CHAPTER 2: REVIEW OF LITERATURE

2.1. Relevant Anatomy & Physiology of the Auditory System

The unique ultra-structural organization of the human cochlea has fascinated researchers for many years, with innumerable studies being performed to understand the complex behavior of the end organ of hearing to various insults, which eventually has lead to the monumental innovation of the auditory neural prosthesis.

The interesting path-breaking discovery that despite congenital or acquired damage to the Organ of Corti due to various causes, the spiral ganglion population within the modiolus survives and remains functional, was the scientific basis upon which the field of cochlear implantation has evolved rapidly to its present day status. Knowledge of the intricate micro-anatomy and patho-physiology of the auditory system remains vital for comprehensively understanding the various electrophysiological and behavioural responses that are evoked by a cochlear implant.

Organization & Function of the Membranous Labyrinth

The compartmentalization of the membranous labyrinth into the Scala Vestibuli, Scala Media and Scala Tympani, provides distinct channels for flow of the endo-cochlear fluids in response to the acoustical impulse, which in turn induce mechanical displacement of the Basilar Membrane, thereby
triggering the Organ of Corti to create electrical nerve action potentials. The cochlear tonotopicity facilitates temporal stimulation of the various regions of the cochlea, according to the intensity and frequency of the acoustical impulse, which get transduced into electrical signals and relay onto the afferent neuronal fibrils and first order neurons in the spiral ganglion.

The Basilar Membrane (BM) extends from the lateral edge of Osseous Spiral Lamina (OSL) to the basilar crest over the Spiral Ligament. It is unique in its dimensions, with an average length of 31.5mm and its width increases from the cochlear apex to base from around 150 to 450 µm. It is microscopically divided into a medial pars arcuata and a lateral pars pectinata. The pars arcuata primarily consists of radial filaments which secure it to the spiral ligament through strong type II collagen with support from specialized cells – the Claudius and Boettcher cells. This arrangement provides the Basilar Membrane with high resilience and tenacity required for optimal displacement with the travelling wave and a frequency specific maximal vibratory property (Clark GM et al, 1988).

The cochlear implant electrode array when placed in situ within the scala tympani, lies underneath and in proximity to the Basilar Membrane. It mimics the natural arrangement of the Basilar Membrane, with the electrodes serially arranged for stimulation according to the ‘place-pitch’ conduction principle (Frijns JH et al, 2001).
The major difference in stimulation via these electrodes is the absence of transduction via Organ of Corti, since sound stimuli externally pre-processed into electrical impulses are directly delivered to their respective regions within the cochlea and trigger the Spiral Ganglia within the Rosenthal’s Canal (bypassing the damaged Organ of Corti) and further conduct these signals to the auditory nerve and onto the auditory brain which perceives it as natural sound signals. Hence, the basic requirement for the success of cochlear implant aided hearing is the presence of surviving Spiral Ganglion population within the damaged cochlea (Hall RD, 1990; Leake PA et al, 1999).
**Fig-2.1**

**Fig-2.1.1:** Internal structure of the cochlea showing alignment of the Organ of Corti, in relation to the Spiral Ganglion within the Rosenthal’s canal and the further formation of the Auditory Nerve Fibers in the Modiolus. The survival of functional Spiral Ganglion population, nearly 35,000 in number (in spite of congenital or acquired damage to the Organ of Corti) is paramount for the success of electrical stimulation with Cochlear Implants.

**Fig-2.1.2:** Ultra-structure of cochlea showing the arrangement of afferent neuronal fibrils; **A:** Apical cochlear turn showing myelinated nerve fibers within osseous spiral lamina (OSL) & **B:** Basal cochlear turn showing Organ of Corti (OC), adjacent to osseous spiral lamina (OSL)

(From: Wright CG & Roland PS, 2005).
Fig-2.1.3: Cross-section of basal turn of the cochlear duct & its schematic representation, showing the internal ultra-structure of the Organ of Corti

(From: Wright CG & Roland PS, 2005).
Pathological Dead Regions in the Cochlea

Dead regions occur within the cochlea where the hair cells and / or the auditory neurons function poorly. A sine-wave that produces peak basilar membrane vibration in a dead region may not evoke sufficient neural activity within that dead region, for that signal to be detected as an acoustic impulse. However, if the signal is sufficiently intense, it may be detected via hair cells / neurons adjacent to the dead region. This phenomenon is termed as ‘off-frequency listening’ or ‘off-place listening’.

Using a frequency-to-place map, the boundary of a dead region can be defined in terms of the characteristic frequency of the hair cells and / or neurons immediately adjacent to a dead region. This is referred to as the edge frequency (fe) and a dead region may have two edge frequencies, an upper and a lower one. However, candidates for cochlear implantation, typically have a dead region that starts at ‘fe’ and extends upwards from there, extending over the basal, middle & apical regions of the cochlea. Psycho-acoustic tests for diagnosing dead regions in the hearing impaired cochlea, typically lead to an estimate of the value of the edge frequencies (Clark GM et al, 1988).

In a similar way, the behavioural & electrophysiological measurements performed through a cochlear implant from the various dead regions within the cochlea, are not alike, since they get influenced by factors like – density of
the surviving spiral ganglion / afferent neuronal population, the electrode-neural interface, gelling effect of sensory elements towards the stimulating electrodes, presence of any intra-cochlear fibrosis or cicatrization due to insertional trauma and spiral ganglion migration towards the electrode array over a longitudinal period of electrical stimulation. Such factors necessitate individual measurement of responses from the various regions of the cochlea (Gluckert R, 2005; Roland PS, 2006).

**Fig-2.1.4: Cadaveric Cochlear Dissection Model:** shown after insertion of cochlear implant electrode array via the Round Window niche into the Scala Tympani. The active electrode contacts are placed immediately beneath the basilar membrane, in order to be close to the modiolus & simulate the tonotopic phenomenon of the cochlea. Today, various types of electrode designs are available - like peri-modiolar hugging / pre-curved, contour & contour advanced, Hifocus Helix, Mid-scalar etc; which favor optimal positioning of electrodes with respect to the neural elements, and provide full
transmission of electrical energy onto the nerve, without causing undue charge-based injury at the contact point. Sound impulses externally transduced into electrical stimuli, arrive at the frequency-band specific electrode contacts along the array and traverse across the basilar membrane onto their respective spiral ganglia and further relay in the auditory nerve fibers at the various sites along the cochlea, thus providing a ‘nature-like’ tonotopic hearing perception (From: Wright CG, Roland PS & Kuzma J, 2005).

The Need for Early Implantation

Research studies using animal models have objectively proved the phenomenon of Neural Plasticity and Scavenging, which happens as the central auditory system undergoes sequential degenerative changes over the duration of hearing loss (Hardie, et al 1998). The presence of hearing in at least one ear provided adequate stimulation in order to preserve the synapses in the inferior colliculus and thus prevent neural scavenging, to an extent (Hardie, Clintock, Aitkin & Shepherd, 1999). But, aberrant connections begin to form between the midbrain and the peripheral auditory system, if hearing is deprived during a sensitive period in early development (Russell & Moore, 1995). Hence, a “Critical Age” for stimulation of the auditory brain exists, which ranges from 1 to 3.5 years in congenitally hearing impaired children beyond which optimal speech & language development may become compromised (Sharma A et al, 2002).
Brown, Tyler and Bertschy, in 1997, studied the influence of age at implantation & duration of hearing impairment, on the outcomes with CI, in a group of pre-lingual children & they reported statistically significant improvement in speech / language skills among early implantees (<5 yrs), as compared to those implanted later (>5 yrs). A further assessment of speech outcomes in CI users has shown that pre-lingual children implanted at a younger age, perform much better than those implanted at older ages (Tobey EA, 2000). The amount of auditory cortical activity thus depends upon the age at cochlear implantation. Recent research has shown that younger implantees with a limited duration of hearing deprivation, have larger areas of auditory representation on their temporal cortex as noted on PET-CT scans & this correlated well with good outcomes as recorded by their Habilitation scores.

Principles of Auditory Neurophysiology & its Assessment

Auditory neurophysiology is complex due to the numerous structural and functional inter-connections occurring between the cochlear nuclei, superior olivary complex, inferior colliculus and the auditory radiations, with approximately 30,000 cochlear nerve fibres, relaying onto a sequence of nearly 10 million neurons in the central auditory system. There are multiple decussations, midline-crossings and an efferent descending auditory network which furthermore make deciphering the signals an enigma. But, the distribution of signals along the cochlea and further onto the higher auditory centers, follows a tonotopic pattern and is based on a temporal integration
principle, which sequentially relays over specific time intervals. Such neuro-physiological responses are influenced by critical factors like age, arborization of the cortex, duration of hearing loss, presence of organic lesions or functional imbalances afflicting the auditory pathway (Kral A et al, 2001; Sharma et al, 2002).

A precise integration of signals along the auditory relay, favours objective monitoring with evoked auditory potentials for confirming optimal functioning of the various centers along the auditory system. Objective measurements in normal or hearing impaired individuals may be performed using near-field evoked responses like ECochG / OAE or far-field responses like BERA, ASSR, MLR / LLR, P300 & CAEP.

In cochlear implantees, a similar battery of electrophysiological measurements, are clinically available to assess the optimal performance of the auditory system in response to signals delivered via the implant. Since, the damaged cochlea is bypassed by the implant, all tests like the ECAP, ESRT, EABR, EMLR, ELLR, P300, MMN and CAEP, are performed as far-field ‘Evoked Telemetry’ responses to intra-cochlear electrical stimulation via the cochlear implant.
2.2. Cochlear Implantation –

2.2.1. Historical Perspective & Current Status

![Cochlear Implant in-situ](image)

**Fig-2.2.1:** Schematic representation of a Cochlear Implant in-situ

The serendipitous discovery of auditory perception following electrical stimulation of the ear, as described in Volta’s experiment, in 1790, has today evolved by leaps and bounds, into the unique realm of cochlear implantation. Following Volta’s cue of the possibility of electrically stimulating hearing, a string of researchers continued to experiment with electrical hearing over the next 167 years, but with little clinical success. Djourno and Eyries reported their first successful stimulation of the acoustic nerve by direct application of an electrode in a deaf person in 1957. Their achievement brought in an overwhelming wave of interest from various parts of the world and soon a string of similar single channel implantations were performed by House, Doyle, Simmons, and others.
The introduction of multi-channel implants by Prof. Graeme Clark in 1967, lead to further advances in micro-electronics and speech processor designs. Over the next fifty years, technological improvements produced refinements in surgery, miniaturization of implants with better electrode designs & precise speech processing strategies suitable for all environments, leading to the evolution of the present day cochlear implant system (Hall JW, 2007). The chronological landmarks which were achieved, during the rapid evolution in the field of cochlear implantation, have been enlisted in the following Table - 2.2.1.

