Electric Field Density Distribution for Cochlear Implant Electrodes.

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Abstract: Microelectrode array of the Cochlear Implants (CI's) is an important component of the device which performs a key function in restoring the hearing process to the deaf patients. CI's are available with different variants in the market with different design of the Electrode array. Due to the importance of Electrode array in CI it's also been researched looking for ways to microfabricate in different design, material, shape and size depending upon the requirement and the application of the device. The traditional fabrication method of the device restricts the electrode usability and its performance. In this paper we investigate and explore different design possibilities for CI electrode array. With the help of COMSOL Multiphysics ® 4.2a we also study the effect of electric field distribution caused due to stimulation of electrode array inside the fluidic environment of cochlea.

Keywords: Cochlear Implants (CI's), Electric field distribution, COMSOL Multiphysics ® 4.2a

1. Introduction.

Cochlear Implants (CIs) are implantable devices that bypass the non-functional inner ear and directly stimulate the auditory nerve with electric currents, thus enabling deaf people to experience sound again. It consists of a receiver-stimulator package, which receives power and decodes the instructions for controlling the electrical stimulation, and an electrode array, which has electrodes placed inside the scala tympani (ST) of the cochlea i.e. near the auditory nerve in order to stimulate the residual auditory nerve fibres as seen in Figure 1. [1]. The electrode array is one of the important components of the CI. Over the past three decades, the design of these electrodes has developed from simple single- channel devices to multiple site arrays consisting of 12 to 22 stimulating contacts. In all the currently manufactured devices, these stimulating sites are built of platinum-iridium alloy (Pt:Ir 90:10) which are molded into carrier of silicone elastomer. This carrier not only holds the contacts in the intended position but also helps easy insertion of the electrode array. The primary goal taken into consideration while designing the electrodes includes deeper insertion into the scala tympani to better match the tontopic place of stimulation to the frequency band assigned to each electrode channel, greater operating efficiency and reducing the intracochlear damage caused during surgical insertion [2].

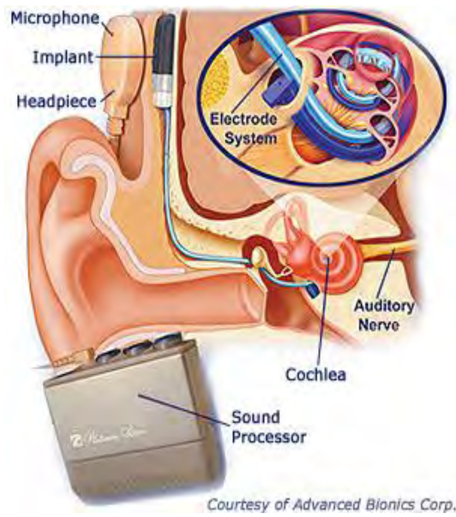

Figure 1. Cochlear Implant with the electrode array.

In this paper we look into the problems associated with the CI electrode array with the description towards the design of the electrode array. Initial simulation study with the help of COMSOL Multiphysics ® 4.2a is presented to evaluate the difference between different design configurations.

2. Problems and methods.

2.1 Problem description.

Electrodes used in CI are in the form of wire form which are still used widely due to their mechanical properties, straight construction and its easy availability. However, such types of microelectrodes lack behind due to their limited application which includes poor reproducible physical and electrical properties, short life spam and reliability due to its construction methods, and incompatibility with the inclusion of on-chip circuitry. On-chip electronics give us the space for large arrays of stimulation sites without increasing the mechanical complexity of the overall assembly. This results with the possibility of combining the electrode-stimulator structures onto a single substrate including the interface circuitry. These improved structures must serve the long term stimulation of the varied tissues along with the system stability when used into a particular application. Taking into considerations all the applications, there is a need of a multielectrode stimulating array with controllable physical and electrical properties. It should also have easily controllable stimulating site areas and spacing's, material stability at the stimulating site, and on-chip circuitry which reduces the wiring to the external leads and the last to provide a controlled stimulus signal to the electrodes [3]. The dimensions of the electrodes must be as small as possible certain enough to withstand the stimulus signal and also not too large to avoid any damage to the surrounding tissues. The material's used in fabricating the electrodes must be biocompatible and suitable for stimulation. At the end the probe must be adequately strong to withstand for the specified biological application and all the materials used must be compatible with reproducibility, high yield and a reliable fabrication process. Thus in order to meet the mentioned necessities silicon micromachining and integrated circuit (IC) technologies can be incorporated. The first thin-film microelectrodes fabricated using the micromachining techniques were constructed of gold electrodes supported on thin silicon substrate as a carrier.

