliminate or utilize the head shadow effect and also rely on the utilization of ear piece acoustics.

CURRENT VERSIONS OF CROS

Classic CROS :

Classic CROS is useful for persons having unilateral hearing loss with the better ear manifesting a slight loss in the speech frequencies range. Better ear though not normal should not be accluded by an ear piece as the objective is to provide the user with improved hearing in the form of mild increase in sensitivity and better auditory discrimination for conversational speech originating from the side of the poorer ear.

Here the microphone is detached from the main component of the hearing aid and located in the vicinity of user poor ear, Figure 3-2. It can be housed either in eye glass or postcurricular units. A polythene tube which is shaped to the contours of the ear, aimed directly into the canal will provide an ideal arrangement for the aid mounted in eyeglass. A greater stability of the physical placement of the behind the ear unit is obtained if tubing is held in the ear with the skeleton type ear mould.

Classic CROS suits well to persons with a unilateral hearing loss and a mild loss in the better ear, but a person having normal or near normal hearing will find it too loud for him in daily listening conditions.

Mini CROS

Mini CROS is identical in its construction to Classic CROS and its purpose is also to eliminate the head shadow effect for a person with unilateral hearing loss. It differs in two respects from Classic CROS. First, tube is not used to carry sound into ear canal but amplified sound is allowed to escape through the nozzle to which the tubing is ordinarily attached. Second, a more powerful aid than that used for Classic CROS is needed, as much of the amplified sound dissipates before reaching the better ear. Mini CROS (Figure 3-3) reduces the consequences of the head shadow effect and eliminates the chances of over amplifying the normal ear.

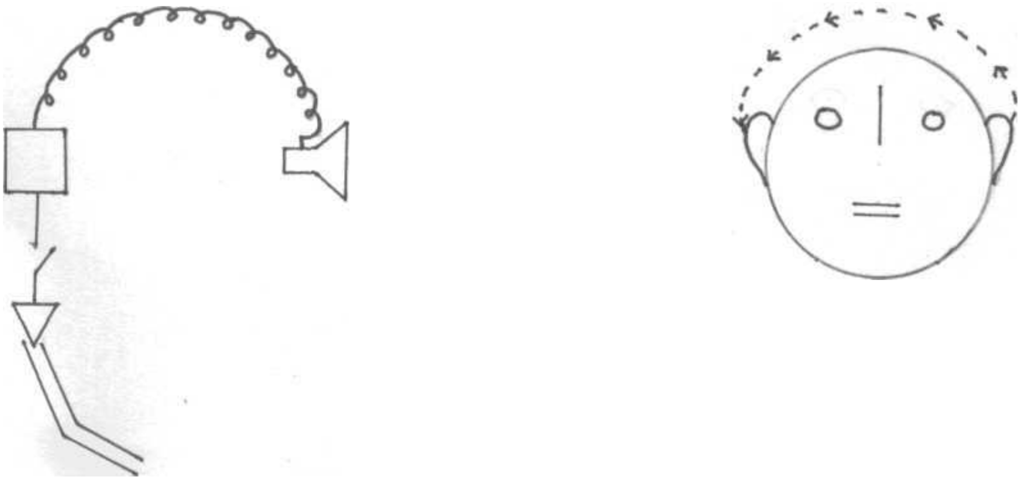

Fig. 3-2 : Classic CROS

Fig.3-3: Mini CROS

Hi CROS

Hi CROS (Figure 3-4) is especially useful for the persons with a bilateral high frequency hearing loss. The objective of Hi CROS is to offer the high frequency and simultaneously reducing low frequency amplification. It utilizes the attenuation of the head to provide greater amplification of high frequency sounds to an open ear canal before encountering acoustic feedback.

Hi CROS produces improvement in auditory discrimination for speech, compared with unaided performance especially at conversational loudness, but in cases of very sharp high frequency loss it fails to provide optimal improvement in speech.

Focal CROS

Ideal candidates for Focal CROS are those persons whose one ear is unaidable and aidable ear has a precipitous drop starting at 1000 HZ, but it may also be used with persons having a bilateral precipitous high frequency loss. In Focal CROS (Figure 3-5) microphone includes a nozzle to which polythylene tube that extends into the canal of unaided ear is attached. It has been found Focal CROS sometimes provides better discrimination for speech than Hi CROS, the reason may be that the localization of the sound pick up in the concha utilizes the resonant

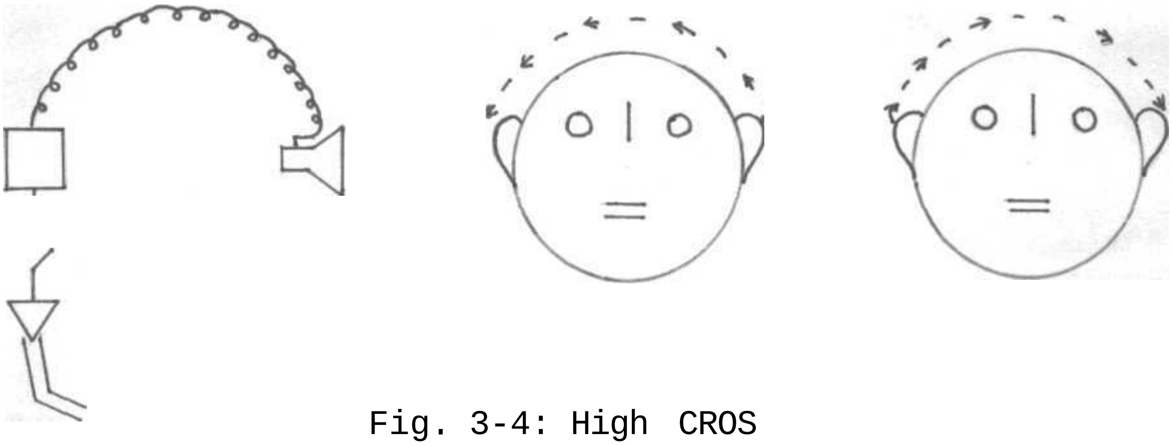

Fig. 3-4: High CROS

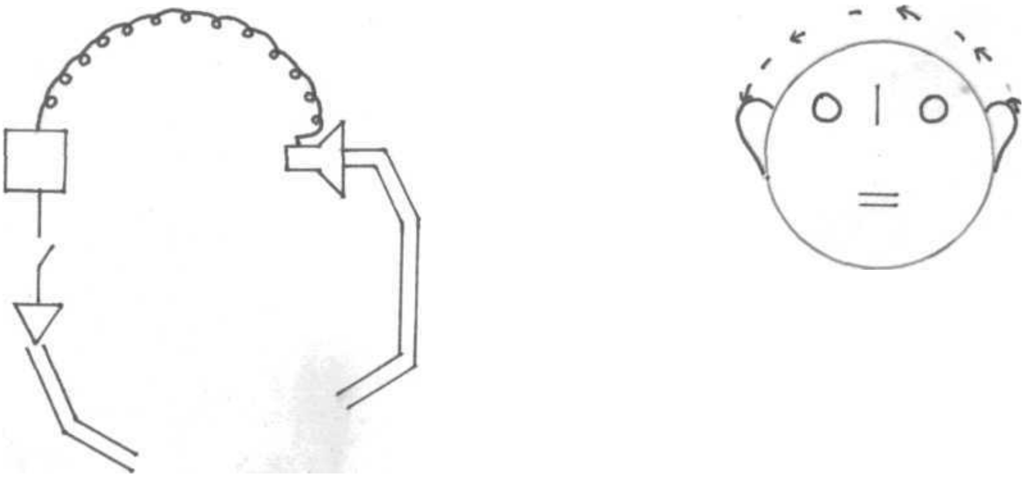

Fig. 3-5: FOCAL CROS

characteristics of ear canal to enhance the high frequency elements of speech. When these enhanced high frequency sounds strike the microphone, it reduces the need for high gain setting, resulting in less distortion from the hearing aid.

The unique location of the microphone suggests ather possible uses for Focal CROS which are:

1. Persons who work or engage in numerous out-door activities, the placement of microphone pick up in concha may lessen annoyance from wind noise.

2. In noisy places, it may decrease the amplification of background noises by increasing the signal to noise ratio by capitalizing on the protection or acoustic shield offered by the auricle. .

Power CROS

The Power CROS (Figure 3-6) differs from the Classic CROS in two ways:

1. A solid or closed ear mould is used in Power CROS.
2. This hearing aid is in the highest gain category for ear level instruments.

Power CROS allows for greater utilization of amplification than the conventional monaural ear level arrangements

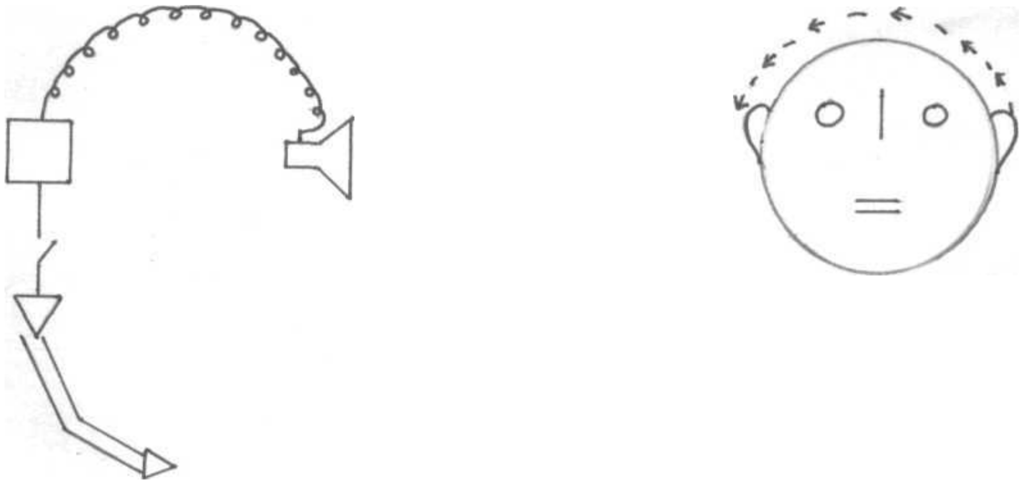

Fig.3-6: Power CROS

located on the side of the aided ear, thus permitting the use of an ear level hearing instead of body borne instruments. It capatalizes on head shadow effect by using attenuation of head shadow to reduce feedback potential for severe hearing loss.

Bi CROS (Bilateral CROS)

The Bi CROS works on the same principle as the single CROS aid. Sounds are picked up by a microphone on the side of the poor ear and routed across the head either over the head via a wire attached to a head band (Figure 5-7) or through a wire attached to the inside of glass frames (Figure 3-8). The latter type has been found to be especially convenient for many users.

With the eyeglass construction a microphone is located in the temple on the side of unaidable ear, which converts the incoming acoustic stimuli to an electrical signal that is transmitted (by way of wire extending from the microphone around on the inside of glass frame) to the hearing aid amplifier located in the temple near better ear. Other microphone mounted on the side of the better ear also delivers electrical stimuli to the hearing aid amplifier. The signals from the two microphones are mixed, amplified and fed to an receiver which delivers the message through a plastic tube to better ear. This tube is held in the ear

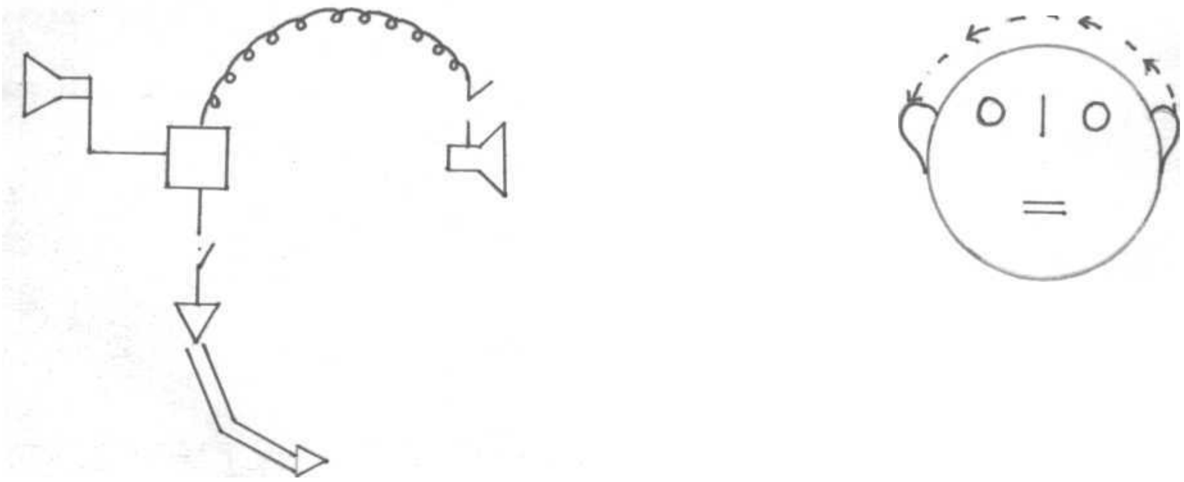

Fig. 3-7: Bi-CROS

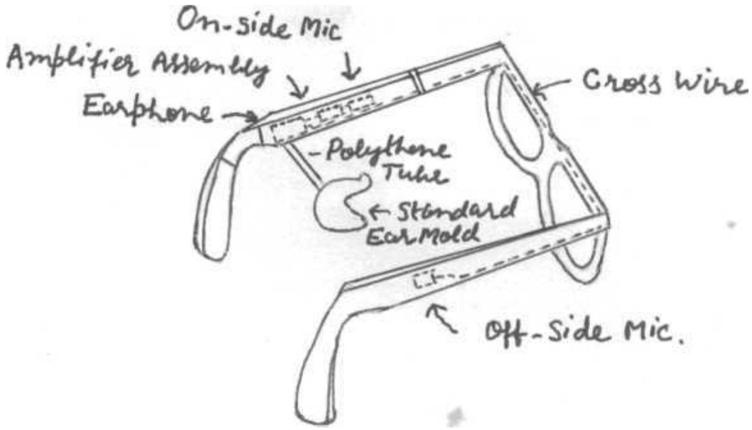

Fig. 3-8 : Bi-CROS

canal by a standard hearing aid piece. This type of hearing aid is illustrated in Figure 3-%.