**Historical Landmarks in the Evolution of Cochlear Implants**

<table>
<thead>
<tr>
<th>Year</th>
<th>Inventor(s)</th>
<th>Achievement</th>
</tr>
</thead>
<tbody>
<tr>
<td>1790</td>
<td>Alessandro Volta</td>
<td>Used electrical current to stimulate the inner ear &amp; published his auditory experience</td>
</tr>
<tr>
<td>1855</td>
<td>Duchenne of Boulogne</td>
<td>Used an alternating electrical current produced by a vibratory circuit to stimulate the inner ear</td>
</tr>
<tr>
<td>1868</td>
<td>Brenner</td>
<td>Published the effects of altering polarity, rate &amp; intensity of the electrical stimulation on the placement of electrodes. He discovered that hearing quality was better with a negative polarity stimulus</td>
</tr>
<tr>
<td>1930</td>
<td>Wever &amp; Bray</td>
<td>Demonstrated that the response to the electrodes from the surrounding area of the auditory nerve of a cat was similar in frequency and amplitude to which the ear had been previously exposed to as in nature</td>
</tr>
<tr>
<td>1936</td>
<td>Gersuni &amp; Volokhov</td>
<td>Found that hearing could still persist after the removal of the tympanic membrane and ossicles, therefore giving an opening for the cochlea to be the site for electrical stimulation</td>
</tr>
<tr>
<td>Year</td>
<td>Authors</td>
<td>Description</td>
</tr>
<tr>
<td>------</td>
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</tr>
<tr>
<td>1939</td>
<td>Stevens &amp; Jones</td>
<td>Showed that electrical stimulus could be transduced linearly or non-linearly into sound vibrations before it reached the inner ear. They proved that the middle ear acted as a transducer, electrical energy could be converted into sound by direct effect on the basilar membrane of the cochlea and thus a direct stimulation of the auditory nerve produced a basic hearing sensation.</td>
</tr>
<tr>
<td>1950</td>
<td>Lundberg</td>
<td>Performed one of the first attempts to stimulate the auditory nerve using a sinusoidal current during a neurosurgical operation.</td>
</tr>
<tr>
<td>1957</td>
<td>Djourno &amp; Eyries</td>
<td>Published their first results of direct electrical excitation on the auditory nerve, using a trans-cutaneous magnetic inductive link, which laid the foundation for clinical research in human subjects.</td>
</tr>
<tr>
<td>1961</td>
<td>William House</td>
<td>Implanted two patients with the first prototype of short term single electrode implants.</td>
</tr>
<tr>
<td>1964</td>
<td>Blair Simmons</td>
<td>Implanted a six electrode unit in an adult cochlea for the first time, the success of which proved the place theory of electrical frequency coding.</td>
</tr>
<tr>
<td>1964</td>
<td>Doyle</td>
<td>Reported inserting a linear chain of electrodes into a patient with total perspective deafness, which provided hearing perception.</td>
</tr>
<tr>
<td>1966</td>
<td>Blair Simmons</td>
<td>Performed extensive studies where electrodes were placed through the promontory and vestibule directly into the modiolar part of the auditory nerve. Thus, each nerve fiber representing different frequencies could be stimulated for the first time.</td>
</tr>
<tr>
<td>Year</td>
<td>Inventor/Invention</td>
<td>Description</td>
</tr>
<tr>
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</tr>
<tr>
<td>1967</td>
<td>Graeme Clark</td>
<td>Invented the multiple-channel cochlear implant prototype for the management of profound hearing loss</td>
</tr>
<tr>
<td>1972</td>
<td>Speech Processor prototype developed</td>
<td>Speech processor worked with Dr House's 3M single electrode implant and was made commercially available. 1,000 patients were implanted between 1972 to 1980.</td>
</tr>
<tr>
<td>1976</td>
<td>William House &amp; Michelson</td>
<td>Studied the effects on patients when small electric currents were used on the promontory during middle ear procedures in deaf individuals. Dr. House implanted several devices in totally deaf volunteers, many of which failed due to the lack of bio-compatibility</td>
</tr>
<tr>
<td>1980</td>
<td>House 3M device</td>
<td>Several pre-/ peri-lingual children were implanted with the 3M single channel device in USA</td>
</tr>
<tr>
<td>1984</td>
<td>Multi-channel Cochlear Implant</td>
<td>In Australia, the Multi-channel Cochlear Implant which enhanced spectral perception and speech recognition was developed</td>
</tr>
<tr>
<td>1985</td>
<td>US-FDA approval for 3M House CI</td>
<td>All basic safety concerns about the long term success of CI in adults and children were mostly resolved &amp; efficacy of CI was established</td>
</tr>
<tr>
<td>1990s</td>
<td>Blake Wilson &amp; The Era of Modern Cochlear Implants</td>
<td>Introduced better speech coding strategies in the processor which provided higher performance. Further technological refinements lead to the evolution of a range of newer FDA approved / CE marked cochlear implant devices, from the various CI Companies</td>
</tr>
<tr>
<td>2000s to Present Day</td>
<td>Bilateral CI, Bimodal Stimulation (EAS / Hybrid), Partial Deafness CI</td>
<td>The basic concepts of multi-channel CI remain the same, but a wide range expansion has occurred in the candidacy. Improvements in speech processing technology, has lead to clear hearing in noisy environment &amp; enhanced music perception skills</td>
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</tbody>
</table>
2.2.2. CI Technology & Surgery : An Overview

Cochlear implants have been established as time-tested electronic devices, used to restore hearing in individuals with severe to profound hearing loss, over the last four decades. The last decade has especially seen tremendous progress and refinement in implant technology and surgical techniques for newer devices. The candidacy for CI has expanded by leaps and bounds to include very young children, those with multiple handicaps, a spectrum of syndromic associations & also individuals with partial high frequency hearing loss having residual hearing in lower frequencies.

![Diagram of cochlear implant process]

**Fig-2.2.2.1:** Schematic representation of electrical hearing as provided via a cochlear implant
The Architecture of Cochlear Implants

The cochlear implant system comprises of an external & internal component, both of which are connected transcutaneously with a magnet during implant use. The parts of a CI device include a directional microphone, which receives acoustic impulses from the environment and transmits them onto a speech processor, which in turn converts these signals into frequency specific electrical signals & transmits them as coded signals, via radiofrequency across a transcutaneous transmitter-receiver/stimulator coil, worn on the mastoid temporal bone. The receiver-stimulator coil in the internal system, decodes these signals, in order to produce a pattern of temporally arranged frequency specific electrical stimuli, which get distributed along the electrode array placed within the cochlea. Since this method follows a pattern of ‘place-pitch’ stimulation, very similar to the tonotopic arrangement of the normal cochlea, these electrical signals are perceived by the spiral ganglion and first order neurons of the Auditory Nerve exactly like in normal ears, thereby providing nature-like auditory perception to the higher auditory centers (Laneau J, 2004).

Recent technological improvements like the digitalization of speech processors with high rate stimulation, current steering, stochastic and fine-structure processing of sound signals, have provided enhanced clarity of complex sound signals in all environments and music perception skills for CI
users, to match a nature-like hearing experience. Although the speech processor codes the electrical signals digitally, the transmission of information onto the electrode array needs to be done serially in an analogue manner, in order to comprehensively provide a temporally integrated sound across the entire speech spectrum onto the higher centers. Present day research is focused on this aspect of implant technology, to invent an option of delivering digital sound signals directly onto the electrodes, which is an exciting possibility in the near future, since it may eventually lead to completely implantable digital CI devices with remote programming options, while obviating the need for a radiofrequency interface with an external speech processor (Niparko JK et al, 2009).

Ultra-high resolution CT scans have now documented the enlargement of Rosenthal’s canal, with growth & migration of spiral ganglion population towards the electrode array over a period of implant use. This promising finding provides numerous possibilities for the restoration of neural elements via a cochlear implant, to alleviate intra-cochlear damage in future. Research is hence, focused on the aspects of stem cell therapy and neural regeneration factors, which may be delivered via drug eluting electrode arrays, for promoting hair cell regeneration (Toh EH & Luxford WM, 2008).
Speech Processing Strategies and their influence on Mapping

Sound processing strategies represent a set of rules that define how the speech processor analyzes acoustic signals and codes them for delivery to the cochlear implant. These codes are processed in the form of Spectral Information and Fine-Timing / Temporal Information and delivered to the electrodes as Analogue and Pulsatile stimulus waveforms. A complete stimulating strategy should ideally address the number of channels selected to reproduce the original spectrum, the number of electrodes activated to generate each channel, the number of consecutive clock cycles required to deliver selected channels and the scheduling of the activating sequence of electrodes (Kiefer, 2001; Koch, 2004; Han DM, 2005; Wilson & Dorman, 2008). The older pulsatile strategies like SAS and pre-processed feature extraction (MPEAK) have given way to newer coding methods of peak picking or place-coding (n-of-m / ACE / SPEAK) and the robust rate-coding envelopes (CIS/CIS+/HiRes).

Coding strategies serve to derive stimulation parameters from the input signal. The simultaneous analog strategy (SAS) closely mimics the natural hearing. All incoming sounds are compressed and filtered into eight channels. These channels are then simultaneously and continuously presented to the appropriate tonotopic electrode. There is no need for selecting speech frequencies in this strategy since the intensity is coded either by stimulus
amplitude, rate or both. Due to this fact the SAS strategy has met with limited success. The SPEAK (Spectral Peak) focuses on the spectral (frequency) properties of sound and is based on the place-pitch selectivity of the cochlea. Since it is a roving, place coding strategy, there can be stimulation of selected speech relevant sites along the cochlea, with spectral peaks or maxima providing detailed spectral information. The SPEAK strategy is characterized by filtering sound into 20 different bands covering the range of 200 Hz to 10,000 Hz. Electrodes are stimulated sequentially and at amplitudes specific for each frequency peak. The outputs for each filter are analyzed and those channels of highest amplitude that contain speech frequencies are stimulated. This method also provides cues for consonant perception and for environmental sounds.