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The present electrodes are a bundle of platinum/iridium (Pt/Ir) wires welded to platinum strips acting as stimulation sites. The current design is limited in electrode count, due to their large size in accordance to ST with restrictions for deeper insertion in ST thus depriving access to low frequency auditory neurons. The benefit of single site stimulation over multiple sites is to gain the freedom of perceiving multiple frequencies over a broader range to match the so-called tonotopy in the cochlea. Also multiple sites gives the audiologist an extra room for better choice of stimulation pattern, allowing fine-tuning of the device for individual patients. Silicon semiconductor technology provides the fabrication of advanced high-density CI electrode arrays with greater number of stimulation sites, integration of electronics, reduced size, multiplexing and specific site selection as per frequency. This results in less area requirements for placing the device with low power for the electronics and other components.

2.2 Initial electrode array design.

This electrode array consists of a silicon substrate material with 16 Titanium nitride (TiN) stimulation sites as seen from Figure 2 below.

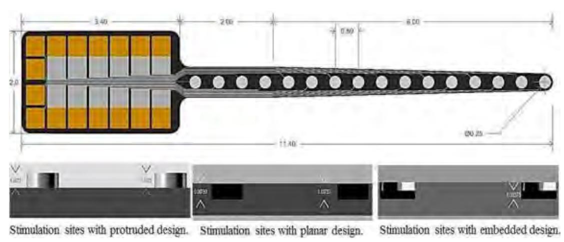

Figure 2. Initial probe design of the electrode array with three different design configuration for the stimulation site.

The array is of 11 mm length and 50 µm thickness without stimulation sites. These sites are 250 µm in diameter with 500 µm centre distance. At the end of the fabrication process the whole device is coated with a biocompatible material to avoid deterioration of the device. The complete fabrication process and other details are published elsewhere [5]. Three different design variations with respect to the stimulation sites are considered as seen in Figure 2. Stimulation sites with protruded design, planar design and embedded design were studied with the help of COMSOL Multiphysics ® 4.2a to observe the electric charge density distribution at the stimulation sites when stimulated under normal conditions as well as saline environment. This is important since greater electric field in the nearby area of stimulation sites may cause over-stimulation and may lead to residual hearing damage. These field should have sufficient value to trigger an action potential in the nerve, but should also be low in the surroundings to avoid the neural damage resulting in loss of residual hearing inside the cochlea. TiN sites are connected to aluminum bond pads of 1.5 µm thick at the base of the array with titanium nitride metal lines of 5µm width and 200 nm thick.

3. Simulation method.

The electric field density distribution study was performed with the help of COMSOL Multiphysics ® 4.2a. To mimic the environment inside the cochlea various assumptions were made. In our 2D simulation study the perilymph environment was created by assuming an space area of 600 x 250 µm filled with perilymph. A cross-section of the part of the stiff probe is considered for simulation purpose as seen from the details in Figure 3. The other material and their properties are listed in table 1.

Table 1: Material properties considered for simulation.

Material Name	**Electrical Conductivity, sigma (S/m)**	**Relative Permittivity ε_r**
Silicon	$4.3e^{-4}$	11.7
Titanium Nitride	5000	100
Perilymph	2	50
Parylene HT	$5e^{-20}$	2.2

The 'Electric Currents (ec)' section of the AC/DC module of the Comsol Multiphysics ® 4.2a is used to simulate the model. This module has some standard properties:

1. Current Conservation (all domains).
2. Electric Insulation (all boundaries, except the bottom line).
3. Initial Values (all domains).

The properties special for this model are:

1. Electric Potential (just the center electrode, its top boundary).
2. Ground (the bottom boundary of the perilymph, as an extracochlear electrode).

These properties make sure that the charge will follow its correct path. Furthermore, all components are initialized on 0 V, except for the stimulated electrode which has the potential of 544 mV [8] regarding to the ground which is the lower boundary of perilymph. In reality, there will be an electrode just outside the cochlea or none at all. In that way, there will be an electric field around the electrode and a potential high enough to trigger an action potential in the nerve ending. A stationary simulation step in combination with the parameter sweep is done to achieve the results.