A candidate for a Bi CROS hearing aid is one who has a better ear with sufficient hearing loss to warrant amplification and an unaidable ear which may be due to variety of causes.

The major advantage is that person can easily hear and understand speech when others are sitting on the hearing aid user's "bad side", while he will also be able to hear those people positioned on the side of good ear. Bi CROS eliminates the need for the person to turn his head in compensation of unilateral hearing problem while others may be able to localize the source of sounds.

Open BICROS

The open BICROS (Figure 3-9) is essential the same as the BiCROS but here the solid ear piece is replaced by tubing or an open mould to provide an unaccluded ear canal, which may be necessary for comfortable and effective use of amplification for a person having high frequency loss in better ear. OREN-BICROS may provide an alternative to CLASSIC-CROS in some eases, depending upon the listening demands and precise nature of the hearing loss in user's better ear.

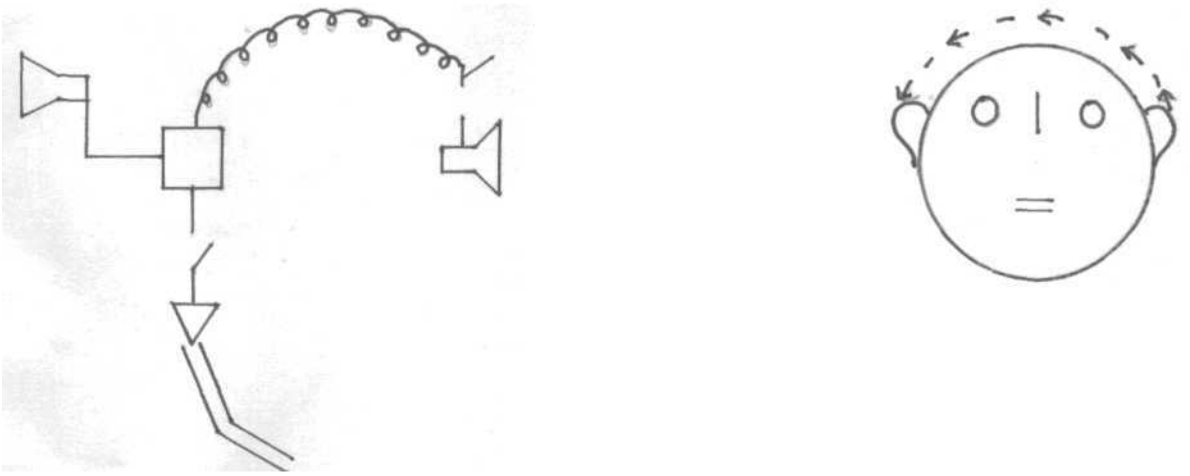

Fig. 3-9: Open Bi-CROS

Uni CROS

Dunlavy (1967) developed UNI-CROS by a unique modification of CROS and it is intended to provide amplification for the persons with a bilateral asymmetrical loss with both ears aidable. UNI-CROS (Figure 4) is designed in one microphone located on the side of the poorer ear, feeding into two amplifiers located on each side of the head and can have different gain characteristics The better ear utilizes CLASSIC or HI-CROS principle while the poorer ear uses conventional monaural amplification with a solid ear piece. Incoming signals are received by the single microphone and routed to both ears in a different manner. In a noisy situation, the user may find it helpful to turn off the monaural aid and use just the CROS amplification.

Multi CROS

MULTI-CROS can be used as a CLASSIC-CROS, BICROS, or OPEN-BICROS or as a conventional monaural aid (Figure 4-1). Basically it is a BICROS hearing aid with a separate on-off switch for each microphone. Thus by activating both microphones, the aid can function as a BICROS or OPEN-BICROS to be used with appropriate ear piece. It becomes a conventional monaural aid when the ipsisateral side microphone is turned off. In a post-auricular assembly, MULTI-CROS can be easily used either for right or left ear. Some eye glass MULTI-CROS

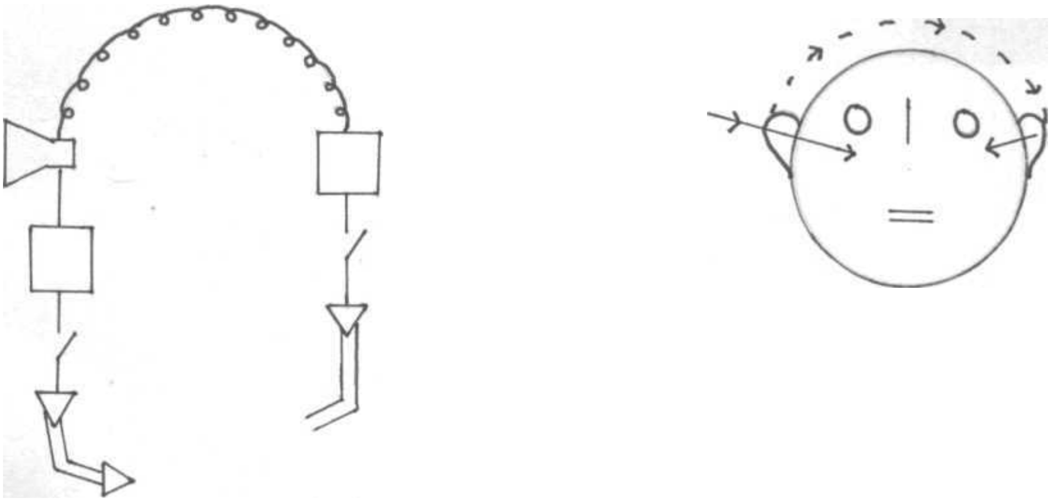

Fig. 4: uni CROS

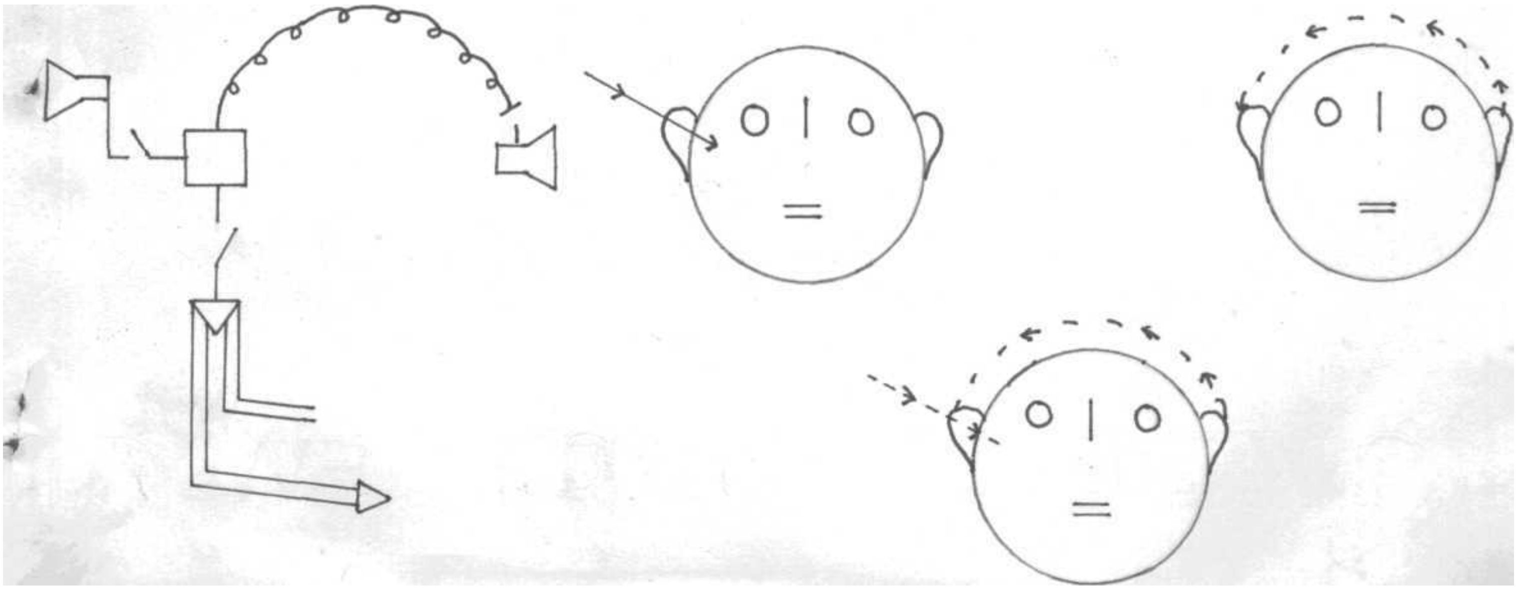

Fig. 4-i: Multi CROS

aids have also been constructed with plunger type temple bars, that can be interchanged without disturbing the wiring, while others can be used as a FOCAL-CROS and any of the other types mentioned. The MULTI-CROS may be used as an IROS (Ipsilateral Routing of Signal), MINI-CROSS and HI-CROS.

It may prove to be satisfactory for persons with unaidable ear, particularly if the better ear manifests a mild to moderate, flat or high frequency loss. MULTI-CROS should be recommended for those persons who can use it with good insight and whose listening needs vary, since it could require manipulating earmoulds or tyubing or both as well as microphones.

IROS

Green coined the term IROS to specify the ipsilateral Routing of Signals to an unacculded ear canal. IROS is a conventional monaural mild-gain hearing aid used with an open ear piece or shaped polythene tubing (Figure 4-2).

While selecting IROS aids for trials, the gain offered in the higher frequencies should be considered, not the average HAIC gain as it could be misleading. Overlooking this precaution can result in overamplification for the patient and rendering him ineligible for IROS amplification because of a feedback problem. IROS is less consipicuous in postauricular units and if a person with bilateral high frequency loss can

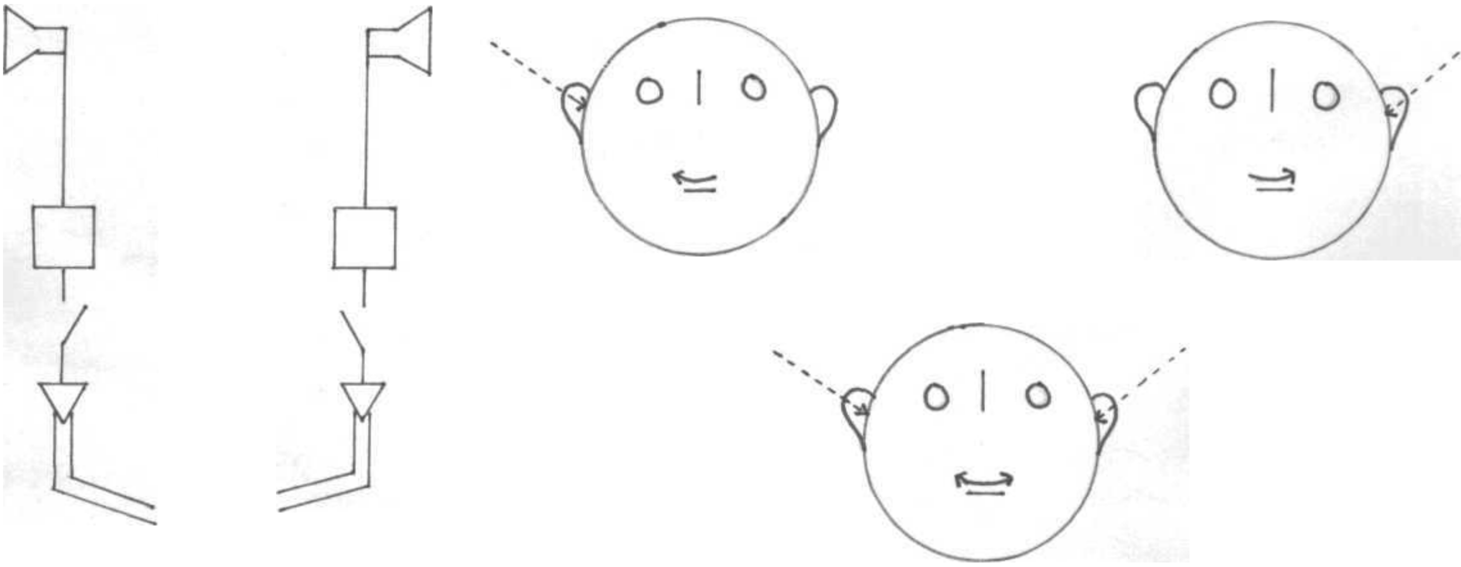

Fig. 4-2 : IROS

use IR0S successfully, binaural amplification can also be considered.