CIS (Continuous Interleaved Sampling) is a rate coding strategy, which provides detailed temporal information, but it stimulates fewer, widely spaced and fixed number of electrodes or sites in the cochlea at a higher rate than SPEAK. This system works by filtering the speech into eight bands. The bands with the highest amplitude within the speech frequencies are subsequently compressed and their corresponding electrodes are stimulated. The CIS strategy uses high-rate pulsatile stimuli to capture the fine temporal details of speech. ACE (Advanced Combination Encoders) is a combination of the place and rate strategies, which helps to optimize the amount of temporal information conveyed by stimulating the cochlea at faster rates (up
to 3500 pps per channel). This strategy gives the Audiologist good flexibility to optimize sound processing for each individual. Using ACE, the Audiologist can select the number of stimulation channels, the number of maxima and the stimulation rate per channel. The ACE strategy filters speech into a set number of channels and then selects the highest envelope signals for each cycle of stimulation. The newer coding strategies are focused on Envelope & FSP (Fine-Structure Processing), which have distinct benefits for comprehensive speech understanding, tonal language coding, three dimensional spatial hearing, hearing in noise & quiet and musical notes appreciation. The future of CI signal coding is focused on the development of virtual channels for sequential & parallel stimulation, at a higher rate using a wider input dynamic range, for optimal & rapid processing of complex sounds in all environments, thereby reproducing the original sound spectrum clearly to the cochlear implantee (Kiefer, 2001; Han, 2005; Wilson, 2008).

Today, all strategies are either CIS-based or n-of-m / ACE / HiRes based. It is accepted that no single strategy is effective for all recipients. For this reason, recent software offer several speech processing strategies for the same implant. This allows the Audiologist to choose what strategy is best for that individual. Currently, the Cochlear-Nucleus systems employ SPEAK, ACE and CIS. The MedEl system relies on the CIS strategy and the Advanced Bionics system follows the HiRes-P/S strategies. It is important to adhere to a single strategy for stimulating the implant, while serially
programming an individual, since any alteration in strategy between schedules, will unduly influence the current levels configured into the MAPs and thereby induce variabilities in subsequent Mapping. For a similar reason, it is not possible to compare cohorts using different implant devices, or cohorts using the same device but with different speech processor models, since a variable bias gets induced due to differences in the electrode configuration and / or speech processing strategy, which will eventually provide results favoring the advanced models or strategies, used in the comparison (Miller CA et al, 2003; Polak M et al, 2005).

**Cochlear Implantation: The Surgeon’s Perspective**

With the candidacy expanding to include a panorama of difficult individuals, CI surgeons face a multitude of challenges while performing surgery in recent times. A meticulous assessment of a candidates’ temporal bone anatomy with high resolution radio-imaging and assessment of their associated medical conditions & fitness for surgery under general anesthesia, are paramount in successfully performing the cochlear implant procedure without any untoward incidents (Miyamoto R & Kirk K, 2001).

High resolution CT and MRI scans greatly aid in exploring the intricate anatomy of the temporal bone and help to identify congenital anomalies of the inner ear like an Incomplete Partition (Type-I), Mondini Deformity (IP-II), Large Vestibular Aqueduct, Common Cavity, cochlea-vestibular Dysplasia,
Cochlear Ossification (congenital or post-meningitic sequelae), Rotated Cochlea or an aberrant course of the Facial Nerve in the middle ear. These scans also help in assessment of the vestibulo-cochlear nerve bundle in order to ascertain the candidacy for cochlear implantation and further decide upon the appropriate per-operative preparations necessary for implantation (Phelps PD & Proops DW, 1999). Apart from the routine audiological test battery used to confirm the candidacy for CI, advanced objective electrophysiological tests like the Trans-tympanic EABR and Cortical Auditory Evoked Potentials (CAEP) help to judge whether a candidate with a malformed cochlea & / or hypoplastic / thin VIII cranial nerve will benefit from the cochlear implant or not.

A judicious planning session needs to be held prior to CI surgery for deciding upon the best approach to the cochlea, which type of electrode needs to be chosen, what technique needs to be used for electrode insertion and how to overcome intra-operative obstacles like a CSF gusher, facial nerve exposure or posterior canal wall / annulus dehiscence (Toh EH & Luxford WM, 2008). Surgery is essentially the same in children and adults because the anatomic structures are of adult configuration at birth. However, in very young children, there is an increased risk of facial palsy, hypothermia and hypovolemic shock due to blood loss (more so, with a bilateral simultaneous CI). A detailed counseling session is mandatory for the parents & family, for emphasizing the surgical procedure including details regarding
the risks involved, the techniques of ‘Switch-On’ & Programming of the Device and the need for intensive Auditory Verbal Habilitation / Therapy (AVH / AVT) for a minimum period of one year in order to match their realistic expectations with the eventual outcomes of cochlear implantation.

**Fig-2.2.2.2:** The Posterior Tympanotomy approach for Cochlear Implantation

The success of cochlear implantation depends on scrupulous attention to technique at all the various steps of the procedure. The conventional posterior tympanotomy approach as shown above is the best approach for access to the cochlea. The ultimate goal of CI surgery is to insert the entire electrode array into scala tympani, with as little damage as possible to the ultra-structure of inner ear. This has become possible while inserting newer flexible, atraumatic electrode arrays, via the Round Window. For children with congenital or acquired malformations of the cochlea, like the Mondini dysplasia, common cavity malformation or ossified cochlea, specialized electrode arrays (like straight / short / compressed array / double / split array) are available to provide the best possible intra-cochlear placement of electrodes for optimal stimulation of the viable neural elements within the
deformed cochlea. Hence, it prevails upon the experienced CI surgeons, who take up these challenging cases, to judiciously choose the best electrode type for overcoming the deformity and complete the implantation successfully.

One of the latest applications of implantable hearing technology combines electric and acoustic stimulation (EAS) into a hybrid device designed for individuals with binaural non-progressive low-frequency residual hearing and severe-to-profound high-frequency hearing loss. While performing such a surgery, great care is taken to preserve the residual hearing by administration of Intra-venous Steroids after performing the cochleostomy or exposing the round window niche, prior to insertion of electrodes. A soft-insertion needs to be performed delicately using a special electrode array with care not to damage the neuronal structures within the cochlea (Walkowiak A et al, 2010). Due to the success of this hearing preservation technique with ‘soft-insertion’ of electrodes in candidates with residual hearing, partial deafness cochlear implantation has successfully come into vogue in recent times and bilateral hybrid implantations are also being performed. Thus, cochlear implantation is heralded as a safe surgery with a negligible percentage of complications, while providing complete restoration of the lost sense of hearing & aiding in the development of speech and language skills, thereby integrating CI users into the normal society and leading to a productive life.
2.2.3. CI Programming & Habilitation Protocols

The cochlear implant is ‘Switched-On’ three weeks after the surgery, providing sufficient time for the wound healing and facilitating adequate period of convalescence for the implantee. For pre-lingually hearing impaired individuals, the ‘Switch-On’ is a dramatic event, since they experience auditory perception for the first time in their life. Very young children and those with additional handicaps may develop fear and aversion to this experience and may refuse to wear the implant further-on. Hence, it requires the nuances of an experienced Audiologist to convince such children to wear the implant and set the correct Mapping levels, based on standard observation techniques, periodically in the initial periods of implant use and subsequently fine-tune the MAPs according to the needs of the implantee, as and when required.

The Art of Mapping

For cochlear implant users to perceive the desired range of acoustic signals from their environment, the features of these sounds must control the electrical stimulation within the cochlea in an appropriate way. Low amplitude speech sounds of different spectral structure should elicit soft percepts and higher amplitude acoustic signals should elicit louder percepts while avoiding uncomfortably loud stimulation. As the useful dynamic range for electrical stimulation is relatively narrow and varies across patients and electrodes,
there is a need to tailor the amplitudes of electrical stimulation for each patient. This can be done by assessing the behavioural feedback to psycho-physical & psycho-acoustical stimulation via the cochlear implant, for a wide range of input signals varying in intensity and frequency across the speech spectrum.

Behavioral responses are the ‘Gold-standard’ method for programming cochlear implants and they are sufficient to obtain accurate electrical threshold and comfort levels for the majority of adults and older children using cochlear implants. Although these levels are reasonably accurate at the time of programming, the threshold and comfort levels tend to change over time (Skinner et al, 1995). As a norm, the behavioral levels are low-set at initial Mapping schedules for providing adequate psychophysical perceptive signals to the new implantee, who would seek to understand & familiarize the sounds signals and gradually these levels are increased later-on in a step-wise manner for each electrode along the array, with additional psycho-acoustical inputs, in order to provide an enhanced dynamic range for electrical hearing with loudness scaling, pitch ranking and electrode sweeping properties, as the cochlear implantee becomes more adapted and conducive for higher intensity stimulation, over a period of implant use.
Identifying Most Comfortable Levels form the basis of Behavioural programming in the MedEl & Advanced bionics implants, while their Threshold levels are auto-set by their Map Law, at 10% of the comfort levels to provide an adequate dynamic range across electrodes. The Cochlear-Nucleus implants uniquely follow a different Map Law, wherein emphasis is on Threshold Level based Mapping with individual Comfort Levels being set at around 70% of the Loudness Discomfort Level, for all the electrodes across the array, altering the width of the dynamic range for each electrode, as necessary. Once a series of Maps are created, as per the implantees’
preference, they are incorporated (fitted) into the speech processor as Programs, which control the presentation of encoded sound information through the implant, within the dynamic ranges for stimulation, as set for a particular sound environment.

The Threshold and Comfort Levels obtained for individual electrodes and stored in the memory of the speech processor control the implant’s function and have a bearing on the loudness of the speech signals in most normal environments. But, these levels may not necessarily provide comfortable speech comprehension in noisy environments and hence stimulation with a particular program may be tolerable for a limited time, but could potentially become uncomfortable over a longer period of implant use. This necessitates regular programming sessions, especially during the first year after implantation, wherein attempts are made to provide a diverse range of MAPs, in order to accustom the implantee to various acoustic environments. Watchful observation of the implantees’ auditory verbal skills over a time of implant use provides useful feedback for the Audiologist, to judge whether the program set for the implantee is optimal or not.
Fig - 2.2.3.2: Comfort Level based Mapping technique shown for a MedEl Implantee
The Auditory Verbal Habilitation Protocol

Cochlear implantees are exposed to intensive auditory verbal habilitation soon after receiving their implants, for a minimum period of one year, in order to make them use the implant optimally and in the right way. Habilitation aims at development of new communication skills, rather than just replacing the lost hearing function. Hearing via the implant is used as an active agent for the sequential development of cognition, intelligence, receptive and expressive language skills in a pre-determined, systematic order as per the St.Gabriels’ Curriculum. Periodical assessment of these learned skills, are performed by professionally trained Habilitationists, using a multitude of standard scoring systems. The most popular among these are the Category of Auditory Performance (CAP) and Speech Intelligibility Rating.
(SIR) scores (O’Donoghue et al, 1999); which have an ordinal, non-linear scale for assessment of the auditory verbal abilities of the implantee, taking into account the number of months taken by the implantee to achieve the skills. Further subjective scores like the Meaningful Auditory Integration Scale (MAIS) and Meaningful Use of Speech Scale (MUSS) developed by Robbins et al, in 1991, include the participation of the parents of the implantees, and thereby provide a feedback to the habilitationist, on the parental perspective of the implantees’ communication skills.

