Study of Cochlear Implants Electrodes Stimulation Based on the Physics of the Ear for Audio Signal Integrity Improvement

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***Abstract* – Cochlear implants are neuro-stimulating devices helping severely deaf people to perceive sounds. This work proposes two models for the optimization of electrodes array stimulation in cochlear implants. The first model was developed according to wave propagation theory applied to the cochlea. The second one consisted on data processing using a Short Time Fourier Transform (STFT) in order to remedy to the first model limitations. The models were tested with an acoustic signal, which frequency belongs to the human hearing frequency range. Frequency and transient representations of the reconstructed signals were compared, emphasizing the better linearity of the STFT model.**

Keywords- cochlear implants; physics of the ear; audio signal processing

I. INTRODUCTION

As electronic technology is getting smaller, more accurate and low power, embedded processing solutions are nowadays available to encapsulate in biomedical applications. Among the disability remedy devices, cochlear implants are used to remedy partial or complete hearing loss. They allow direct stimulation of the auditory fibers with an electrodes array designed to reproduce the stimulus that would be generated by a healthy cochlea (Fig. 1).

To do so, an external part of the hearing devices is located within the outer ear and contains a microphone that captures the acoustic waves and transforms them into an electrical signal. Then, this signal is up-converted from baseband (audio signal from 20Hz to 20kHz) to the transmit frequency defined by the chosen ISM standard (13,56MHz, 433MHz, 866MHz or 2.45GHz). The corresponding receiver is located within the patient's head, close to the skull. It is composed of a demodulator, a data processing unit and a set of electrodes driven by electrical signals that will contract the cochlea and stimulate the auditory nerve [1][2].

However, the reconstructed signal is far from being the same as the original sound due to the fact that the number of electrodes is limited (around 20 depending on the type of implants) and the total contact area between the cochlea and the electrodes is small compared to the total cochlea length, resulting in signal distortion. As stated in [3] patient speech recognition is mainly influenced by the audio signal (80% of speech recognition is achieved using only auditory indications, 40% using only lip reading and 90% using both). As a consequence, signal integrity has to be preserved through cochlear implant signal processing, to allow optimal patients hearing sensation.

To improve the signal reconstruction with a limited number of electrodes, two models are compared: the first one using wave propagation theory within the cochlea and the other one using an ad hoc Fourier Transform (FT) to find the electrical stimulus of each electrode. This allows having a simple and yet accurate evaluation of the auditory nerve stimulation generated by a sound cochlea and comparing it with FT based electrodes stimulation.

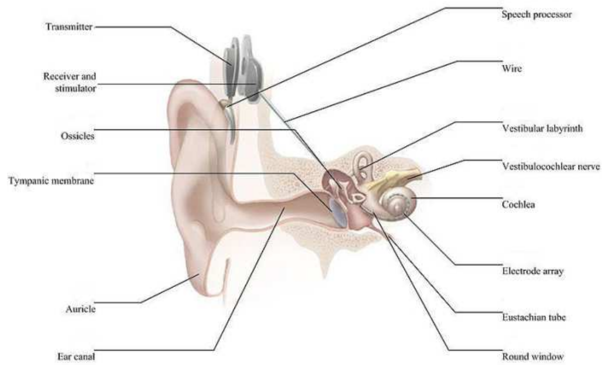

Figure 1: Ear anatomy and cochlear implant **[4]**

The second section gives an overview of the physics of the cochlea and how a cochlear implant can remedy a cochlea malfunction. In the third section, the theory of wave propagation is detailed in order to show how a sound cochlea model can be implemented. The fourth section aims to describe which frequencies are stimulated by cochlear implants. Finally, the last section gives a comparison between the signals obtained with a cochlear implant using the different electrode stimulations.

II. COCHLEA PHYSICS

When an acoustic wave traveling inside the outer ear channel strikes the eardrum, the ossicles located inside the middle ear amplify the sound intensity. This is achieved by producing a lever effect. The last ossicle (stirrup) hits the oval window of the cochlea. The fluid (perilymph) inside the scala vestibuli of the cochlea provides support for mechanical waves propagation. The amplification provided by the ossicles aims to correct the loss of energy associated with the impedance change at the air to liquid interface [5] (Figure 2).