FROS

It is a refinement of IROS (A.R. Dunlavy, Unpublished data), here the microphone is placed in the frame of eyegalss, just in front of the temple bar hinge. Dunlavy claims to achieve approximately 10 dB more gain with FROS (Front Routing of Signals) than if the microphone were mounted in the temple bar or in a postauricular case. FROS should preclude the need for HI-CROS in some cases and may provide binaural amplification for some persons with a bilateral high frequency hearing loss.

Bi FROS

Bilateral front routing of signals utilizes two microphones, one on each side of the frame. These microphones send signals to a receiver in the temple on the same side. BIFORS is a binaural system and has advantages of greater gain capability, directional characteristics and open moulds if necessary.

Double FROS

Two microphones, but only one receiver are employed. It can take the place of BICROS fitting in some cases.

BIFROS 270

Bi FROS 270 is the newest variation of FROS and uses four microphones and two receivers. Two microphones are located in the frame and one in the each temple. The two microphones on each side deliver signals to the receives on the same side.

According to Dunlavy (1974) this 270° sound stimulation provides "a more full rounded sound" and potentially allows front back localization, which is not often demonstrable with most hearing aid configurations.

CHAPTER 7

RISK OF HEARING AID USE

POSSIBILITY OF INJURY BY AMPLIFIED SOUNDS

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RISK OF HEARING AID USE

POSSIBILITY OF INJURY BY AMPLIFIED SOUNDS

Research has been relatively inconclusive and some has been contradictory regarding sensorineural hearing losses resulting from the use of high gain and high output hearing aids. However the balance of recent evidence has been to the effect that appropriate fitting of powerful aids is not likely to be a cause of further hearing deterioration (Hine and Furness, 1973).

Review of the literature identifies scattered articles dealing with this problem. Kastern and Braunlin (1970) presented a case study in which they could create deterioration in patient's hearing through the use of a hearing aid. Their patient was a 10 year old girl with bilateral, moderate sensorineural hearing loss. She wore a body type hearing aid satisfactorily in one ear for 14 months and then began to complain that aid was not helping as much as it did previously. An audiometric evaluation showed marked worsening of the hearing in the aided ear and no change in the hearing of the unaided ear. Subsequent hearing aid evaluation was done and she was advised not to use her present aid. Re-evaluation of her hearing 7 weeks later showed 20 to 30 dB improvement in the previously aided ear. Eastern and Braunlin were able to show temporary deterioration in the aided ear, regardless of which ear wore the hearing aid. The aid in question had an average gain of 34 dB and average saturation sound pressure level of 120 dB. They recommended a less powerful aid of 30 dB gain and l00 dB saturation sound pressure level. When the less powerful aid was fitted to the patient and worn alternatively between ears, evaluations verified that temporary deterioration of hearing was no longer present. Their conclusions included that all children, and their fitted hearing aids be carefully re-evaluated shortly after initiating hearing aid use* They also suggest that clinicians have some means of verifying the electracoustic characteristics of each hearing aid.

The question of traumatic hearing loss from personal amplification was raised as early as 1940 by Holingren who affirmed that in his experience hearing aids never affected hearing adversely. Divergence of opinion on this issue is reflected by two national conference groups on two continents in 1967. While the World Health Organization stated that no evidence existed to support the contention that use of a hearing aid could cause deterioration of hearing, the American Speech and Hearing Association in 1967 that the question of whether amplification can endanger residual hearing had not been adversely resolved.

Harford and Markle (1955), Ross and Truex (1957) and Staloff (1961) present case studies which link temporarily hearing loss to hearing aid use. In each case removal of hearing aid permitted the hearing of the aided ear to return to its presided level. Placing the hearing aid in opposite ear created deterioration of hearing in that ear also.

Robert (1970) presented a case study of child who showed a 40 dB decrease in hearing level in the aided ear which did not recover even though the aid was removed. Barr and Wedenberg (1965) conducted extensive etiological evaluation in an attempt to isolate the cause of continuing deterioration of sensori-neural hearing loss. He classified children into two categories: endogenous and exogenous. The findings of this study indicated that individuals with exogenous (acquired hearing losses due to maternal rubella) showed no sign of detereoration despite the constant use of hearing aid. Exogenous cases arising from meningitis, however and 50% of the endogenous cases showed continuing detereoration of the residual hearing, but this appeared to be spontaneous and not attributable to hearing aid use. Roberts as well as Barr and Wedenberg (1968) suggest that use of amplification in children with progressive sensorineural hearing loss may cause acoustic trauma and thus accelerate the hearing deterioration in the aided ear.

The evidence for powerful hearing aids causing threshold changes in the aided ear certainly seems to confirm this unfortunate circumstance as a real possibility to be considered by clinicians. Further developments will hopefully assist in identifying children, in advance, who might suffer trauma from hearing aid use. Danaher and Picket (1972) noted in subjects with profound hearing loss that their most comfortable hearing level may actually be 125 dB SPL, with a loudness discomfort level of 128 dB SPL. Many patients with profound hearing loss have no loudness discomfort at any level, while other patients with seemingly similar hearing loss, have loudness tolerance problems so severe that they can not tolerate any type of amplification.

Naunton (1957) reviewed charts of 120 patients selected from a population of 1480 cases. He compared threshold from the non-aided ear with the aided ear of these patients, and concluded that changes in hearing due to hearing aid use are statistically and clinically non-significant. On the contrary in a similar study by Kinney (1961) with 178 patients, sufficient numbers of unilateral trasmatic hearing loss were attributed to amplification and he recommended that aids fitted to children with sensorineural deafness be limited to less than 40 dB of gain. Kinney also recommended frequent audiologic follow up visits for children who wear hearing aids.

Macrae and Farrant (1965) support Kinney's finding using 87 children. They divided the children into two groups, classified the maximum power output of their hearing aids ie., moderate and high output. They also considered etiologic factors, but, unlike Barr and Wedenberg, concluded that deterioration in hearing was the same whether the loss was exogenous or endogeneous. They supported Kinney's contention that there was a correlation between hearing aid output and the amount of deterioration. They concluded that:

(a) individuals with sensorineural hearing loss should be fitted with limited maximal power output hearing aids,

(b) frequent follow up of aided children should be required,

(c) children should alternate the use of aid in each ear whenever practical,

(d) users should be cautioned about wearing hearing aids in high ambient noise environments.

Macrae (1968) found substantial temporary threshold shift in the aided ears of the children with sensorineural deafness following use of powerful hearing aids. He measured the hearing levels of four children from a school for the deaf on Friday afternoon after the children had worn their hearing aids all week. He then kept the aids, and deprived the children of amplification for 66 hours, until the following Monday morning which again deteriorated after four hours of hearing aid use.

Beliefler et.al. (1968) conducted the study to know the effects of high gain amplification on children in a residential school for the deaf. The audiometric records of 58 children covering an eight to ten year span were statistically analyzed to determine if hearing aid amplification had a deleterious effect on the residual hearing. In 25 cases, the unaided served as a control for the aided ear. In the remaining 33 cases, the objective was to statistically select the aided ear. It was hypothesized that if amplification had an effect on hearing loss over a period of time, it would be apparent from the comparison of the aided and unaided ears. The result indicated no differences between ears for subjects during the years of study.

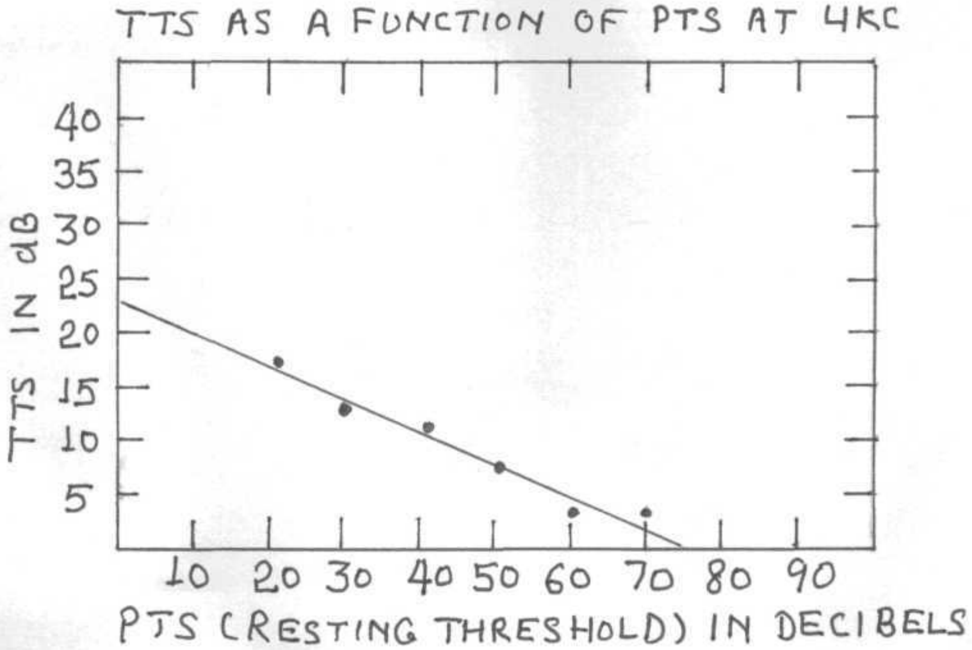
Derbyshire (i976) administered to one hundred children "before and after" pure bone tests of hearing over a period of at least two years, nine months, during which they wore high gain and/or high output hearing aids. Three clear trends were shown by the data obtained. First, there was no evidence that the wearing of aids caused deteriorations of hearing. Ears to which aids were not fitted tended to give slightly less good readings on the re-test compared with first test results. The only frequency which showed any deviation from the first of these trends was 4000 H2 where there was no evidence of general recorded improvement with the use of amplification.

The use of amplifying device for the hearing impaired child in the first year of his age entails some difficulties in anatomical, physiological, central nervous system and nursing respects. These are additional therapeutic risks resulting from maturity problems of the auditory system and diagnostical impediments within first 6-8 months. Biesalshi (P) and Stange (G) (1975) took measurements of the central hearing potential in 21 children of various age groups and 11 adults with normal hearing and studies of progressive disturbance confirm the presented doubts about and prove the risk of a too early and too intensive application of a hearing aid. It is cautioned against providing a child with a hearing aid before the 9th to 12th month without a thorough evaluation.

There have been one or two allusions to the possibility of injury to hearing of infants by incautious exposure to very loud sound. There are two situations in which danger might occur. One is the case of the child whose ears are normal or near normal but who, because of central condition does not develop speech and is mistakenly given amplified sound. The second situation is the case of the child who is actually hard of hearing from some sensori-neural disorder and whose hearing loss might be increased by overloading the ear with sound. The output of high gain aid is enough to produce a hearing loss in normal or near normal ears if they are worn consistently with high gain outputs.

Extensive research has shown that the production of noise induced hearing loss is inversely proportionate to the level of resting threshold i.e., as the resting or permanent threshold level increases, the amount of noise exposure necessary to produce a further elevation of hearing level increases. When a maximum permanent hearing level is reached at approximately 75 dB, further increase of hearing level is very unusual, particularly in frequencies below 2000 C.p.s. Figure 4-3 demonstrates this phenomenon very well, since it has been shown that if no temporary threshold shift occurs, no change in resting threshold is produced.

Rojskjaer (1960) reported three hundred and ninety cases, both adults and children, with different kind of hearing diseases of the inner ear and treated by hearing aid for five years or more, were re-examined. He found that 9 cases who since the first hearing examination had developed additional hearing loss in the ear using the hearing aid. This ear alone was worse than before; the other ear was unchanged. It might be a series of coincidences and the possibility that it might be progression of the original disease in the hearing aid ear only, cannot be excluded. Against the chance of coincidences was the fact that no cases were found with the opposite findings, that is further impairment of hearing in the ear without the hearing aid.



