In spite of the continuous progression in cochlear implant technology there is still a broad field for investigation in the direction towards the achievement of a more physiological hearing.

House and Urban (Los Angeles) are considered to be the first authors, which implicated cochlear implants on a significant number of patients – about 100. These were single channel implants with transcutaneous information transmission. In 1967 another scientist begins investigations over the possibility for electric stimulation of auditory nerve - Graeme M. Clark (Melbourne), presently director of Australian Bionic Ear and Hearing Research Institute. Inspired by the work of Simmons, he starts theoretical research over multichannel cochlear implants introduced through the round window. In the beginning of the eighties, Clark operates his first patient. He is one of the pioneers of modern cochlear implantation.

The German otologist Zollner together with the physiologist Keidel, are the first to outline the characteristics of CI, which are important for achievement of auditory perception. The principles defined by them are conceptual for modern cochlear implantation:

1. Transcutaneous transmission of information.
2. Platinum electrodes with diameter of 0.35 mm
3. Intracochlear placement of the electrode array.
4. Definition of frequency areas from 300 to 3000Hz.
5. Number of contacts: 20 to 100.
6. Stressing on the importance of tonotopic frequency distribution for sound coding.

Cochlear implants have gradually advanced from the single-channel devices to the modern multichannel ones. The place coding of frequency with multiple-electrode implants is the main advantage of these systems over the single electrode implants. Single-channel stimulation can produce only periodicity pitch, and information transfer is insufficient for speech discrimination. For the optimal place coding of frequency is very important the combination of stimulation mode, electrode geometry and interaction with ganglion cells.

Modern Cochlear Implants differ in number of electrode contacts and coding strategies. Most of them can use more than one coding strategy and both ways of stimulation: monopolar and bipolar. In monopolar mode, the current runs between one active electrode and a referent electrode located retrocochlearly. In bipolar mode it travels between two adjacent electrodes – one stimulating and the other - referent. Bipolar stimulation allows better spatial selectiveness, but the monopolar mode can achieve higher loudness levels with lower current. Monopolar stimulation may also allow localized stimulation, if the electrode contacts are placed close to the neurons. There is also the so-called common ground stimulation in which the current passes from the active electrode to each of the others, which have been connected to constitute a single reference electrode. The current levels required to elicit threshold (T) and comfortable listening (C) levels were, in general, higher for BP stimulation than for CG stimulation and were lowest for MONO stimulation. The electrodes on the Nucleus array have a relatively large surface area- 0.44 - 0.66 mm², compared to MedEL and Clarion arrays, which have surface area of about 0.14 mm². With bipolar stimulation, if the electrodes are small and in close proximity to the spiral ganglion cells, significantly lower current levels are enough to elicit auditory perception.
Within the implant processor, spectral shape is coded by filtering the signal into several frequency bands, and then mapping the filtered signals onto appropriate electrodes. This strategy is based on “place” coding in the normal auditory system. In an implant system, the effective number of independent bands, or channels, is limited by the number of electrodes and by the relative lack of isolation between electrodes. SPEAK strategy selects 6-8 spectral maxima from the outputs of 20 band filters. The output voltages are presented on a place-coding basis, non-simultaneously at a constant stimulus rate. SAS use 8 fixed filters by simultaneous monopolar or bipolar stimulation. Simultaneous stimulation could lead to unpredictable variations in perceived loudness due to current interaction. To avoid this is necessary to separate channels temporally and spatially. CIS strategy stimulates multiple channels non-simultaneously, but at a higher rate. The effective number of channels is less than in a normal ear, and is typically less than eight, although the effective number may be somewhat greater with recently developed modiolar-hugging electrodes. There is some controversy about the number of channels required for adequate speech perception, but it seems likely that in difficult listening situations, such as when background sounds are present, at least 16 channels are required.

The electrode-neural interface is one of the most important limitations of the present technology. Scalar electrode arrays have limitations in the specificity of neural stimulation because of the reduced number of effective channels which provide information. If a save and reliable intraneural electrode system could be developed, improvement in stimulation specificity is possible.