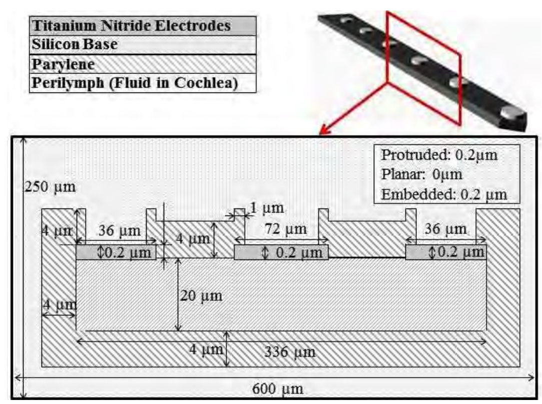

Figure 3. Part of the stiff probe with the cross section in red considered for COMSOL simulation.

The meshing of the model is done by free triangular shapes having maximum element size of 40.2 µm and a minimum of 0.005 µm. A growth rate of 1.2, curvature resolution of 0.3 and the narrow region resolution is 1. The number of elements is approximately 60000.

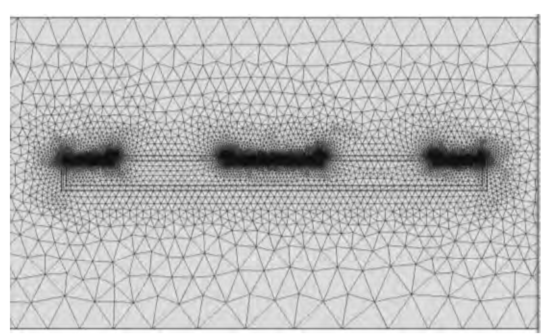

Figure 4. The mesh pattern of the model using COMSOL Multiphysics ® 4.2a simulation.

4. Results.

Electric field distribution at the surface and the potential in the form of contour lines for the protruded shape design is as shown in figure 5. As perilymph which is equivalent to cerebrospinal fluid (CSF) is nothing but a mixture of high K^+ and low Na^+ concentration, it's conductivity is less in respect to the other materials so the filed distribution is not extended far enough.

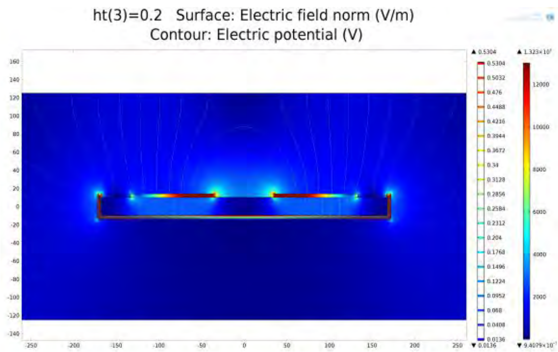

Figure 5. Electric field distribution of the model for the protruded design.

Electric Potential at 30 µm from the surface for the three different design configurations is as shown in figure 6. There is no significant change observed for the three design variants.

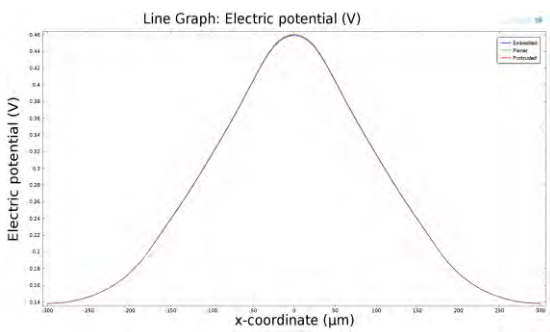

Figure 6. Electric Potential from the surface for the three different design configurations.

7. Conclusions.

As seen from the results from Figure 6 the influence of the electric filed distribution for all the three variants is not significant. Also from the microfabrication point of view the stimulation site with the protruded design is favored resulting in a proximal distance between the nerve and the stimulation site .

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9. Acknowledgements.

This work (SMAC-it project) is supported by the Dutch Technical Foundation, STW (Project 10056) with its processing work supported by The Delft Institute of Microsystems and Nano electronics (DIMES), TU Delft, The Netherlands. The authors would also like to thank Advanced Bionics ® for their support in this project.