As the implantees learn to listen via the implant, they climb through an auditory skills pyramid, from a stage of auditory awareness / sound association, onto the stage wherein they develop auditory processing and comprehension through closed-set and open-set interactions. As this happens, they simultaneously develop their speech skills from a stage of phonating isolated words, onto the formation of full-fledged sentences. Acquisition of enhanced auditory receptive skills and useful levels of spoken language attained through cochlear implants provides an opportunity to integrate into the process of normal education and achieve scholastic skills, which heralds the successful outcome of cochlear implantation.

Habilitation is extremely challenging in children with multiple handicaps and complex needs. Hence, it is imperative for the Habilitationist, to wear a thinking cap and cater to the individual needs of the implantee, by monitoring
the progress of the implantee, and by determining his / her areas of strength & weakness and set goals accordingly. It is paramount for them to work in tandem with the Audiologist who provides the MAP for stimulation via the implant, since any poor performer needs to be troubleshooted at the earliest, for verifying the optimal settings in their MAPs. If necessary intervention with Re-Mapping and enhancement of the habilitation protocols need to be pursued in order to eventually match the expected outcomes of cochlear implantation in such individuals.

2.2.4. CI Electrophysiology – Principles & Methods of IT, ECAP, ESRT & EABR

Prior to cochlear implantation, objective measures like the Trans-tympanic EABR and Promonteric Stimulation (prom-stim) may be used to select the ear for cochlear implantation, among older individuals with prolonged period of auditory deprivation (Zwolan & Kileny, 2004). During cochlear implantation, ECAP, ESRT & EABR, can be used to assess device integrity and to measure the amplitude growth function of the auditory nerve response (Mason SM, 2004). This assures the implant team & the family members, that the implantee is receiving auditory signals optimally via the implant. Post-operatively, these tests aid in objective programming of the speech processor and can also be used as a possible predictor of implant performance. Establishing precise electrical thresholds and comfort levels through behavioral methods is challenging in very young children and those
with developmental delay or multiple disabilities and in such cases electrophysiological measures have taken precedence (Brown CJ et al, 2000; Shallop JK et al, 2000).

**Electrode stimulation modes and array structures**

There are different stimulation modes by which electrodes deliver signals to the cochlea and these modes vary with the different models of implants available today. Monopolar stimulation refers to electrode stimulation with respect to a single extra-cochlear electrode that serves as a reference ground. Bipolar stimulation refers to electrode stimulation in reference to a neighboring intra-cochlear electrode. Dual electrode stimulation is the coupling of two adjacent intra-cochlear electrodes as the active electrode. The dual mode is preferred for electrophysiological studies, due to better neural recording capabilities. The electrode arrays may possess single contact points placed towards the modiolus (as in Cochlear-Nucleus Implants) or paired contact points on either side of the electrode array (as in MedEl Implants). Recent developments have given rise to pre-curved electrode arrays which are intended to closely approximate with the modiolus. These electrode contacts lie closer to the basilar membrane, need lesser current to stimulate the nerve & thereby improve spatial specificity of stimulation. Such variations influence the pattern of electrical flow across electrode-neural interface and therefore impedance to current flow may vary with respect to electrode orientation & proximity to neural elements along the cochlea.
Impedance Telemetry (IT): IT shows the amount of resistance present to passage of current across the electrodes and thereby displays the conditioning of electrode-neuronal interface for receiving electrical stimulation via the implant. IT helps in the measurement of the voltage developed across the active electrodes during stimulation in order to identify electrode anomalies. Electrode impedance is calculated by dividing the voltage at the electrode by the current flow through the electrode. The ground path impedance assessment is considered to be vital for the integrity of the implant. IT can be tested in Monopolar, Bipolar, MP1+2 (Dual) and in Common Ground modes. Hence, it is mandatory to perform IT measurements on all electrodes across the array, prior to each schedule of electrophysiological testing & Mapping (Finley CC, 1990; Liang DH, 1999; Henkin Y, 2003). Impedance Telemetry is routinely performed in all the
available modes by the software and the most conducive mode for electrode stimulation is chosen by default, as per the electrode design & implant model.

**Fundamentals of ECAP Testing:** ECAP Telemetry is a quick and non-invasive way of recording the electrically evoked synchronous responses of the peripheral auditory nerve fibers. The cochlear implant receives the evoked action potentials, by amplifying signals from the intra-cochlear electrodes and these signals are encoded and transmitted back to the speech processor by radio frequency. Thus ECAP measurement is a wireless bi-directional communication of data between the programming hardware and the implant in-vivo. The neural response resulting from a stimulus presented at one electrode within cochlea is measured at a neighboring intra-cochlear electrode.

The ECAP is a triphasic waveform, with a small positive peak P1, negative trough N1 & positive peak P2. The P1 latency lies in the range of 100-300 µsec and the amplitude from N1 to P2 is denoted in µvolts. The negative peak N1 appears at about 220 to 400 µsec following the stimulus onset and this is followed by much smaller positive peaks or plateau (P2) occurring at about 600 to 800 µsec (Abbas, Brown & Gantz, 1998; Cullington, 2000). The twin-peak P1 and P2 waveforms are mainly found on apical electrodes, and may be related to the activity of both the peripheral dendrites and the more central axons (Stypulkowski and van den Honert, 1984; Lai and
Dillier, 2000). Since ECAP is an early latency response, it needs to be separated from the stimulus artifact, using a forward masking and subtraction paradigm which exploits the refractory property of the auditory nerve (Abbas et al, 2000). ECAP is described by the amplitude of the evoked response as a function of stimulation level. ECAP thresholds and slope can be determined from the amplitude growth function and this amplitude increases in an ordinal sequence, with an increase in stimulus level. ECAP testing measures the excitability of the auditory nerve at different cochlear locations and hence, it reflects upon the surviving spiral ganglion population across the cochlea (Brown et al, 1999).

Fig - 2.2.4.2: Basic set-up necessary for performing ECAP (NRI) measurements in an Advanced Bionics Implantee
ECAP measurements are useful in estimating the amount of psychophysical information needed to fit a CI, but may not predict performance / outcomes with the device (Kevin & Susan, 2001). The basic clinical applications of ECAP measurements in the present day include confirmation of device integrity and physiological responsiveness of recipient, selecting the optimal stimulation rate and for setting Maps for very young / ‘Difficult to MAP’ children. Intra-operative

ECAPs are used to verify the integrity of the cochlear implant, to establish proper placement and functioning of the intra-cochlear electrodes and to objectively document the auditory nerve response to electrical stimulation via the CI. Post-operative ECAPs can be rapidly generated electrode-wise across the array and this helps to assist in programming individuals who cannot provide reliable behavioral responses and also for verification of the accuracy of inconsistent behavioral responses, across the array.

ECAP thresholds are believed to fall within the dynamic range of Behavioural MAPs and therefore Audiologists may feel assured to tune up the Behavioural levels based on ECAP thresholds, since that will result in audible percepts for the implantee (Di Nardo et al, 2003; Hughes LM et al., 2000). In many cases, ECAP thresholds have been observed to follow a contour or shape similar to that of a Behaviourally created MAP. When the contour of
MAP levels across electrodes is different for T-levels versus C-levels within a subject, the ECAP will often mirror the shape of one of those functions, usually the T-levels (Hughes & Vander Werff, 2001). Hence, Audiologists may prefer to set the profile and tilt of an initial MAP, based on ECAP thresholds across the array, since this is a reasonable starting point for creating a first-approximated MAP, based on which they can begin to condition a child to respond behaviorally.

Newer ECAP modules, have added features like automated stimulus parameters, faster sampling of signals, higher amplifier gains with lower noise and better masking / artifact reduction methods, all of which precisely reflect upon the neural activity induced by electrical stimulation at various regions of the cochlea.
Fig - 2.2.4.3: Electrode-wise ECAP (ART) data acquired for a MedEl Sonata Implant
**Fundamentals of ESRT Testing:** It is defined as a neuromuscular electrical reflex mediated through the brainstem, identified by the bilateral simultaneous contraction of the stapedius muscles in the middle ears of an implantee, in response to a direct electrical stimulation delivered to the auditory nerve, by the cochlear implant. In a cochlear implantee, the electrically evoked stapedial reflex can be visually identified by tendon contractions intra-operatively and measured post-operatively using the reflexometer of an impedance bridge, in a manual reflex decay mode, from the ipsilateral or contralateral (non-implanted) ear, in response to electrical stimulation through the implant, provided the normal middle ear integrity is confirmed with tympanometry.

ESRTs show a threshold and also demonstrate amplitude growth up to a point of saturation, similar to Acoustic SRTs. A one-second burst of biphasic stimulus impulses (~250 pps) at a supra-threshold current level, are typically used to evoke ESRTs, through any of the electrode channels of the CI. Stimulation is gradually increased until a sufficient deflection is observed in the reflex decay window and a standard bracketing procedure (with small increments in current level) is used to determine the actual stapedius reflex threshold (Battmer et al, 1990; Hall JW, 2007).

ESRT has a few limitations, like the need for presence of normal middle ear function and integrity of the facial nerve and the acoustic reflex
pathway, which is paramount for eliciting reflexes at all schedules of testing. The presence of ESRT response confirms that the implant is functioning and that the neural pathways of the peripheral auditory nerve and the lower regions of brainstem are intact. ESRT thresholds can be a good estimate of comfort levels for speech processor programming for subjects, who are unable to perform psychophysical behavioural tasks (Spivak et al, 1994).

ESRT modules are present in the Mapping software platforms of the various companies and ESRT measurements have identical stimulus parameters to those of the psychophysical MAPs, including the pulse duration and stimulus repetition rate. This allows a direct comparison between ESRT and behavioral measurements, making the ESRT data potentially more predictive for purposes of fitting the CI. Studies involving experienced implant users have demonstrated predictable relationships between the reflex threshold and perceptual behavioural judgments, which fall between the most comfortable level and level of loudness discomfort (Hodges AV, 2003). Studies on ESRT reflex measured intra-operatively on anaesthetized patients show a less robust relationship with behavioral percepts, but have been demonstrated to provide benefit, as a starting point for setting maximum loudness during programming (Jerger J et al, 1988).

In very young children where performing a conditioned response task is often difficult, obtaining a reflex on even a single electrode, provides
assurance that the stimulation being used to condition the child is clearly audible. ESRT thresholds tend to over-predict comfort levels in the initial periods of implant use especially in children and programming such subjects, using ESRT alone may lead to over-stimulation of the implantee (Spivak et al, 1994; Brown CJ et al, 1996).