One of the potential places for introduction of the intraneural electrode is the modiolus. We studied the modiolus with light microscopy and stereomicroscopy aiming to enlighten the possible placement of an electrode along its axis. The ideal option would be if the electrode could enter canalis longitudinalis modioli. Our examination of longitudinal sections of the modiolus revealed spiral ganglion cells and their axons (fig.1, 2). They course downwards spirally and form the cochlear nerve. In the central part of the middle and basal coil some nerve fibers pass through a straight longitudinal canal
without bony structures between them. During stereomicroscopic dissection of the human cochleae, the cochlear nerve was observed at the base of the cochlea (fig.3) and along the modiolus from the basal coil to the apex.

Measurements of the length of modiolar axis were made (fig.4). The central processes of the neurons of the spiral ganglion pass through spiral canals immersing from the cochlear coil and fusing into a single longitudinal canal to form the cochlear nerve.

The modioli obtained from human cadavers were also observed by light microscopy. Light microscopic investigation of the serial sections perpendicular to the axis of the modiolus also showed the longitudinal canal and initial portion of the cochlear nerve clearly. The single form of the longitudinal canal and cochlear nerve appeared at the levels of the basal and middle coils. The spongy bony structure of the modiolus was located just around the nerve fibers. Transverse sections show many small oblique canals around. The length of modiolar axis is about 5-6 mm. The modiolar nerve fibers join to form the Cochlear nerve 2-3 mm after leaving the tractus spiralis foraminosus. This means that electrode with length of about 8 -10 mm can be implanted without fatal damage the nerve trunk. Nowadays there are electrodes with such dimensions. For example the “Nucleus Hybrid” 10 mm with 6 half-banded electrodes and MED-EL compressed electrode array - 12 pairs of electrode contacts equally spaced over a length of 12.1 mm.

The future CI technology will provide implants with large number of contacts e.g. 72, formed on a flexible polymeric substrate (liquid crystal polymer). Traces and spaces can be as small as 12.5 x 12.5 micrometers. This large number of small electrodes placed in close proximity to the modiolar fibers would require lower current levels to excite the nerve. In addition, deposition of iridium oxide can increase the charge-carrying capacity of noble metal electrodes. Traces can be spaced appropriately for high density construction, and electrode contacts can be placed precisely. They will produce independent fields, which, can be driven simultaneously. This will make possible the use
of a larger number of channels for simultaneous stimulation of several cell groups inside the nerve. In this way the electric activity of the normal cochlear nerve can be imitated and more precise phase information could be achieved. High surface area cochlear implant electrodes with much smaller geometric surface areas than current designs might be used in the future to increase the number of stimulating electrodes along the carrier. Potential problems with an increase in charge density for a common stimulus resulting from decreasing the geometric surface area would be reduced by the enlarged real surface area of such electrodes. Electrochemically modified platinum electrodes, with a real surface area approximately 75 times greater than the current standard Pt electrodes of the same geometric size, had shown in vitro a low polarization and electrode impedance, as well as a low residual direct current.

These facts suggest that implantation in the modiolus could be possible. Modiolar implantation could be an alternative to standard CI initially in cases with severe cochlear ossification and some form of cochlear malformations.

References


1. **Fig.1** A peripheral section through the modiolus shows the presence of numerous spiral ganglion cell bodies (HE x 30)
**Fig. 2** On this section is demonstrated the absence of significant blood vessels running along the modiolar axis. (HE x 12)
Fig. 3 A longitudinal section through the cochlea and internal auditory meatus is performed. The basal, middle and apical turns of the cochlea are shown. The separate nerve fibers join to form auditory nerve trunk about 2 mm after emerging from tractus spiralis foraminosus.
**Fig. 4** This photo shows a longitudinal section through the cochlea. GSPN – greater petrosal nerve; GG – ganglion geniculi; AN – auditory nerve; FN – facial nerve. The distance between marks is 1mm.