TTS AS A FUNCTION OF PTS AT 4KC

Fig. 4-3

The patients were using their hearing aids all day and every day.

It is important to examine woth positive and negative results with respect to the future, including the risk of treatment itself, but this risk must not restrain us from treating our hard of hearing patients, as well as hard of hearing children with little residual hearing with aids. The hearing aid is of such great importance for the acquisition of language and speech and for education as a whole that we ought not to igpore it, since the residual hearing in children is of no use if sounds cannot be amplified. The further developments of audiology should assist in the elimination of the risk of acoustic trauma by amplified sounds, particularly by identifying in advance those patients especially sensitive to injury by noise.

Ross and Lerman (1967) recommended the following to alleviate, or minimize the possibility of traumatic hearing loss related to over amplification:

1. Additional hearing loss is most likely related to use of extremely high levels of maximal power output which exceeds 130 dB SPL; hearing aid recommendations for children with sensorineural hearing loss should include only hearing aids with less than 130 dB SPL MPO.

2. The traumatic hearing loss may be related to

hearing aid use, the incidence seems quite small and this concern is by no means to be interpreted as contra-indicatory to amplification. Denyinga child hearing aid during the critical language years may only be saving his bearing for no good use. If the aid is fitted too late it will not help any way.

3. Frequent follow up, audiometric and hearing aid evaluation is an absolute must for all children with sensort-neural hearing loss who wear hearing aids. Aided children may be reevaluated twice a year.

CHAPTER 8

COCHLEAR IMPLANT AND OTHER METHODS OF STIMULATION

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The Cochlear implant attempts to replace a function which has been entirely lost in the cochlea. Many people mistake cochlear implant for a new type of hearing aid. In cochlear stimulation attempt is made to replace the functions of lost or damaged hair cells by transforming the mechanical energy of sound to electrical energy, which directly excites or stimulates the remaining auditory nerve. Only the patients with loss of hair cells but with remaining viable auditory neurons can benefit from this device. Keeping this in view it becomes necessary to differentiate various types of losses grouped under the term sensorsneural hearing loss. These losses can be divided into four types:

1. Sensory
2. Neural
3. Brain stem, and
4. Central.

When the sensory cells of the cochlea are not capable of stimulating the auditory nerve, the sensory hearing loss results. An example of this loss is which occurs from ototoxic antibiotics.

Neural hearing impairment occurs when the auditory nerve is diseased and does not conduct impulses to the central nervous system generated by the hair cells. Acoustic neuroma can cause this type of loss.

A central hearing loss results, when the central nervous system is unable to make meaningful interpretations of the electrical impulses received from an intact peripheral mechanism.

Above listed definitions are theoretical, only extensive testing will determine type and degree of hearing loss. But usually the sensory loss is most often encountered.

Differentiation of Type of Sensori-neural Hearing Loss: Selection of Patients with Cochlear Implantation

First part of the testing involves the determination of the site of deficit in the inner ear and central mechanism by the techniques of the electrocochleagraphy, brain stem audiometry, and cortical audiometry (House and Brachmann, 1974).

It is possible to measure:

1. The Cochlear microphonic from the hair cells.
2. The eight nerve action potential from the auditory nerve.
3. The potentials generated in brain stem.
4. The potentials from the auditory cortex,Conclusions can be drawn as to the type of loss from the Table:

THEORETICAL TYPES OF SENSORI-NEURAL HEARING IMPAIRMENT

1.	Cochlear Microphonics absent	Sensory loss
2.	Cochlear Microphonics present Action Potential Absent	Neural loss
3.	Cochlear Microphonics present Action Potential present Brain Stem response absent	Brain Stem deficit
4.	Cochlear Microphonic present Action Potential present Brain Stem response present Evoked cortical response absent	Cortical deficit.

It may not be possible to classify the losses as stated above as the production of subsequent potential is dependent on the presence of an adequate potential proceeding it Therefore when the Cochlear microphonics is absent. recording of potentials will not determine the intactness of the central mechanism and thereby separate sensory from neural impairment.

The second part of the testing makes this differentiation which consists of electrically stimulating the patients inner ear through the same needle as was used for recording by applying a low frequency (30 to 120 HZ), low voltage (.3to 1.2 V) alternating current. These parameters were experimentally found most effective in earlier studies.

The brain stem response can be measured from intra cochlear electrodes. Its amplitude grows with the increasing stimulus intensity.

The impairments may be divided into sensory or neural depending on whether or not patients perceive an auditory sensation, through the periphally applied electrical stimulation, and patients selected for cochlear implant surgery. All the patients thus selected have responded to the permanently implanted device for directly stimulating the auditory nerve.

Brackmann (1976) tested 225 patients with electrical stimulation, approximately two-thirds of these patients have responded to electrical stimulation. A number of patients experience a painful sensation due to stimulation of Jacobson's nerve, which in turn limits the amount of current to be applied to the promontary electrode.

Chouard and MacLeod (1976) used operative testing with placement of an electrode directly on the round window and reported that in 59 patients so tested, all but three patients who had removal of the auditory nerve secondary to acoustic neuroma removal responded to electrical stimulation.

Brain Stem and Cortical Stimulation

Another method which attempts to restore the hearing in deaf patient is by stimulating the auditory cortex of the brain. There is a great deal of basic research and limited clinical research on direct stimulation of the auditory cortex which give promise to the possibility (Dabelle et.al., 1973).

It has been shown in human studies that stimulation of discrete areas of the brain will produce different pitch sensations. This offers a hope that a prosthesis could be constructed which would transmit speech information to the brain in a coded form.

Auditory Nerve Survival

For a cochlear implant to be effective, some auditory neurons must remain viable after there has been total hair cell degeneration, but the exact number is not known (Wallach et al., 1973). Lawrence and Johnson (1973) on the basis of histopathologic studies have shown that there is loss of some of the ganglion cells in all cases where there is severe loss of hair cells. In almost all cases 5 to 10% of the ganglion cells and neurons remain. The neurons are metabolically and physiologically independent of hair cells, but they are some what dependent on the supporting cells of the organ of corti. Therefore any cochlear prosthesis must respect the integrity of organ of corti as much as possible.

The four types of damage which can occur to the neural tissue on cochlear implantation are:

1. Mechanical
2. Chemical
3. Thermal
4. Electrolytic.

An electrode should be designed to produce the minimal amount of mechanical injury to the remaining cochlear structures and auditory nerve. The scala tympani appears to be the best site to peace on electrode to produce the least amount of cochlear trauma. Simmons (1967) states that placement of scala tympani electrodes in animals have shown little damage to the hair cells or neurons. An electrode in the scala vestili would very likely injure Reissner's membrane which is found to be associated with neuronal degeneration.

Patients who have been implanted with silastic molded scala tympani electrode have shown little changes in stimulus parameter that they have been implanted. Similarly, patients implanted with a wire electrode in a scala tympani have shown little changes in their ability to be stimulated over a period of implantation of up to five years.

Simmons (1966), Simmons and Glattke (1972) state that electrodes directly placed into the auditory nerve through the modiolus are well tolerated in cats and in one case of chronic implantation in human. All of this evidence suggest to the likelihood that the auditory nerve will survive mechanical effect of long term electrode implantation.

The second type of damage which can occur from electrode placement is chemical, which can be minimized by selection of chemically inert material that are resistant to alkalis and corrison like platinum, iridum and titanium. Teflon, silastic and various types of varnishes are available for insulation. Teflon is least likely to develop insulation leaks. Another material Pyre ML has also been well tolerated in cochlear implants of human and animals.

The third type of damage is thermal which may occur from the electrical stimulation of neural tissue but the current levels necessary to stimulate auditory nerve are very low and thermal damage is unlikely. The heat dissipation capabilities of the tissue environment exceed the beat which is produced by the current adequate for stimulation.

Wallach and Cowden (1974) have shown that modiolar electrodes directly in the auditory nerve have the lowest levels of current necessary for auditory stimulation. Scala tympani electrodes require a current in the microampere range of less than 200 ampere and well below the level that is found to cause damage by heat.

The fourth type of injury which can occur due to electrical stimulation of neural tissue is electrolytic but use of alternating current is less likely to produce electrophoresis than a direct current. The low currents necessary for stimulation reduce the hazards of electrolysis and it also not likely that charged ions such as chlorine will accumulate in sufficient quantity to be injurious.

The loss of surface would be other undesirable effect of electrolysis. Experience has shown that electrodes small enough to be introduced to scala tympani or modiolus will have sufficient area so that they will have long life inspite of small amount of electrolysis that occur.

Information Transfer By Electrical Stimulation

It is difficult to exactly reproduce and duplicate Cochlear functions. A good speech discrimination up to 4000 HZ can be obtained by stimulation through a single electrode. But this will only provide periodicity pitch and hence the speech discrimination will be limited.

The work on auditory nerve stimulation was initiated in early 1960 (House, 1976, House and urbon, 1973). The three patients had five wire electrodes planted and showed possibility of pitch discrimination for frequency 25 to 400 HZ. Increase in pitch was also reported when electrodes were stimulated near to the basal end of the Cochlea and when the pulse repitition rate was increased.

The dynamic range of electrical stimulation is much reduced than that of normal cochlea. King et al., (1972) reported it to be 4 dB. Simmons (1966) found a dynamic range of 15 to 25 dB in his subjects and they did not learn to identify words or phrases. Merzenich et. al., (1973) also confirmed the finding of narrow dynamic range. Hanson and Lauridsen (1975) noted very narrow dynamic range to electrical stimulation and encountered problem with the skin in the area of exit of electrodes.

Michelson (1971) found that patients respond to sinusoidal electrical stimulation across the frequency range of 25 to above 10 KHZ. Pitch changes as a function of stimulus frequency from approximately 50 Hz to 500 Hz, above it pitch changes very little. Electrode selection, stimulus repitition rate and current intensity influence the subjective pitch sensation. Chouard and MacLead (1976) reported recognition of speech and popular melodic tunes among their subjects.

Clark (1975) recommended implanting a series of electrodes through small openings in the proximal and distal portions of the basal turn and middle Cochlear turn for auditory nerve stimulation. Clark et. al., (1975) also found that electrodes pass more easily from the apex basal ward. than vice-versa and considered it to be the preferred method of insertion.

The major goal of direct eighth nerve stimulation is hearing and good speech discrimination. The implanted patients had greater sense of security as they hear the background noise at normal intensity level. The cochlear stimulation also imparts a rhythm to speech which is of great value in lip reading. Single channel stimulation does not provide good speech discrimination. Independent investigations of other groups have also confirmed these findings.

Future of Direct Eighth Nerve Stimulation

The single channel stimulation does not provide sufficient information to understand speech, it has been estimated that eight to twelve channels stimulating different populations of nerve fibres may provide necessary information. Mladejousky et al., (1975) has shown that nerve fibres respond entirely differently to electric than acoustic stimulation in animal studies, hence it is not likely to duplicate the functions of cochlea with artificially produced information.

Hilgly sophisticated cochleare prosthesis will also provide a different type of information than is normal to central nervous system. It is not known if the brain will be able to decode the information provided by prosthesis and bering a good speech discrimination. A congenital deaf person without a prior memory might learn this new code with more facility than a patient with auditory memory who must unlearn one language prior to learning another.

It may be that electrical stimulation of fibres whose characteristic frequency to acoustic stimulation is in speech frequency will not respond in this manner to electrical stimulation necessiating the placement of electrodes into apex. A finer and flexible electrodes will be needed to reach the apical portions of cochlea where speech frequencies are normally perceived. This information might be produced by stimulating eight to twelve discrete group of fibres near the basal end of the cochlea.

Speech signals will have to be divided into eight to twelve band widths, each hand width stimulating a separate electrode. It is naive to think that anything approaching near normal level of stimulation could be achieved, though it is hoped that enough information will be conveyed centrally so that brain can interpret it meaningfully.

CHAPTER 9

REHABILITATION EQUIPMENT OTHER THAN HEARING AIDS

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REHABILITATION EQUIPMENT OTHER THAN HEARING AIDS

Sound amplification units have been used for many years to educate the hearing handicapped people. Watts (1969) said that at the beginning of 19th century, Itard made a systematic approach to use the residual bearing of the hearing handicapped. The types of equipment which are available for deaf education can be dividied into following types of educational sound amplification systems:

(a) The Individual Auditory Trainer
(b) The Induction Loop (Audia Frequency)
(c) The Induction Loop (Radio Frequency)
(d) The Radio Frequency Free Field System
(e) The Self contained individually worn unit.

Individual Auditory Trainer

The individual auditory trainer has two microphones (one for the student; one for the teacher), an amplifier, an attenuator for each earphone and pair of earphones (Figure 4-4). The unit can be battery or mains operated. The model could be portable or earable.