Hence, a consensus of opinion has been derived from results of various research workers, whereby initial ESRT based maps are to be made by setting maximum stimulation levels at approximately 20% below ESRT thresholds. With the current generation of multi-programmable speech processors, MAPs made at -20%, -10% and at the reflex threshold can be provided to the implantee at initial programming. Following such a method, should help to avoid over-stimulation via CI, which is of particular importance when fitting very young children (Hodges AV, 2003; Gordon KA, 2004; Hall JW, 2007).
**Fig-2.2.4.4:** ESRT measurements as performed for Subject-A of the MedEl implant group, with electrode-wise psychophysical stimulus delivered by Maestro 4.0.1 software and responses recorded from Ipsilateral / contra-lateral ear, in a manual reflex decay mode, using the Reflexometer of an Inter-acoustics AZ 26 impedance bridge, after confirming normal tympanometry in the test ear.
**Fundamentals of EABR Testing:** It is a far-field recording of changes in the electrical activity in the brainstem, in response to auditory stimulus in the electrical form, provided by the cochlear implant. The EABR consists of a series of vertex positive peaks occurring within the first 6 to 8 ms, following electrical stimulation of the auditory nerve. The general morphology of the EABR is typically similar to that of acoustic counterpart, and the most prominent peak of the EABR is also the wave V. The primary difference between the EABR and the acoustic ABR is that EABRs recede with significant shorter peak latencies.

In general, the absolute latencies for all EABR component waves are observed to be around 1 to 1.5 milliseconds shorter than the typical acoustic auditory brain stem response (ABR) mean latencies (Fig-2.2.4.6.). Typically, wave V of the EABR has latency of 4-5ms, while at the high stimulation levels the latency of the V peak of ABR is 5-6ms. The amplitude of the EABR peaks are greater than the ABR peaks, presumably reflecting the greater neural synchrony associated with electrical, as opposed to acoustic stimulation of the auditory system. Another major difference between both is that the latency of the individual peaks of the EABR changes very little with stimulation level. The lack of change in latency is probably due to the electrical stimulation which bypasses the normal cochlear mechanics and activates nerve fibers directly.
The amplitude of EABR peaks are found to be greater than the acoustic ABR peaks, because of the greater neural synchrony associated with the electrical stimulus, as opposed to acoustic stimulation of the auditory system. But, these relatively larger amplitudes of the electrical signals produced by the implant, makes the recording of EABR difficult, since there is considerable disturbances from the EEG waveforms and artifacts which are also simultaneously triggered by the implant. Thus, it is paramount to prepare the optimal settings necessary for EABR recording, by synchronizing the stimulating & recording modules and by fine-tuning the pre-amplifier settings / frequency band filters as required, in order to negate the super-imposing artifacts / EEG waves, while meticulously studying the EABR responses.

EABR is a reliable, objective alternative to behavioral techniques for estimating threshold and comfortable levels for electrical stimulation in ‘Difficult to MAP’ scenarios. But, the results of threshold estimation with EABR in patients with CI are not as accurate as the estimation of behavioral thresholds with conventional ABR in children using Hearing Aids. One factor affecting the accuracy of threshold estimation with EABR is the electrode location within the cochlea and the neural density at that region. Unlike ABR, the EABR is unable to quantify the intensity of the stimulus accurately, since there is considerable difference in the pitch ranking and loudness scaling properties of the input signals, as delivered by the speech processor in an implantee (as compared to a natural acoustic signal threshold used for
evoking ABR). While ABR identifies the exact threshold of hearing in dB for a hearing impaired individual, EABR thresholds tend to fall much higher than the actual electrical threshold levels to be set in the MAP for the implantee, especially in the basal regions of the cochlea. In most cases, the EABR has lower threshold with shorter latency and larger amplitude for a stimulus evoking the apical electrode sites, while the basal regions require a higher intensity of stimulus, for evoking a similar EABR response (Brown et al, 1994; Hodges, Ruth, Lambert & Balkany, 1994; Shallop, Van Dyke & Mischke, 1991; Truy et al, 1998).

Lusted, Shelton & Simmons, in 1984 compared the electrode sites and demonstrated that scala tympani placement resulted in clearer changes in growth of amplitude for different degrees of neuron loss than electrodes placed outside the cochlea. Hall JW in 1990 reported on measures of EABR of rats, in which he demonstrated a correlation between growth response magnitude and nerve fiber survival. Wave 1 the auditory nerve response showed the strongest correlation, while later peaks of EABR showed poor correlations.

Kilney & Zwolan in 2004, characterized the trans-tympanically evoked, peri-operative EABR and defined its relationship with pre-operative hearing, age and hearing loss etiology on 59 children (10 to 60 months of age) who had received cochlear implants. There was no difference found between
wave V latency obtained from the younger (10-36 months) & older (37-60 months) children. This study highlights the fact that EABR testing is reliable across subjects of different etiology and duration of hearing loss.

A number of investigators have used intra-cochlear stimulation to measure EABR responses. The EABR responses in all these cases are consistent, both across the studies and implant types (Abbas PJ, 1999; Brown CJ, 2000; Mason SM, 2000; Gordon KA, 2004; Davids T, 2008). The difference between pre-implant (TT-EABR / Prom-Stim) and post-implant intra-cochlear EABR studies are likely related to the proximity of the electrodes to the nerve and the consistency of the electrode placement as in the case of intra-cochlear EABR stimulation. The amplitude and latency of wave V of the EABR waveforms need to be plotted as a function of the stimulus current level, in order to identify the exact threshold of EABR responses. The amplitude of the response generally increases with increasing stimulus levels and the latency of the peak tends to decrease slightly.
Fig-2.2.4.5: Basic Set-up & Montage necessary for EABR measurements, shown as performed for Subject-A of the MedEl implant group, psychophysical stimulus delivered to the electrodes by Maestro 4.0.1 software via the Diagnostic Interface Box 2.0 and evoked responses from the ipsilateral ear, recorded using the Intelligent Hearing Systems (IHS) SmartEP (Evoked Potential) software module (Version 3.91USBez), in a synchronized paired computer.

Fig-2.2.4.6: Representative EABR waveforms as evoked via the cochlear implant, showing typical wave morphology & latency patterns and their comparison to an acoustic ABR waveform. Standard differences in wave latencies between ABR & EABR are highlighted in a table (Ref: Abbas PJ, 1991).
Fig-2.2.4.7: A representative EABR threshold measurement shown as performed for Subject-K of the MedEl Combi40+ implant group, with the psychophysical stimulus delivered by Zebra DOS software (version 3.0) and evoked responses recorded from the implanted ear, using the Intelligent Hearing Systems (IHS) SmartEP (Evoked Potential) software module (Version 3.91USBez).

EABR Stimulation of different electrodes within the implant can result in different sensitivity, but all show similar changes of amplitude and latency with the current level. This facilitates the acquisition of similar data from adjacent electrodes located in the same region of the cochlea which is being tested, thereby providing an option of selecting any electrode from an offset along the array for EABR analysis, for predicting MAPs for that region of the cochlea (Hall JW, 2007).
**Standard Parameters used for Electrophysiological Measurements**

*(An example of default parameters shown from the Cochlear-Nucleus Implant Group)*

<table>
<thead>
<tr>
<th>Impedance Telemetry (IT)</th>
<th>Stimulation Modes = CG, MP1, MP2 &amp; MP 1+2; Stimulus Current Level = 100 micro-volts; Pulse Width = 25; Recording: 0-30 k-ohms</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Evoked Compound Action Potential (ECAP)</strong></td>
<td><strong>Probe:</strong> Stimulating Active Electrode = Screening (Mid Array ~ 14) + Full Sweep of All Electrodes / Alternate Electrodes; Indifferent Electrode = 1-22, CG</td>
</tr>
<tr>
<td><strong>Neural Response Telemetry (NRT)</strong></td>
<td>Stimulating Mode = MP1; Stimulating Current Level = 1-255 CL; Stimulation Pulse Width = 25-400 microsecs; Stimulation Pulse Gap = 25; Stimulation Rate = 5 - 400 Hz</td>
</tr>
<tr>
<td>Electrically Evoked Stapedial Reflex Telemetry (ESRT)</td>
<td>Measurement Options:</td>
</tr>
<tr>
<td>---------------------------------------------------</td>
<td>----------------------</td>
</tr>
<tr>
<td><strong>Method 1</strong> – Intra-operative Visual Recording</td>
<td>Current Levels = 1-255 CL</td>
</tr>
<tr>
<td><strong>Method 2</strong> – Impedance Bridge Testing in Ipsilateral / Contralateral Ear</td>
<td>Pulse Width = 25-400 µs</td>
</tr>
<tr>
<td></td>
<td>No. of Pulses / Burst = 1-10,000</td>
</tr>
<tr>
<td></td>
<td>Stimulus Interface Gap = 6-60 µs</td>
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<td></td>
<td>Stimulation Rate = 100-8000 Hz</td>
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<tr>
<td></td>
<td>Stimulation Duration = 0-5000 ms</td>
</tr>
<tr>
<td></td>
<td>Stimulus Repetition Rate = 0.005-1000 Hz</td>
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</table>

<table>
<thead>
<tr>
<th>Electrically Evoked Auditory Brainstem Response (EABR)</th>
<th>Stimulation defaults:</th>
<th>Measurement defaults:</th>
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<tbody>
<tr>
<td></td>
<td>Single cycle pulse trigger</td>
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<tr>
<td></td>
<td>Indifferent electrode: MP1</td>
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<tr>
<td></td>
<td>Stimulus current level = 150 - 220 CL</td>
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<tr>
<td></td>
<td>Stimulus pulse width = 25 - 100 µs</td>
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<td>Stimulus inter-phase gap = 7</td>
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<tr>
<td></td>
<td>Number of pulses per burst = 1</td>
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<tr>
<td></td>
<td>Burst stimulus duration = 0.057</td>
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<tr>
<td></td>
<td>Burst repetition Rate = 35 Hz</td>
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<tr>
<td></td>
<td>Stimulus Rate = 21.1/milliseconds</td>
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<tr>
<td></td>
<td>Stimulated Electrodes = 1, 14, 22</td>
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<tr>
<td></td>
<td>Electrodes placed on Vertex (Cz) / high-forehead (Fz) / CL Mastoid (Ref) and low-forehead (CG)</td>
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<tr>
<td></td>
<td>RF free period: 10 ms</td>
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<tr>
<td></td>
<td>Number of sweeps: 1000</td>
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<td></td>
<td>Averaging: Basic / Alternating polarity</td>
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<tr>
<td></td>
<td>Artifact rejection: Disabled</td>
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<tr>
<td></td>
<td>Trigger delay: 0 µs</td>
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<tr>
<td></td>
<td>Sweeps / Artifact = 1024/1 to 1413/60</td>
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<tr>
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<td>PP Amplification = 14 - 21</td>
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<td>SNR = 0.7 - 3.2</td>
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<td></td>
<td>Pre-amplifier Gain = 60 - 100 dB</td>
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<td>Recording Mode:</td>
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<td>Bipolar / Multi-polar</td>
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<td>Alternating Stimulation via CI Trigger pulse</td>
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<td>Artifact Subtraction by Polarity Reversal</td>
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<td>Filters = 100-1500Hz</td>
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<td>Peak Records = 1.4-7.8 ms</td>
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<td>High Pass Filter = 1-150Hz</td>
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<td></td>
<td>Low Pass Filter = 3-5 KHz</td>
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<td></td>
<td>Amplification Rate = 50,000 to 100,000</td>
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<tr>
<td></td>
<td>EAP Signal Record = 1-10 milliseconds</td>
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2.3. Literature Survey:

Recent Concepts & Practices in Cochlear Implant Electrophysiology & Mapping

Technological improvements in cochlear implantation, including the development of newer speech coding strategies and programming modules, in addition to the advances in surgical techniques and the clinical expertise available today for comprehensive habilitation, have improved the outcomes & quality of life in CI users. The present day CI programming, has become more dependent on higher stimulation rates, provided by the new age speech processors which favor enhanced auditory perception abilities, hearing in noise, speech clarity and music perception skills. The speech coding strategies adopted and implemented in the various types of cochlear implants, offer different stimulation rates, which needs to be optimized by the implant audiologist according to the candidates’ preference.

In clinical practice, choosing an optimal rate for stimulation becomes a trial and error method based on behavioural inputs from the candidate, but younger children may not be able to identify the best rates for stimulation, since a rate which offers best quality of sound, may not offer the best speech perception. In a similar way, it may often be difficult to distinguish between an optimal MAP versus an effective MAP, based only on behavioural responses to high rate stimulation via the cochlear implant (Skinner MW, 2000 & Overstreet EH, 2004). Auditory perception at high stimulation rates may also
be influenced by the refractory property and recovery function of the nerve, which affect the perception of loudness via temporal summation, since a fast recovering nerve fiber will transmit more impulses than a slow one. To overcome this mismatch, newer implants have been incorporated with automated stimulation rates derived from inbuilt formulas, based on the pulse width and averaged electrode voltage. The default rates for stimulation with automated pulse widths, thus maintain the same charge throughout the electrode array. This helps to protect the electrodes against any physical damage which may occur due to stronger current stimuli, beyond the electrodes’ compliance limit.

**Impedance Telemetry:** Studies have shown that impedance levels reflect upon the integrity of the electrode-neuronal interface, but do not necessarily have any significant influence on the Electrophysiological thresholds and Mapping levels. A high impedance value in an electrode does not provoke a higher EP threshold. The possible reason for this finding is that the nerve responses to the electrical stimuli are controlled by a spectrum of factors like the refractory property, recovery function, neuronal density at the site of stimulus and influences of higher centers on the auditory nerve (Saunders E et al, 2002; Henkin Y et al, 2003). The Impedance levels across the array, gradually stabilizes over time with implant use. This is possibly due to the gelling effect of electrodes with the tissue interface and neural reorganization around the electrode array, which facilitates better conduction of electrical impulses with lesser resistance as the auditory nerve becomes more
conducive to electrical stimulation over time (Abbas et al, 2001). In practice, it has been observed that the impedance values across the electrodes keep varying by a few Kilo-Ohms with time, but they need to be within the normative levels for optimal conduction of electrical impulses, without undue resistance to the passage of current. It is a routine practice that Implant audiologists check the impedance telemetry, both intra-operatively after placement of the cochlear implant in-situ and also prior to each schedule of Mapping / before attempting an electrophysiological measurement, in order to make sure that all electrodes are conditioned for electrical stimulation and thus confirm the implant integrity at the time of testing.

**Need for Post-operative Electrophysiological Testing**

In the past, researchers compared both intra-operative and post-operative electrophysiological tests, especially ECAPs for setting MAPs and an initial consensus was derived from results published by brown CJ et al, 1994; Parkinson AJ, 2002; Hill K et al, 2004; and Mason S, 2004. These Authors proposed the use of both intra-operative and post-operative ECAP tests for setting initial MAPs, since their studies had shown that there was no statistically significant differences between the two longitudinal measures and hence both were equally useful during Switch-On & at subsequent initial Mapping schedules. These authors also suggested that, subjects who do not produce sufficient ECAPs at the time of ‘Switch-On’, possibly due to low-set
comfort levels with reduced tolerance to loudness of the stimuli, may benefit from goodness-of-fit MAPs created based on intra-operative ECAP profiles alone. This concept has been commercially utilized in recent implant software, which has provision for incorporation of intra-operative ECAP thresholds into the Mapping program when necessary. Hence audiologists do not feel the need for performing post-operative ECAP tests, unless required as in ‘Difficult to MAP’ situations.

But, intra-operative tests do not truly reflect upon the behaviour of the Auditory Nerve at the time of the cochlear implant ‘Switch-On’ and programming. There are significant variations in the current levels of various electrophysiological tests post-operatively due to factors like; wound healing, reduction in neural tissue - electrode interface (the gelling effect), alteration of electro-chemical gradient within cochlea, neural re-organization, conditioning and adaptation of auditory nerve and the influences of higher auditory centers on the nerve’s response to electrical stimulation, all of which are absent while the subject was tested under General Anesthesia, in the operating room.

A series of researchers at the Cooperative Research Center (CRC) for Cochlear Implants in Melbourne, Australia analyzed the relationship between intra-operative and post-operative ECAP thresholds and their application for programming the Cochlear-Nucleus Devices, during the last decade. Their studies have proved that post-operative ECAP profiles are more effective in
Mapping, than an intra-operative ECAP profile. A similar opinion has also evolved from other studies on intra-operative versus post-operative ESRT & EABR profiles & their clinical application for implant programming (Battmer RD et al, 1990; Kileney PR et al, 1991; Hodges AV et al, 2003).

In a comprehensive research work done by Hughes LM et al, in 2001; a longitudinal analysis of electrical impedances, ECAPs and behavioural measures was performed in a group of Cochlear-Nucleus 24 CI users. The primary goal was to study the changes in these measures over time and compare the results between 35 adults and 33 children involved in this study. The various parameters were sequentially recorded in these cohorts up to 2 years of implant use. The results showed that changes in impedance telemetry, ECAP thresholds and Amplitude growth function of ECAPs were similar among adults and children, up to the first 2 months of CI use. Beyond the first 2 months, children exhibited significant increase in electrode impedance, ECAP thresholds, slope and rising psychophysical T-levels, while these same measures were relatively stable in adults. The major inference from this study is that, if ECAP thresholds need to be used for Mapping, it is best to make these measurements at the same time intervals as the device programming, rather than using the ECAP measurements made at the initial programming session (Switch-On) or earlier (intra-operatively), for setting the Mapping levels at later visits.
Hence, there is a paradigm shift in the consensus of opinion in the present day that, a one-time confirmation of optimal implant function assessed intra-operatively with electrophysiological measures, does not impress upon the post-operative behavior of the Auditory Nerve to electrical stimulation via the cochlear implant, which may vary over a period of time. Thereby, post-operative electrophysiological tests need to be performed during the habilitation period or beyond, in order to predict optimal Mapping levels, whenever a ‘Difficult to MAP’ situation is encountered in the clinical scenario (Brown CJ et al, 2003; Hill K et al, 2004; Gordon et al, 2004; Polak M et al, 2005; Caner et al, 2007 and Basta D et al, 2007). Most implant manufacturers have provided in-built electrophysiological test modules, in their programming software for this purpose, but they do not emphasize their use, unless when a trouble-shooting situation is encountered.

Implant companies are yet to provide a clear-cut, comprehensive tool guide or flow-chart algorithm, for the optimal application of these objective measures in real-life ‘Difficult to MAP’ situations, which adds to the dilemma of the Implant audiologist, who may often face such a practical situation in clinical practice. A possible reason for the above scenario may be the paucity of uniform & consistent results while using electrophysiological measures. The realm of post-operative electrophysiology is still evolving, with new data emerging from reputed CI clinics across the world, providing new insights into this intriguing entity, with ample scope for further research.
The Enigma of Mismatch between
Electrophysiological Thresholds & Behavioural Levels

The three clinically available objective measures, namely - ECAP, ESRT & EABR have been successful in providing working MAPs for cochlear implantees, but all of them have also shown certain drawbacks when clinically applied, as discussed in the literature below.

A) ECAP versus Behavioural Levels: ECAP measurements are the most popular, user friendly, rapidly performable method for providing reference values for Mapping, but they do not predict the optimal behavioural levels for setting in a MAP, due to various reasons as below. ECAP thresholds tend to be higher than the behavioural threshold levels and may fall in-between threshold & comfort Levels, within the dynamic range of the MAP, but in 1/3rd population of implantees ECAPs may also exceed the comfort levels especially in the basal electrodes (Smith et al, 1994). Strong correlations have been reported between ECAP thresholds and psychophysical behavioural thresholds, while weaker but significant correlations have been found between ECAPs and comfort levels in various studies by Brown CJ, 1996 & 1998; Abbas PJ, 1999; Hughes LM, 2000; Franck & Norton, 2001.

The obvious mismatch noted between the ECAP thresholds and behavioural current levels, has been implicated to the different rates at which
the stimulus is delivered to the Auditory Nerve via the implant. ECAPs are evoked at much slower stimulation rates, in order to clearly identify the morphology of the nerve action potentials and to study the properties of amplitude growth and recovery function of the nerve. Behavioural programming is done at higher stimulation rates by fast temporal processing of sound impulses, in order to deliver the complete sound signal without any loss of information and thus provide the best auditory perception. Hence ECAP levels cannot be directly incorporated into the speech processor as programming levels, without any available behavioural inputs from the cochlear implantee.