The microphones may be handheld, round the neck type, or desk type. The microphone for the pupil may be attached

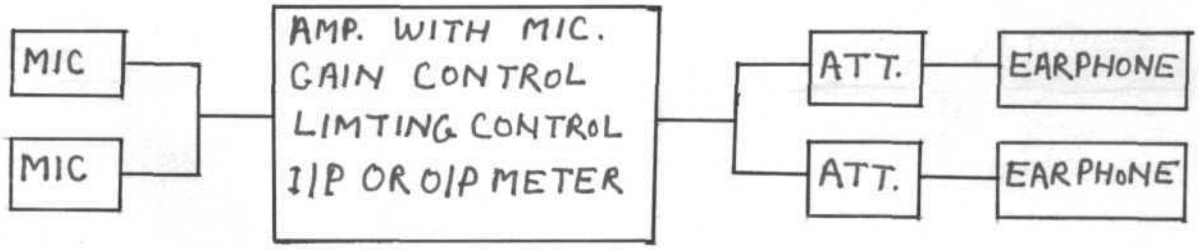

Fig. 4-4 : Individual Auditory Trainer

by boom to the head set. It has the advantage that child speaks at a constant distance from the microphone, but acoustic feedback may cause problem with this type of microphone. Two microphones used should be of similar characteristics and there gain control should also be separate to compensate for any voice differences of student and teacher. Voice level can be monitored by a meter or visual indicator. This can also be used as a positive visual reinforcer for child's responses.

The frequency response of the amplifier should be flat and wide (± 3 dB in the range 50-8000 Hz) and should have provisions for altering this response. Variable bags and treble attenuation controls can also be provided. The amplifier may also incorporate peak clipping, automatic volume control, automatic gain control or compression amplification.

Derbyshire and Reeves (1969) studied the performance of 24 children on three different bone settings of an auditory trainer. They found that the settings predicted to give the best discrimination score did in fact do so.

The attenuators could be continuously variable or could change in 5 dB steps.

The earphone used in the auditory trainer should give extended and substantially flat response. The Telephonics

TDH-39 moving coil is best suited for this purpose. The type of earphone cushion is also important. Considerations should also be given to fitting of the instrument and comfort.

Liltler and Rice (1964) concluded that profoundly deaf child does not need an average SPL greater than 125 dB, or a frequency response higher than 4000 Hz. John (1957) produced an auditory trainer with a response up to 8000 Hz. Watson and John (1960) found that deaf children could benefit from extended high frequency range. Ewing and Ewing (1964) supported the above findings. Dale (1962) advises to use levels up to 185 dB SPL. Ling (1965) stressed the need for amplification below 300 HZ for certain group of cases.

The following additional features may be provided in an auditory trainer:

(i) Input: for tape-recorder, record player, radio or television; an induction pick up coil.

(ii) Output: for vibrator or visual attachment like speech emphasis or 'S' indicator.

Stereo Auditory Trainer

Stereo auditory trainers have two microphones and amplifiers to produce a true binaural effect (Figure 4-5).

Poulos (1950) found improvement in speech perception of deaf children when they listened binaurally instead of

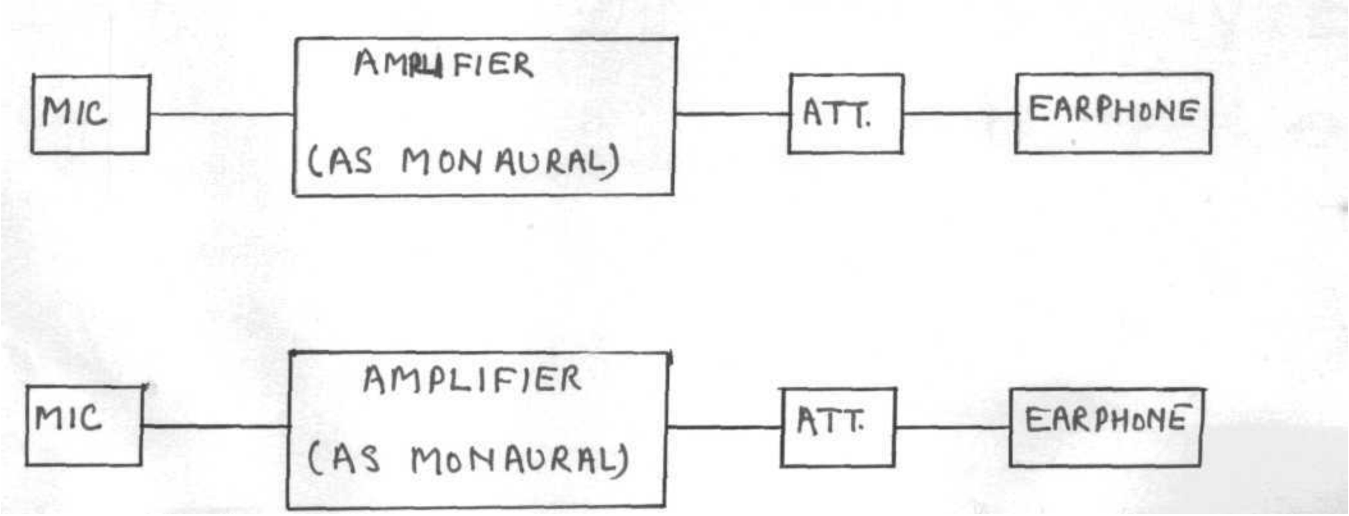

Fig. 4-5: Principle of the binaural Auditory Trainer

monoaurally.

Induction Loop System (Audio-Frequency)

This system is used in a classroom and has microphone, an amplifier, a loop wire round the room (Figure 4-6). Individual aids should have induction pick up coil.

The microphone converts the teacher's voice into electrical signals which is amplified and fed to the loop. When this electric current flows through a wire, it produces magnetic field within the room. This induces an electrical signal in the coil of the hearing aid, which is amplified within the aid and is converted into sound in the earphone.

The teacher's microphone might he designed to operate at a close range. It will reduce the masking effect of the background noise. The teacher can maintain his mobility in the room with the help of long lead. Heath and Lane (1969) recommend the use of fixed microphone, so as to enable the child to appreciate all sounds created within the room. The output of the amplifier should be sufficients produce the magnetic field, which will saturate the aids when the gain (Volume) control of the aid is set at maximum. Martin (1967) reports that the power is sufficient to operate loop systems in the classroom. Many aids are designed for use in magnetic field strengths of either 10 or 100 $m.A.m^{-1}$.

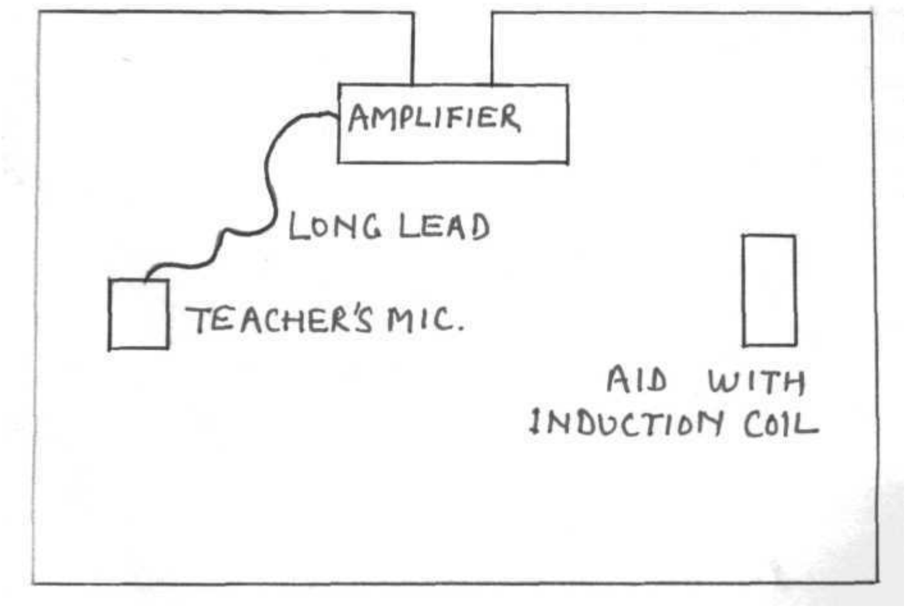

Fig. 4-6: Induction Loop System (Audio-Freqaency)

The loop should be around the walls of the room at the height at which aids are used and its impedance matched to amplifier output. Such a loop, would radiate a magnetic field outside the actual area of a loop for a considerable distance both horizontally and vertically. This makes the use of loop system difficult. The over spill problems are due to excessive amplifier output power.

Most hearing aids have the induction coil in vertical position. These aids are intended to be used in the horizontal magnetic field. The performance of the aid will be impaired, if it is used in any other position than normal one. Aids may be designed for efficient use in any direction. The hearing aid has a function switch which allows the selection of:

(i) "microphone only" operating
(ii) "Coil only" operating
(iii) "microphone and coil" operating.

The choice of above will depend upon the acoustic environment and type of teaching being done.

The main advantages of loop system in a classroom:

(i) The masking effect of background noise is eliminated, because of teacher's closeness to the microphone.

(ii) The distance between child and teacher does not affect the communication.

A charger is provided to recharge the battery of both the units.

Each class in the school for the deaf can be alloted radio frequency. The number of assigned frequencies will depend on the total band width provided by authorities and on the band width used by the equipment. The frequency response of the narrow band width radio systems will be poorer than wide band systems. Figure 4-7 and 4-8 shows Induction Loop System (Radio Frequency) and Radio Frequency Free Field System.

The advantages of the free field RF system are:

(i) Student can monitor their own voices.

(ii) Teacher's voice is received by the pupils at a constant level.

(iii) The units are adjustable as far as possible to suit each student's hearing loss.

(iv) Movements of the teacher and student in the classroom are not restricted.

(v) The system is also operatable outside the school e.g. educational visits.

(vi) Adjacent classrooms can use systems without the danger of interference from each other.

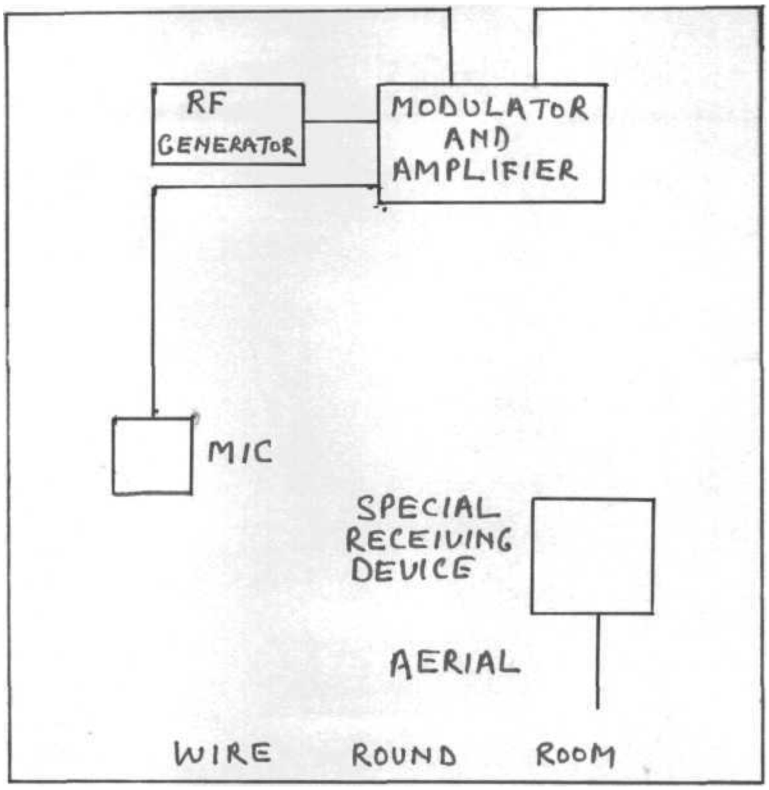

Fig. 4-7: Induction Loop System (Audio Frequency)