A fundamental study done by Abbas PJ et al, in 2001 established the fact that ECAP measures alone are not very reliable for implant programming. Abbas & his colleagues revealed chronological changes in ECAP (NRT) over time from the day of surgery in their study group of 36 children, using the Cochlear-Nucleus Implants. Statistically significant changes in the NRT thresholds of these children were observed until 3 to 8 months following initial stimulation. Measures of NRT slope in the children did not stabilize even until 12-months post-implantation. These Authors opined that longitudinal trends in NRT measures followed the threshold levels more closely than comfort levels in the Nucleus implant group which they studied but, NRT did not reflect upon the vital factors like an expanding dynamic range and rising maximal comfort levels, which are paramount for optimal implant fitting over a period of implant
use. Smoorenburg, et al in 2002, proved that the recovery function of the auditory nerve when analyzed via the Nucleus NRT software, helped to define the stimulation levels needed to program initial MAPs in infants and children. They found it useful to create offsets between T-NRT and T-Levels across the array and thus help in defining the tilt of the MAP. They concluded that ECAP thresholds, have a positive correlation with comfort levels, but cannot be used to predict the overall comfort levels and tilt of the comfort profile, since prediction of slope of the maximum stimulation curve, which is the most critical factor in speech perception, from the ECAP thresholds was quite poor.

Many reasons have been documented in literature for the poor match between the ECAP thresholds and the Behavioural Levels to be programmed in the MAP. Factors affecting ECAP profile include – electrode impedance on the probe and recording electrodes, radial distance of the probe and recording electrodes from the modiolus, actual focus of contact between the electrode & the neuronal elements, neural density at the point of stimulation, refractory period of the nerve fibers, masker probe induced artifacts, voltage measurements across the electrodes, demographic variables such as etiology of the hearing loss, duration of auditory deprivation, implantation age and duration of implant use (Pfingst BE, 1999; Rebscher SJ, 2001; Morsnowski A, 2003; Shpak T, 2004).
Researchers like Almqvist B (1998), Miller CA (2000), and Morsnowski A (2003), aimed to identify the goodness-of-fit between ECAP thresholds and Behavioural profiles, in order to establish a common correction factor for incorporation into the fitting method, in ‘Difficult to MAP’ candidates. But, they were unable to arrive at a consensus regarding standardization of a correction factor based on offsets formed between the reference T-Levels and their corresponding ECAP thresholds, since they found that subjects showed a range of inter-electrode and inter-patient variabilities which occurred over time, during their studies.

**Effect of Stimulation Rate on the relationship between ECAP & Behavioural levels**

Frank & Norton, in 1999 have reported that there is no one-to-one relationship between ECAP thresholds and behavioural measures used to program the speech processor, since they found that these recordings followed no specific patterns and considerable variability was found across the electrodes and among the subjects. The following reasons substantiate their observation. The degree to which electrophysiological thresholds of the auditory nerve could replace or supplement behavioural testing methods, depend upon the pulse rate of the stimulus used for electrophysiological and behavioural measurements.
The rate of stimulation at which ECAPs are measured is necessarily limited by the time needed for each recording to be elicited on the software and by the refractory properties of the nerve. ECAP measurements arise from the auditory nerve activity within the millisecond immediately following each current pulse. Thus, the pulses must be spaced far enough apart in order to allow this measurement to be free from additional stimulus artifacts. If the rate of stimulus is very high, the amount of activity evoked by each pulse would be reduced by refractory effects of the nerve, making the response more difficult to detect.

On the contrary, behavioural levels used to program the implant, are generally measured at relatively higher pulse rates, ranging from 250 to 1000 pulses per second, depending upon the speech coding strategy being used. SPEAK strategy uses 250 pulses/sec, CIS strategy uses 1200 pulses/sec and ACE uses 900 pulses/second. This high stimulation rate tends to decrease the behavioural threshold and comfort levels, since the psychophysical responses to this high stimulation rate depends not only on the evoked activity for each pulse in a pulse train, but also on the neural activity integrated over a period of time on summation of a number of stimulus pulses. The electrophysiological thresholds do not temporally integrate in the same way as the behavioural thresholds and hence, there lies a disparity between these two thresholds, even when measured on the same electrode at the same time (Brown CJ et al, 1999; Skinner MW et al, 2000).
Brown CJ et al in 1998, compared ECAP thresholds to psychophysical threshold levels in Nucleus implantees, for single biphasic pulses and found a good correlation of $r = 0.89$ between them. She inferred that ECAP thresholds measured at 35Hz are more related to behavioural thresholds obtained for 500 millisecond pulse trains of 35Hz, than to behavioural thresholds for 500 millisecond pulse trains of 250Hz. Subsequent studies that compared ECAPs and behavioural thresholds showed correlation ranging from $r = 0.5$ to 0.6 in adults (Smoorenburg et al, 2000) and between 0.2 to 0.7 in children (Hughes et al, 2000; Gordon et al, 2002).

Zimmerling and Hochmair, in 2002 measured post-operative ECAP thresholds at 80Hz and behavioural thresholds and maximum comfort levels at 80Hz and 2020Hz in 12 Ineraid implantees. They concluded that ECAP thresholds at 80Hz were highly correlated with behavioural threshold measures at the same rate ($r=0.89$), but correlated less with behavioural threshold measured at 2020Hz ($r=0.6$). Comfort levels moderately correlated with ECAP thresholds at 80Hz ($r=0.5$) but correlations were poor at 2020Hz ($r<0.2$). Murray & McKay, in 2005 reported that correlations of ECAP thresholds with behavioural measures decreased with increasing stimulation rates. Their data showed correlation values ranging from $r = 0.94$ at 35Hz rate, 0.7 at 250Hz rate and 0.6 at 1800Hz rate respectively. They also noted that ECAP thresholds were located higher in the behavioural dynamic range as the rate increased, possible due to the fact that higher the rate,
behavioural threshold and comfort levels become lower. Fewster and Dawson, in 2005 also reiterated this fact in their work, where ECAP thresholds fell close to comfort levels when high stimulation rates were used. Basile et al, in 2004 reported that in general, ECAPs obtained at low stimulation rates are better in morphology and amplitude than those acquired at higher rates.

The electrode design and rates used in the speech coding strategy have a significant effect on the behavioural thresholds and ECAP measurements (Eisen & Franck, 2004; Han DM et al, 2005; Polak M et al, 2005). Researchers have also proved that speech perception is better for psycho-acoustically created behavioural Maps, than for NRT based Maps with behavioural adjustments (Smoorenburg GF et al, 2002; Seyle K & Brown CJ, 2002). Hence, the clinical application of ECAPs for device programming is often limited to identifying the minimal thresholds required for nerve stimulation at initial Mapping, but later psycho-acoustical feedback with speech stimulus is mandatory in order to redefine the ECAP based Map, according to the subjects needs.

Brown CJ et al, in 1998 reported that correlations may improve by using additional limited behavioural data along with electrophysiological measurements. Initial trials with a raw and random electrode-wise correlation between these measures provided inconsistent and varied correction factors.
across the array. Later, Hughes LM et al, 2000 & Smoorenburg GF et al, 2002; showed that behavioural threshold and comfort level data obtained from a single middle electrode on the array can be used to create offsets with its corresponding NRT level. Hughes further described that the difference in current level (NRT vs T-Level and NRT vs C-Level) recorded at the mid-electrode, may be used as a correction factor and applied to the NRT values on other electrodes for predicting optimal behavioural threshold and comfort levels across all the electrodes. This method improved correlations between the measures in his study from 0.6 upto 0.9. But, since electrophysiological & behavioural responses are not similar across the array, and may widely vary between the apical and the basal array, this method was not found to be clinically very successful.

In order to improve the correlations across the array, Gordon et al in 2004, investigated with a combination of two separate correction factors, one for the basal and mid-array and another for the apical array since, greater variabilities had been observed in the apical array leading to lower correlations in the previous studies. Using this method Gordon could obtain improved correlations across the array, of $r = 0.95$ for ECAP vs T-Level and $r = 0.93$ for ECAP vs C-Level. The mean average difference between the actually measured behavioural levels and statistically predicted behavioural levels across the array, reduced to lesser than 10 programming units by this method. Gordon’s study proved that it is better to divide the electrode array
into offsets preferably as apical array, mid-array and basal array and select one representative electrode in each array for performing correlations with their behavioural levels. This method provides three correction factors across the array, thus increasing the accuracy of predicting optimal behavioural levels for each region of the cochlea independently. Since ECAP measurements when used alone without any behavioural inputs were not able to provide a reliable MAP, researchers looked further into other available objective measures like the ESRT & EABR for aiding in Implant programming. In general, it has been observed that basal electrodes require higher current levels for stimulation, than the apical electrodes. The higher behavioural comfort levels noted in the basal array imply that louder impulses are required to address the basal region of the cochlea, which has higher density of spiral ganglions and codes for higher frequencies of auditory stimulation.

A study by Hughes LM & Abbas PJ, in 2005 analyzed the electrophysiologic channel interaction, electrode pitch-ranking and behavioural thresholds among two cohorts of Cochlear-Nucleus implantees, using ECAP measurements and they concluded that there was no statistically significant difference with respect to the ECAP measurements, noted between the straight array and the peri-modiolar, contour advanced array groups, even though the contour advanced group possessed lesser electrode channel interaction and improved pitch-ranking ability. Such a panorama of behavioural levels required to be set across the electrode array, cannot be
provided comprehensively by ECAP measurements alone, since they do not reflect upon the accurate psycho-physical / psycho-acoustical properties of the electrical impulse, processed via the cochlear implant. This necessitated the use of ESRT & EABR which reflected better on the comfort levels needed to be set in the MAP, due to their higher intensities of stimulation. As the concept of Map Law evolved to include, charge based electrode programming with comfort level based fitting, ESRT has taken precedence in setting MAPs in recent times.

B) ESRT versus Behavioural Levels: ESRT measurements have been commonly used for predicting comfort levels especially for very young, pre-lingual, hearing impaired cochlear implantees, in whom identifying the optimal loudness of the electrical signal is paramount for providing the best fitting program. It is known that in the initial period of implant use, ESRT thresholds may over-estimate the comfort levels and they may be a good indicator of maximum comfort levels, rather than most comfortable levels. Hence, audiologists setting an ESRT based initial MAP for an uncooperative child, must be cautious in order to avoid any Mapping level above the ESRT thresholds, which may induce an uncomfortable response to acoustic stimulation in the child and aversion to further implant use. At later stages of implant use, ESRT levels may fall in close proximity to the most comfortable levels (Spivak et al, 1994; Hodges et al, 2003; Stephan et al, 2000).
ESRT has been found to be of greater value than ECAPs for the estimation of behavioural comfort levels (Gordon et al, 2004; Han et al, 2005; Caner et al, 2007). Investigators have concluded that ESRT show high correlations with behaviourally obtained comfort levels and help to predict the maximum comfort level pattern across electrodes (Jerger J et al, 1988; Stephan K, 2000; Walkowiak A, 2010). Fitting the speech processor based on ESRT data has been shown to result in speech perception scores equal to or better than those achieved with conventional fitting techniques (Shallop JK, 1995; Almqvist, 2000, Bresnihan M, 2001). In general it has been observed, that implantees using ESRT based MAPs have lesser discomfort and better preference to wear the implant in loud environments. But, especially among very young children, initial fitting measures with ESRT alone has not been very successful due the inherent nature of ESRT thresholds to over-estimate the comfort levels, which may result in the setting of too loud a MAP in such children, thereby inducing an aversion for implant use among these children (Walkowiak A et al, 2010 ; Van Den Abbeele T et al, 2012).