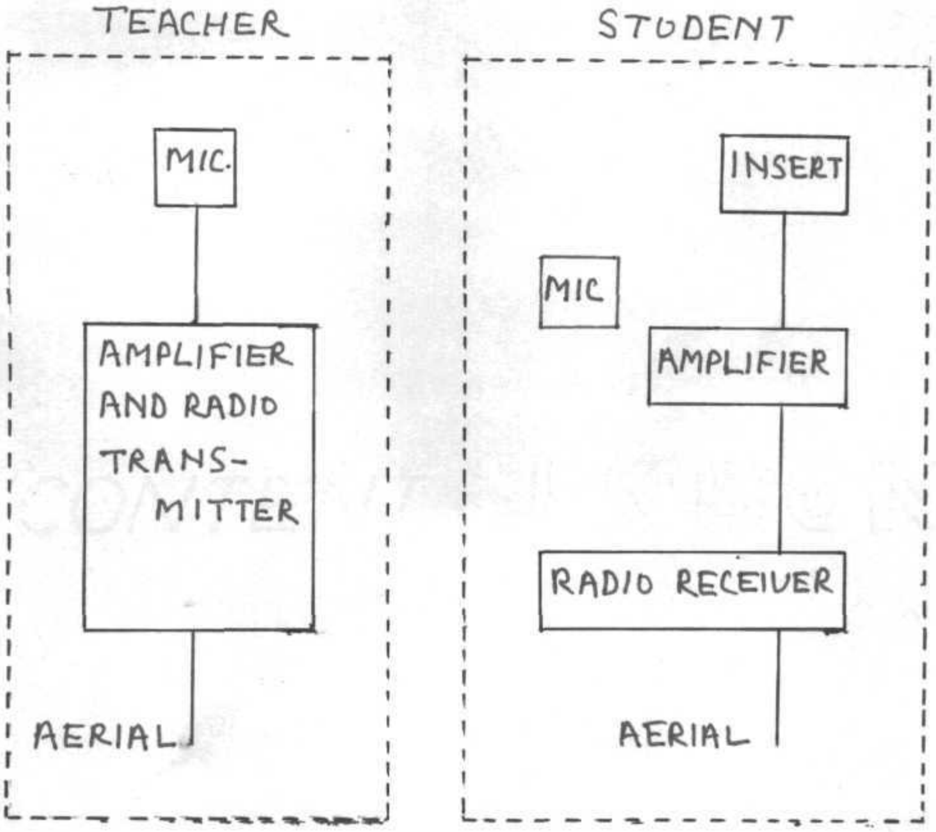

Fig. 4-8: Radio Frequency Free Field System

The only major drawback is that, once the child is out of the school, he will have to change to another form of aid. Another disadvantage is that the child will not be able to pick up his other classmates, if they are out of range of the microphone on his or teacher's unit.

The teacher can have the freedom of movement by substituting the conventional microphone and long lead.

Self Contained Individually Worn Unit

This may consist of a combination of radio microphone receiver and an individual aid (Figure 4-3) or a combination of radio microphone receiver, loop and individual aid which can be worn (Figure 5).

The advantages of this unit is that the child uses the same individual and which has been found to be suitable for him and the additional facilities can be added as the situation requires.

These systems can be used with a radio frequency loop system or a free field radio frequency system. These units can also be separated and the aid used as a normal hearing aid.

Ancillary Equipment

There have been attempts to design the equipment specifically for the needs of the auditory handicapped.

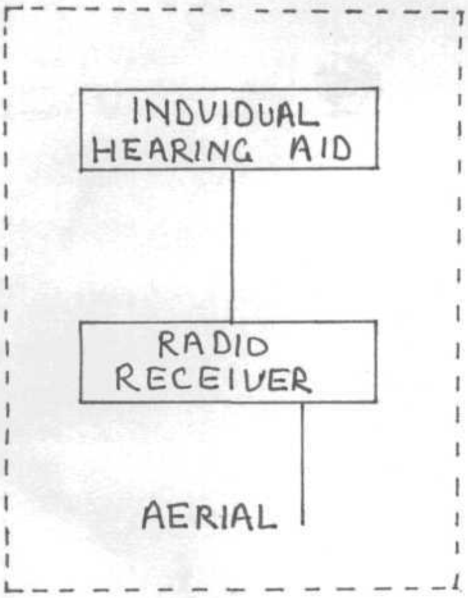

Fig.4-9: Self Contained Individually Worn Unit.

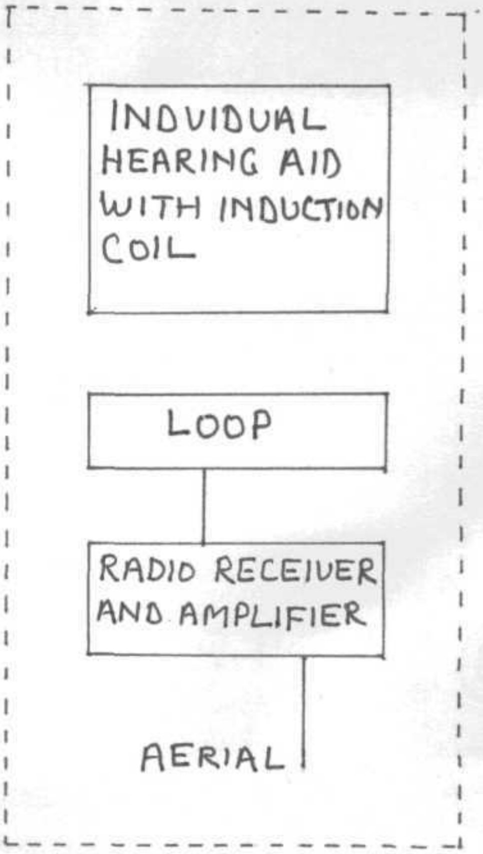

Fig. 5: Self Contained Individually Worn Unit With Loop

These aim at supplementing or replacing the sound signal with visual and tactile information. Watts (1965) and Good (1973) have traced historical and the technological development of these devices.

Visual Devices

Numerous visual devices of variety are available now and they can be divided into two:

1. Which display some specific characteristic of speech, and
2. those which display a more complete speech pattern.

The first category includes:

"S" Indicator:

This can be of use in cases with high frequency sensory neural haring loss. Speech sounds are fed into the microphone, which is then analyzed through selective filtering and the "S" content is visually displayed by a meter indication or a flashing light. In training young children and to increase their motivation, a relay can be used for operation of toys and slide projectors. Risberg (1968), Mortony (1969) and Pickett (1971) found it to be a useful equipment.

The Intonation Indicator

This is employed for training of voice pitch register in deaf persons where auditory feedback of fundamental frequency is not possible. A contact microphone, held against the throat picks up the vibration frequency of vocal folds, and the fundmanetal frequency may be displayed on a meter or an oscilloscope,

The Nasal Indicator

The nasal indicator is helpful in teaching the partially hearing handicapped the aspects of nasalization and denasalization. Here the contact microphone is held against the alae of the nose, microphone picks up the vibrations, amplifies it. The intensity of these vibrations is registered on a meter.

The Rhythm Indicator

The speech rhythm is displayed with the help of microphone, on a long persistance cathode ray tube arranged so that vertical spot deflection is independent of frequency and in direct proportion to voice amplitude. The instrument is important in training the deaf children, as the improper rhythm will affect the intelligibility of speech of the deaf child.

The second category contains visual speech equipment which displays a more complete speech spectrum. Most of the devices use band pass filters to analyse the frequency composition of the speech and display this as varying vertical columns of light. Fant (1960) described a device which provided sixteen columns of light, the height of the column representing the energy within that band width. Secrson (1965) described the use of such apparatus in speech therapy.

Montgomery (1967) reported about a portable visual speech device which produces running visual representation of speech sounds on 7.5 Cm screen of the oscilloscope. The still photographs of characteristic speech are available for comparison.

Gitlits (1972) describes an oscilloscope type of visible sppech apparatus which conveys the rhythmic structure of the speech and also converts the audible speech into symbols.

Tactile Aids

Mulholland (1963) reported that use of equipment to interpret speech signals through the sense of touch has not been of much benefit to the deaf child. Udan (1960) indicating that it is possible to make the child aware of sounds through sense of vibration.

Various models of tactile vibrators consist of microphone, an amplifier and transposer to transpose frequencies within the limited sensitivity range of the skin. Picket (1971) gives this range as 100-800 Hz. The vibrator can be used on the different parts of the body like wrist, the sternum, the clavide and the cervical vertabrae. At the same time amplified sound can also be fed to the child's ears. Neate (1972) who used simple tactile vibrators with selected group of children thinks it worthwhile to use additional techniques while training the deaf.

Recoding Devices

In cases of profound hearing loss, where no useful high frequency hearing exists, equipment has been produced which transposes the high frequencies down to the low frequencies that can be usefully used. Johannson (1959, 1966) recommends a form of transposition that superimposes frequencies above 4 KHZ on the linearly amplified low frequency band. The details of the transposing type hearing aid are shown in the figure.

Block diagram of a commercially available body worn frequency transposing hearing aid is shown in Figure 5-1. Channel 1 allows all signals to pass through while the lower channel (Channel 2) filters off all sounds below 4 KHZ and then effectively transposes these high frequency signals.

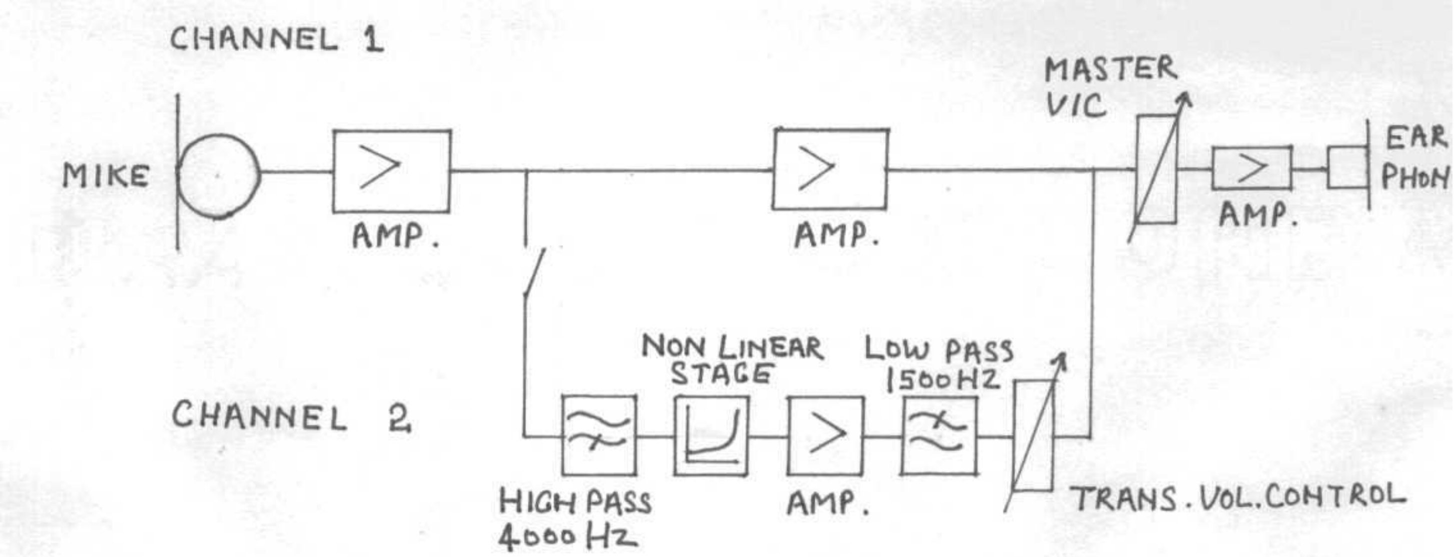

Fig.51 Transposing Hearing Aid.

These signals contains a large amount of low frequency energy which is then passed back to normal channel through a low pass filter and superimposed on channel i. A switch allows the possibility of using the aid as a conventional body worn instrument or as a transposing aid.

Pimonow (1968) has described a device which can be considered an alternative to frequency transposition. Here the input sound is analyzed and than synthesized in such a way that the output is contained within a narrow frequency range.

SPECIAL DEVICES

Signal Detection

A patient who has a loss in low and mid frequencies and greater loss in higher frequencies ("ski drop" loss) often faces a problem and fails to hear the door bell or telephone ring. This problem can be solved either by amplifying the stimulus or substituting it with one which can be easily heard. Various telephone companies may have telephones with different signal bells ranging in frequency from 800 to 4000 HZ. Selection can be made depending upon the requirements of the particular patient. The ringing telephone can also be signalled by an auxiliary control device to activate a flashing lamp. This will be of great help to the severely hearing impaired person.

Another problem for the hearing impaired person will be the inability to hear the sounds from another parts of one's own home. Awareness of specific sound such as door bell or cry of the baby can be enhanced by installing a special electronic switch, which converts sound into either visual or vibratory impulses. From the sound sensing device, lamps or vibrators can be wired to various other rooms. For example, when the baby will cry, the light will flash throughout house, wherever the installations have been made.

Automatic Vakina Devices

Another problem which confronts the deaf is awakening to the sound of alarm clock. Automatic waking devices consists of an electric clock into which, a bed side lamp, buzzer or vibrator can be plugged dependingupon the choice of the user. The buzzer can be placed under the pillow, while the vibrator can be attached to bed frame. Hence the light and vibrator will be able to awake the person.

Electronic wrist watches has been built which when are pre-set for alarm, at the exact time a small spin will come out of the watch as well sound of the buzzer and keep on pricking the skin of the person, till he wakes up and stops it. Such watches can be of definite help to the deaf population.

Radio and Television Listening

Listening to a radio or television viewing can be trying for the family as well as deaf person. The attachment of the earphone to the audio-output can be of great help. A small loud speaker can also be placed near the hard of hearing person's chair. The television audio signal from television sound control can be fed to a portable radio receiver, the gain of which could be adjusted through radio control. But if they are still troubled by other sounds in the house induction coil apparatus can be used and aid switched to a telephone position.

CHAPTER 10

TUBE FITTING

CHAPTER 10

TUBE FITTING

Tube fitting, or fitting of a hearing aid without an ear mould has proved very useful in dispensing the hearing aids. The following are the basic advantages of the tube fitting:

1. When the ear mould is used, the person gets the feeling that his ear is plugged and accluded with something. With the tube fitting, the person even does not feel the tube in place.

2. When the hearing is normal in the low frequency, but there is greater loss in the high frequencies, tube fitting is ideal for this type of 'ski slope' sensori-neural hearing loss.

3. The tube fitting appears to be more 'normal' to the users than the close mould fitting as it leaves the canal open and does not interfere with residual hearing.

4. Tube fittings are quick and easy to make. If the tube fitting can be utilized for particular type of hearing loss, the hearing aid can be fitted immediately, evaluations can be made and user can walk out wearing the aid after fitting.

5. The tube fitting is also used for tinnitus maskers on patients with normal or near normal hearing. The ear is left open to allow the hearing mechanism to act normally, and at the same time masking sound is fed into the ear to assist in tinnitus relief.

Procedure

Following steps may be used to make tubes for tube fitting. Most people develop their own style and methods of tube making after a little practice. The equipment needed for tube fitting is as follows:

(i) a small bowl of water or alcohol trubbing or isopropyl alcohol),
(ii) a hand blower bulb
(iii) electronic solder
(iv) various types of tubing
(v) a hot blower, and
(vi) hand tools such as tweezers, side cutters and needle-nosed pliers.

1. Select and cut off a piece of the tubing desired. It is better to take a bit longer piece than required at this stage. A longer piece than will be used is cut and excess can be trimmed off later.

2. Push a piece of solder into the tubing. This is done to give the tubing stiffness and memory, as it should

hold the ear contours when it is shaped. The solder should be dipped in water or alcohol to lubricate and allow the solder slide easily into the tubing. 0.50" diameter ectronic solder will be the right size into a .076" internal diameter standard tubing. Do not push the solder all the way through the tubing, but recess about l/l6" inside the end of the tubing. This will prevent damaging or scratching the ear canal with the solder when the tubing is placed inside the ear canal.

Heavy wall tubing should be chosen for tube fitting, this will reduce the acoustic feedback and damp out the undesirable acoustic resonance.

3. Place the tubing inside the ear canal and bend up around the pinna. Insert the tubing down to the canal within ½" to 3/8" of the eardrum. Some fitters go further down in the ear canal upto ¼" of the drum. But great care should be exercised, while observing through the otoscope, as to prevent damaging the eardrum. The distance of the tube from the end of the ear drum will affect the acoustics of the fitting. Make an upward and backwards bend in tubing, as to match the contours of the external auditory canal, it is useful to bend the tubing a bit extra towards upwards, so that when the solder is removed, tube should not touch the canal due to its own weight but remain in the centre.

4. When the desired! bend in the tubing is achieved, heat it with hot air-blower to reform it for 1 to 1½ minutes. Move the tube around constantly in the air stream to prevent localized overheating and sagging. Fingers or hand should not be put in the hot air, as it can be quite painful.

5. After heating the tube, drop it immediately into the bowl ofwater or alcohol. Now remove the solder from the tubing using the pliers and blow out the excess water with the hand bulb, some fitters claim that alcohol makes the bend set better than water.

6. As soon as the tube becomes dry, reheat the canal end of tubing with the hot air blower, to round off the end which is sharply cut. This will prevent scratching or irritating the ear canal from the shapp edge. Care should be taken not to overheat the rest of the tube and make it sag. The sharp end can also be smoothed by applying a drop of nail polish on the end of the tube.

If at this stage tube does not conform to the contours of the ears, it should be discarded and another one prepared.

The tube is fitted in place in the ear and the excess is cut off next to the top of the pinner where the end of earhook will fit. If the tube is well-fitted, it will hold a temple or an ear level aid securely. Many persons may feel that instruments will fall off if they bend forwards. This fear can be alleviated by the use of a free field ear mould. The free field ear mould contains just a. think, skeleton of material to fit in the concha. Though when the ear is occluded by any material, it will produce some acoustic effect, but the free field ear mould will bring no change or have minimal effect in a totally open canal. Even at the cost of slight change, a greater security can be provided to the user.

Acoustics

To have relieve and easy to use method of measuring the effects of tube fittings, the usual way of performing acoustic test is by using KEMAR. KEMAR is the Knowles Electronic Manikin for Acoustic Research that represents the physical dimensions of an average adult human ear. KEMAR provides a very convenient way to gather comparative data that can be used to fonn the basis for fitting techniques for all types of hearing aids.

The acoustics of occluding the ear with an ear mould are often not understood clearly. But this can be easily understood from the curves drawn in the figure,fig 5-2A is the free field frequency response with nothing in KEMAR's ear no ear mould tubing or anything else. This shows a frequency curve very close to data derived from the real ear.

The graph in the Figure 5-2(B) is taken under the

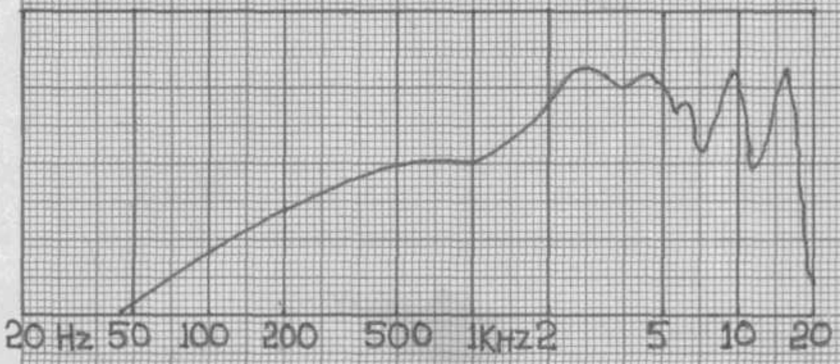

Fig.5-2(A) : KEMAR Free Field Response with unoccluded ear canal.

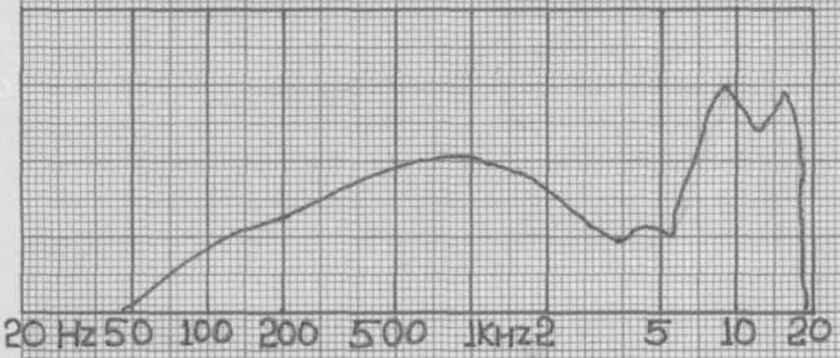

Fig. 5-2(B) : KEMAR Free field response with standard lucite mold with standard bore and no vent.

conditions, but a standard lucite ear mould with a standard bore and no vent is used.

When an ear mould is put in KEMAR's ear, it causes loss of 14 to 24 dB in the area between 2000 and 5000Hz. But this is in speech frequency and which is important for understanding speech.

One of the advantages of keeping the ear canal open, as in a tube fitting, the occlusion loss does not occur. The ear can function normally with the sound passing down the ear while hearing aid assists the residual hearing by providing amplification only where it is needed.

The following figures describe the performance of hearing aid under different measurement and fitting conditions. The figure 5-3 (A) is taken using a 2 cc coupler, under standard ANSI measurements conditions.

Figure 5-3 (B) is the same instrument run on KEMAR using a standard ear mould with a standard bore and no vent.

Figure 5-3 (C) was run on a KEMAR using a tube fitting.

Considering 1000 Hz as a reference point, it becomes apparent that low frequencies have been rolled off and high frequencies have been emphasized. Starting with a flat curve under closed mold conditions, now there is rise of 14 dB from l000 to 2000Hz.

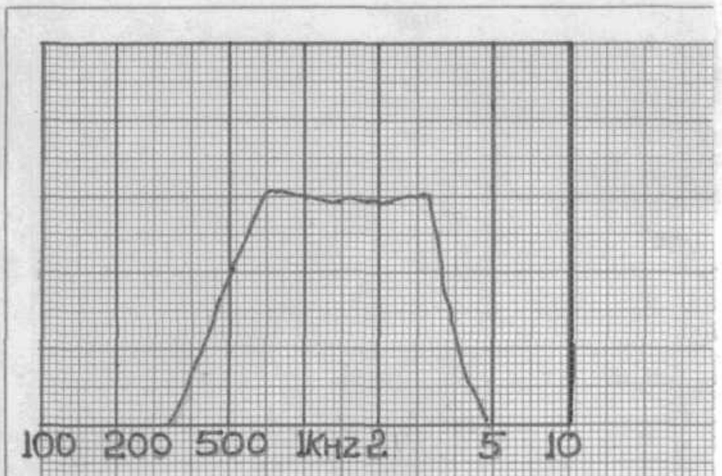

Fig.5-3(A): behind the earmodel run on standard 2 cc coupler,

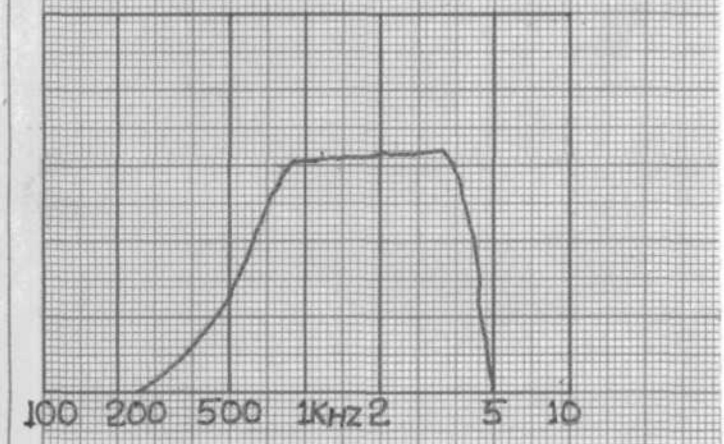

Fig.5-3(B): Same model run on KEMAR with closed ear mold.

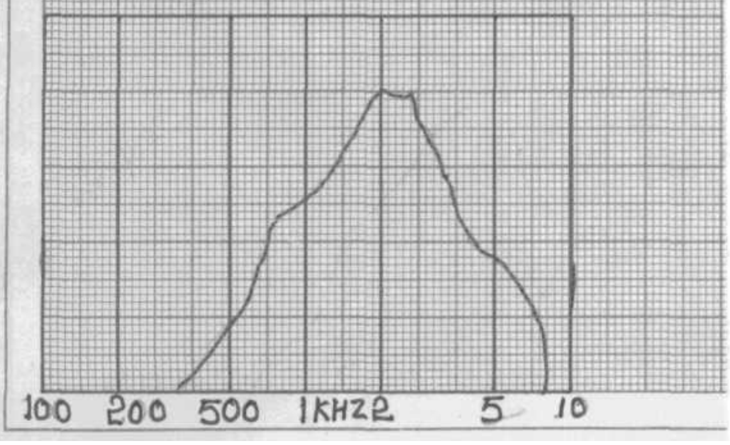

Fig.5-3(C): Same model run on KEMAR with tube fitting.

When an instrument for tube fitting is to be selected, it is necessary that instrument has a flat frequency response on measuring with the 2 cc. coupler. There is no general agreement on how flat the curve should be; but typically, from flat at 2000 HZ (with respect to 1000 HZ) to 6 to 8 dB rise should work well. This can be seen in the figure 5-3(B) and 5-3(C). Figure 5-3(B) has a flat curve but when tube is fitted, the response shifts to that of 2 c. Figure 5-4 shows a similar tube fitting effect to Figure 5-3 for a temple model hearing instrument. Figure 5-4(A) shows the response of KEMAR with a closed mould with a standard bore and no vent. Figure 5-4 (B) is with that of tube fitting. It can be seen from the diagram that under tube fittings, there is rise in the high frequency and roll off in the low frequencies.

There are two factors which can be changed by the filter while doing the tube fitting.