Unlike ECAP & EABR, ESRT thresholds are found to significantly rise over a period of implant use, in accordance with the increased tolerance to higher levels of stimulation shown by increasing comfort levels and an expanding dynamic range among cochlear implantees (Gordon KA et al, 2004). Accurate estimation and fine-tuning of most comfortable levels and loudness balancing are of greater value, while applying an ESRT based
method for programming young children, rather than construction of an overall MAP profile, only using ESRT measurements. Hence, ESRT has been the popular tool of choice, to aid in MAP stabilization and to avoid over-stimulation among CI users with fluctuating / inconsistent comfort levels, which can happen due to a greater stochastic ability (Shallop JK, 1995; Ferguson et al, 2002; Miller CA et al 2003). By and large, ESRT based prediction methods are found to be more useful than ECAP based prediction methods, for setting Psychophysical behavioural levels for pediatric implantees, though their comparisons with comfort levels are significantly influenced by individual variabilities among the subjects, with regards to their age at implantation and duration of implant use.

C) EABR versus Behavioural Levels: Literature suggests strong positive correlations between acoustic BERA thresholds and subjective behavioural thresholds for specific acoustic stimuli among normal individuals (Gorga et al, 1985; Coats & Martin, 1997). Similarly, research has also shown that EABR thresholds have good positive correlations with behavioural responses for the same electrical stimulus delivered via the cochlear implant (Abbas PJ & Brown CJ, 1991). Hall JW in 1990, found a strong correlation between the amplitude growth of wave 1 of EABR and the number of surviving spiral ganglion cells. This correlation was found to be much weaker for the subsequent waves in the EABR. Hall also described that wave 1 of Jewett’s potentials corresponds to the ECAP potential and therefore ECAP thresholds
are directly proportional to the number of auditory nerve fibers activated at the site of stimulation. This was a possible explanation for the ECAP thresholds falling close to the psychophysical behavioural threshold levels, while the EABR waveforms usually require a stronger stimulus higher than the threshold current level, since the responses are evoked from the brainstem.

Brown CJ et al, in 1999 reported reasonable positive correlations for EABR with maximum comfort levels, and recommended its application in the setting of MAPs. In her study, she observed that EABR thresholds fell approximately two-thirds of the way between the subject’s behavioural threshold and comfort levels. But, she also emphasized the existence of considerable individual variabilities in her series, between the subjects and in the same individual over a time of implant use. She attributed this variability to the temporal integration of the electrically synchronized stimuli, across subjects. CI users who were able to perform excellent temporal integration of the high rate sound signals, had larger differences between their Behavioural levels and their EABR thresholds, than those subjects, who did not process their high rate signals well enough. Hence, Brown concluded that in order to improve the correlations between EABR and Behavioural levels for a specific electrode, it was necessary to estimate the temporal integration ability of the CI user.
In a subsequent study, Abbas PJ et al, in 2000 opined that EABR threshold patterns mostly tend to remain unchanged during the first year of implant use and hence EABR may be useful for objective programming of implants, throughout the period of habilitation. This observation favored the inclusion of EABR into the electrophysiological test battery provided by the various Implant manufacturers. Further research focused on the assessment of EABR properties at various sites along the implant array. Gordon KA et al, in 2004 observed that EABR thresholds were higher than ECAP thresholds across the array, due to the need for a higher energy of stimulation via CI that is required to elicit a recordable action potential from the brainstem. Gordon inferred that a pattern of gradual ascent in EABR thresholds may be noted from the apical array towards the basal array, possibly due to higher neuronal tissue density in the basal regions of the cochlea, which need more current for a cumulative response from the brainstem.

It is accepted that EABR is more reliable than other objective measures for predicting Behavioural levels, since it is more consistent across the array and remains stable on longitudinal assessment over a period of time. But, EABR has not found much acceptance in the practical scenario, since it requires technical expertise with an advanced set-up and a cooperative implantee. It is also found to be cumbersome, time-consuming & impractical to be done electrode-wise, in order to comprehensively program a cochlear implantee (Abbas et al, 2000; Brown et al, 2003; Shpak et al, 2004).
Need to Combine Objective Measurements for Predicting Behavioural MAPs

Since each of the objective measures (ECAP, ESRT & EABR) showed inherent disparities as above while predicting behavioural MAPs, researchers arrived at a consensus of opinion, to combine these measures into a test battery, in order to obtain more consistent MAPs. Gordon & colleagues, in 2004 pioneered the concept of combining the three objective measures for achieving optimal cochlear implant stimulation levels in children. They observed that a combination of such non-behavioural measures can aid in the determination of useful cochlear implant stimulation levels, particularly in young children and infants with limited auditory experience.

Gordon proposed that in order to overcome the disparities among these measures while predicting MAPs, appropriate correction factors need to be generated, based on individual correlations with available behavioural levels and then may be clinically applied to produce an optimal MAP. She suggested that correction factors to predict threshold levels should be based on ECAP thresholds and EABR thresholds, and maximum stimulation levels must be determined by using the ESRT thresholds. She believed that such correction factors need to be made for at least one apical and one mid-array / basal electrode, taking into account the age of the child and these factors may have to be revised during the first year of implant use.
Need for Correlating Electrophysiological Measures with Behavioural Responses

Electrophysiological measurements recorded from the Auditory Nerve are quite different from the behavioural responses observed in the cochlear implantee, mainly due to the fact that the electrical nerve responses recorded from the peripheral auditory system does not primarily match with the behavioural responses to the same current level of stimulation, since they are influenced by the auditory processing, which happens at the higher auditory centers. Electrophysiological measurements are usually performed at default stimulation parameters that are different from the stimulation rates eventually used during cochlear implant programming.

Sensitivity and neural reactions recorded to electrophysiological stimuli are bound to be different from the behavioural reactions recorded at higher rates of stimulation, used while programming (Shepherd RK & Javel E, 1997). A higher stimulation rate is used in Mapping for optimal processing of stimuli necessary for speech comprehension, while a lower stimulation rate is preferred while performing electrophysiological measurements, since accurate neural thresholds can thus be identified (Craddock et al, 2004; Kaplan-Neeman et al, 2004; Gordon KA et al, 2004; Davids T et al, 2008).
Behavioural response elicited by electrical stimulation with a cochlear implant electrode, is understood to be the result of a combination and superposition of the following phenomena occurring at three different levels:

(Level-1) Electrode - tissue impedance and positioning of the electrode contact towards the neural tissue. The higher thresholds for electrophysiological responses at the basal electrodes are possibly due to the physical current distribution among the more dense network of neurons;

(Level-2) Neural preservation and excitability status and refractory properties of the auditory nerve fibers and (Level-3) Cortical and behavioural reactions to the excitation patterns in the higher auditory pathways as influenced by the age at onset of deafness, cognition, intellect, hearing aid usage and duration of hearing deprivation prior to implantation.

All electrophysiological measurements clinically used like the ECAP, EABR & ESRT objectively record events occurring at levels 1 and 2, yet take no account of the variability present at the higher auditory centers. This necessitates the need for their correlation with any available behavioural level, in order to be able to optimally predict further behavioural levels across the electrode array.

Behavioural responses are immensely influenced by higher auditory circuits and electrophysiological measurements of the peripheral auditory system alone cannot substitute or replace a behavioural MAP accurately.
Behavioural responses to stimulation via the implant vary widely between very young children and older children, wherein factors at level-3 play a major role and there also exist inter-personal variabilities between subjects. Inappropriate Behavioural Mapping in infants / toddlers may over-estimate comfort levels, as compared to older children, since these subjects are unable to provide correct auditory feedbacks to the psychophysical stimuli provided by the implants. Hence, age may also be a factor influencing the variability of the behavioural levels across subjects (Sharma A & Dorman M, 2002).

The advent of Cortical Auditory Evoked Potentials (CAEP) and Positron Emission Tomographic CT-Scan (PET-CT), have today provided some objective insights into the interesting events occurring at the level 3, with respect to age at onset of hearing loss, duration of auditory deprivation, lingual status, cognitive mental functions, age at implantation and duration of implant use, among pediatric cochlear implant users.

Research with cochlear implant aided Acoustic & Electrically Evoked CAEPs / PET-CT imaging by studying their correlations with the behavioural levels / outcomes among a spectrum of pediatric cochlear implantees, may probably provide the way forward in future to solve the mismatch which exists, while applying current methods.
Electrophysiological Thresholds versus Behavioural Levels:

An Example of Mismatch

The following Figs - 2.3.1 & 2.3.2, represent the mismatch noted between the electrophysiological thresholds and the behavioural comfort levels (M-Levels) at different points of time of implant use, found in a cochlear implantee of the study group (Subject A) using the Advanced Bionics HiRes 90k implant with Harmony speech processor. The NRI, ESRT & EABR threshold values have been directly plotted onto the graph for a comparison with their respective M-Levels (y-axis = current units – CU & x-axis = electrode array El 1- apical & El 16 - basal electrode).

Fig - 2.3.1: Plotted after ‘Switch-On’, at 1 month of implant use in Subject A, shows that; EABR thresholds (measured on 3 electrodes across the array) lie close to the Behavioural M-Levels, ESRT thresholds (measured on 4 electrodes across the array) over-estimate the Behavioural M-Levels, while NRI thresholds (measured for all electrodes across the array) fall on a mean average of around 65% of the Behavioural M-Levels and are variable from electrode to electrode.
Fig 2.3.2: Plotted at 12 months of implant use in Subject A, there is more proximity between the ESRT & EABR thresholds and Behavioural M-Lowels, but NRI thresholds still fall on a mean average of around 80% of the M-Lowels and are variable from electrode to electrode.

These graphs highlight the practical fact that direct incorporation of electrophysiological thresholds especially ECAPs onto the MAPs for fitting a subject with inconsistent behavioural responses (which is the popular method followed in clinical practice) is fallacious at any time of implant use, and will produce sub-optimal performance with habilitation, unless such MAPs are further fine-tuned with behavioural inputs subsequently. The existence of such disparities, reiterates the need for a correlation between electrophysiological and behavioural measurements, based on which a statistical method for predicting optimal behavioural levels can be developed, prior to clinical application of the electrophysiological parameters for implant fitting. The present research work was focused on developing such a clinically useful statistical prediction method, which would provide a way for optimal Implant fitting.