1. Depth that tubing goes into the ear canal.
2. Internal diameter (ID) of the tubing used.

Depth: When tube is put further in the ear canal it serves two purposes. First, in this way moregain is achieved before feedback occurs as the effective length between the receiver and microphone is increased and more of the ear canal provides better baffling of the sound.

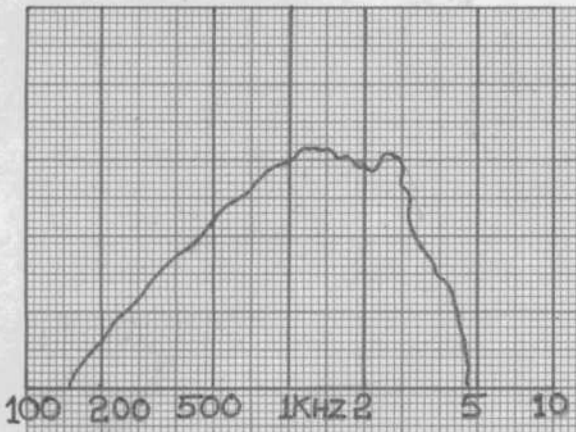

Fig.5-4(A): Temple Model on
KEMAR with standard
ear mould with standard
Bor vent.

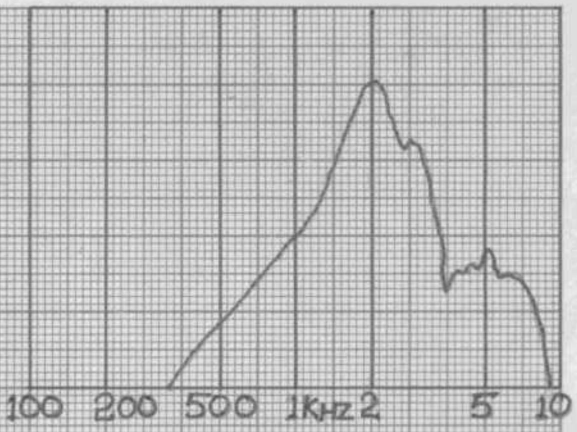

Fig.5-4(B) Same model run on
KEMAR with tube
fitting.

Secondly, as the tube is pushed deeper in the ear canal, low frequencies are emphasized* Usually additional low frequencies are got desirable for tube fitting and it can cause problems some time bow frequencies tend to create a booming or echoing sound for the user. The depth of the tube may become a compromise between the instrument gain and desired frequency response.

One can start at the maximum depth in the canal and then trim the tubing back a small amount at a time till the user reports that the booming sound is absent.

Diameter

Another factor, which can be varied in tube fitting is the diameter of the tube. If the ID of the tube is large, low frequencies will be emphasized and high frequencies dumphasized. This is demonstrated in Figure 5-5B which shows the characteristic of a standard tube with an ID of 0.76" and can be used as a reference curve. In Figure 5-5(A) is the response of .093" ID tube. The build in low frequencies compared to Figure 5-5(B) can be noted here. Figure 5-5(C) shows the response of tube with ID of .053". The,toll off in the low frequencies as compared to Figure 5-5(B) can be seen.

With a little experience and experimentation, various things with the tube fitting can be achieved. Figure 5-6(A)

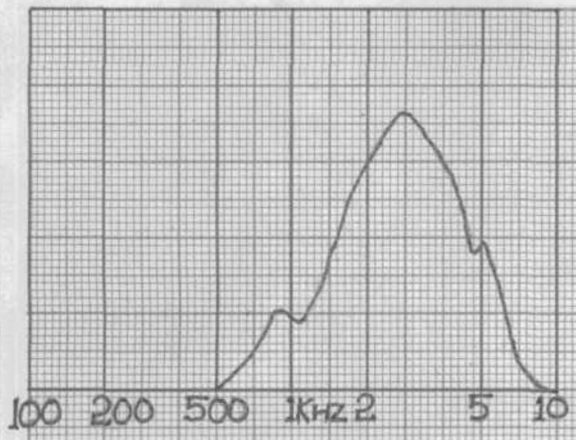

Fig.5-5(A): Behind the ear model run with #ll tubing

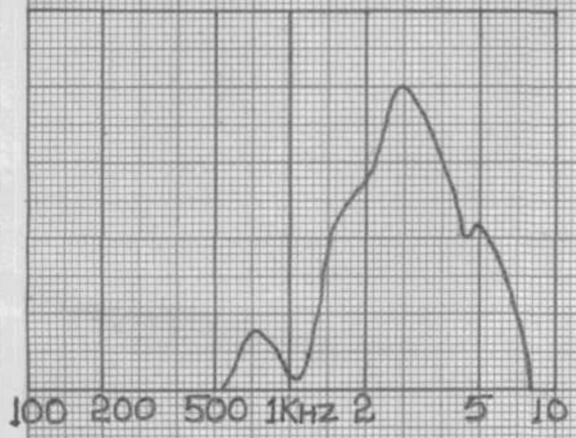

Fig.5-5(B): Same model with # 13 tubing

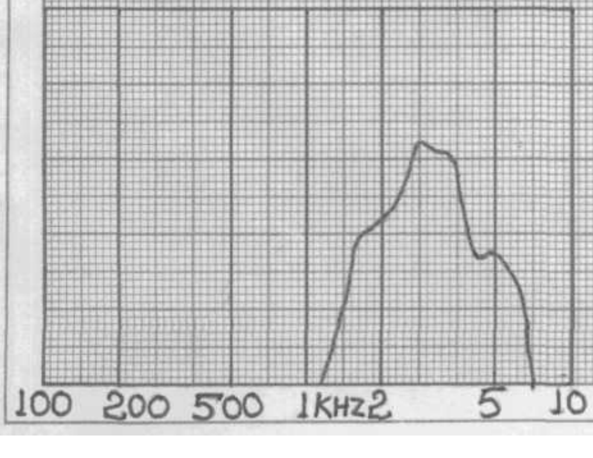

Fig. 5-5(C): Same model with #16 tubing.

is an example of unconventional tube fitting. Here 2 cc. coupler response shows an instrument with greater energy in low frequencies. With tube fitting the response obtained is shown in figure 5-6(B).

This might be useful as a conventional fitting on a person who has a ear discharge and the ear has to be kept open. If the aid has a low frequency gain and it is removed with tube fitting than it will be possible to obtain a conventional type of curve with an open ear canal.

Tube fitting is a useful skill which can be easily applied for gaining various benefits from it.

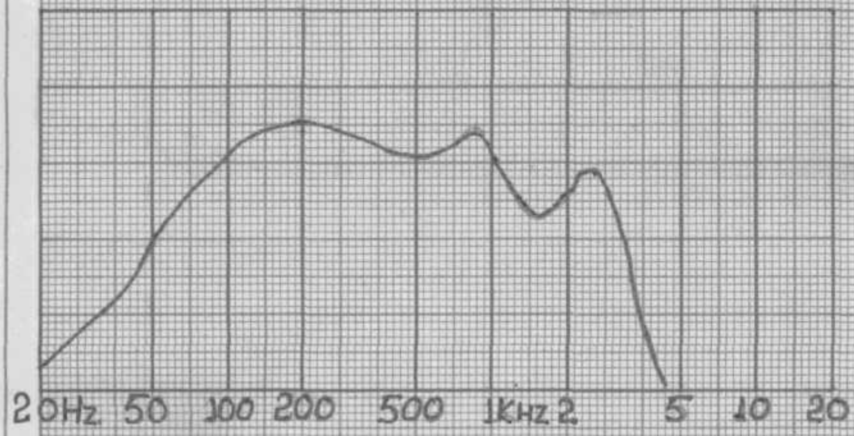

Fig.5-6(A): Low frequency emphasis model run on 2 cc Coupler.

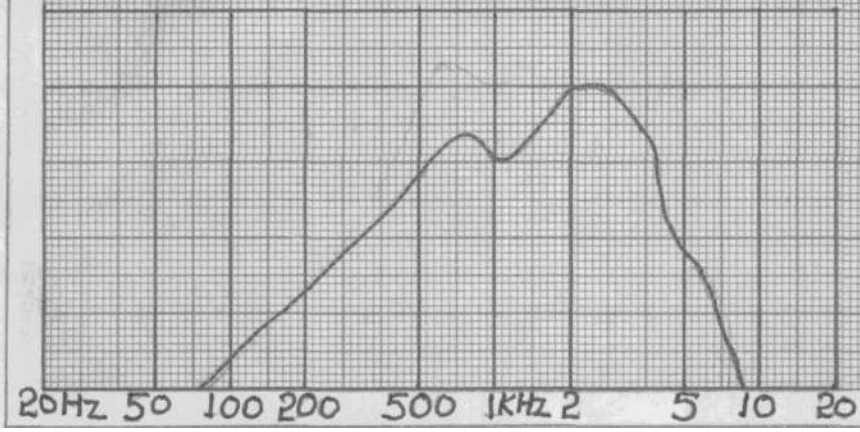

Fig.5-6(B): Same model run with tube fitting.

CHAPTER 11

NEW TECHNOLOGY : HEARING AIDS TODAY AND TOMORROW

THOUGHTS ON THE FUTURE

CHAPTER 11

NEW TECHNOLOGY : HEARING AIDS TODAY AND TOMORROW

THOUGHTS ON THE FUTURE

Advanced level of micro-minaturization technology has brought revolution in hearing aid design, making the size of hearing aid very compact and small. The development of electret condenser microphone has a significant impact on the hearing aid development as it has a broad, smooth frequency response and low vibration sensitivity. This has made possible high gain, forward facing microphone, behind the ear aids and has made all in the ear aids practical. Magnetic receiver technology has reduced overall size of the transducer while providing high frequency range and response smoothness in hearing aids*

Electronic signal processing in hearing aids has improved the amplification of speech signals considerably. Signal processing involves the systematic and controlled transformation of a signal by the system to improve the reception of the signal. Automatic Gain Control (AGC) circuits are important type of signal amplifiers, as they control and modify the speech signal in order to make it suitable for the hearing impaired. When AGC has frequency selective characteristics in addition to input compression circuits low frequency background noise in the environment can be reduced. Frequency selective input compression circuits filter out low frequency energy before it enters the circuit's AGC feed back loop thus preventing the low frequency energy from controlling the gain of the high frequency information carrying portions of the speech signal. If this low frequency filtering can be appropriately combined with input compression to provide a speech signal processing system it will perform well in noisy listening conditions.

The directional microphone can also be used to improve listening in noisy conditions. Lentz (1976) and Beck et. al., (1980) reported measurable difference in directional characteristic among hearing aids. Directional characteristics referred to as super cardiod show greater ability to reduce noise which comes from different angles behind the listener. The super cardiod pattern maintains directionality across the entire frequency range of the hearing aid. This is more desirable than predominantly low frequency directionality especially in hearing aid fitting utilizing large vents or open ear fitting where only high frequency amplification is needed.

The use of improved controlled acoustic damping elements combined with acoustic line principles in hearing aids with extended frequency response has advanced hearing aid technology considerably. Most manufacturers now include damping elements and acoustic filters within the hearing aid or its optional ear hooks to smooth the frequency response or to produce response emphasis in the frequency range between 2500 and 3000 Hz. This principle is commonly referred to as "insertion gain" its purpose being to provide additional amplification in the 2500 and 3000 HZ region so that it will compensate for the loss of natural amplification of the ear canal, when the ear canal is occluded by an ear mould.

The hearing aid industry can benefit from latest technology of microprocessor, as it will be the major component to be used in hearing aids and bring great improvements in the performance of amplification for the persons with hearing loss.

Hearing in noise and acoustic feedback reduces the intelligibility af speech for the person using the aid. Graupe (1979) worked on separating desired voice and other signals from unwanted noise, this signal processing schemes can be implemented by combining both analog and digital signal processing special integrated circuits for hearing aids. These processors will reduce the feedback and background noise and also compensate for amplitude non-linearities controlled response across all or part of the speech frequency range may also be incorporated. It might be possible to individually programme such device with advance of technology.

The technology to develop and produce such a programmable microprocessor hearing aid system is available today.

These systems measure the near stationary noise spectrum during pauses in speech and the spectral parameters of noise are identified during these pauses. This is used to control the response of an adaptive filter which improves signal to noise ratio of the processed speech and noise. Such system can also incorporate additional programmable response filtering, compression with variable compression ratio across frequency, multi-band compression and other parameters found to be beneficial.

It will not be possible to apply this technology fully to wearable hearing aids, as our understanding regarding the characteristics of suprathreshold hearing and its relation to hearing loss and selection of hearing is limited. The designer and programmer of micro-processor-controlled digital signal processing systems must know the signal characteristics that are required to improve the ability of the impaired ear to receive speech signals in quiet and noise. This requires a cooperative and combined effort from the various professionals involved in the rehabilitation of hearing handicapped.

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