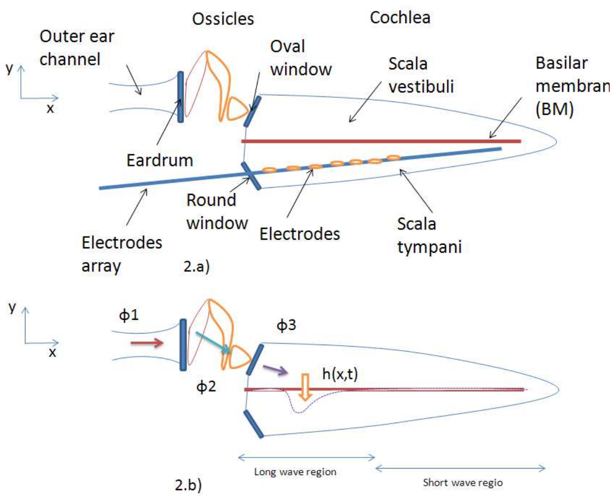

Figure 2.a) ear schematic with inclusion of the electrodes array inside the cochlea. Figure 2.b) ear schematic and the wave propagation leading to a basilar membrane height change (h(x,t))

More precisely the cochlea works by converting a wave propagating inside the perilymph into nerve a stimulation using the complex organ of Corti structure. Inside the cochlea, the Outer Hair Cells (OHC) are stimulated by a soft membrane (Basilar Membrane: BM) vibrations and release chemical messengers which excite nerve cells.

According to biophysical theories [6],[7] when a mechanical wave propagates inside the cochlea, the BM distorts to absorb the wave energy. In consequence, the height (denoted as h(x,t) in the following parts) of the BM excitement mainly depends on the sound intensity as well as the position along the x axis, which is the distance from base of the cochlea. The width, denoted as W(x), of the basilar membrane excitement around the resonance greatly depends on x as shown in the Section III. As the number of excited OHC depends on W(x) the number of auditory fibers excited is hence associated with the x variable.

If one or many functions described above are not working properly, the patient may suffer from hearing loss or deafness. In this case, a cochlear implant may substitute to the natural sound processing to provide the auditory nerve stimulation via an electrodes array sunk inside the scala tympani (Figs. 1 and 2).

The electrodes array (which is the connective link between the implant and the cochlea) is composed of a limited number of electrodes stimulated by an electric current. The electric current at proximity of the auditory fibers permits theirs direct stimulations even with nonfunctional OHC.

We chose the device CI422 with Slim Half-Band Straight Electrode manufactured by Cochlear® [8], which is composed of 22 electrodes to implement our models. Then we compare the results with a FT approach, applied to only the frequencies stimulated by the electrodes, to select the electrodes.

III. SOUND WAVE PROPAGATION AND BASILAR MEMBRANE DISPLACEMENT.

The wave propagation theory is quickly reviewed in this section for a better understanding of the biophysics of hearing and for determining the characteristics of the basilar membrane resonance in terms of amplitude and width. This theory and the computations are extracted from [9] and [10], the main results are recalled here for the reader's convenience.

A sound wave φ_1 propagating along the x axis is characterized by:

$$\varphi_1(x,t) = A_1 exp^{i(k1*x-\omega*t)} \tag{1}$$

Where A_1 is the sound amplitude, k_1 is the wave number and ω is the pulsation.

At the oval window interface, the mechanical wave is converted into a wave propagating in a viscous fluid, resulting in impedance mismatch. Therefore, the transmitted power from the mechanical wave into the liquid wave is damped by 29dB.

As demonstrated by various authors, the main function of the ossicles in the middle ear is to reduce the power loss due to the impedance mismatch therefore the wave power at the oval window interface is equal to the power of the air wave minus 3dB [11][12].

The wave propagating inside the scala vestibuli (φ3) is expressed as:

$$\begin{aligned}\varphi 3(z,t) = \frac{A_3}{2} * [&\exp\big(i(k*z^* - \omega*t)\big) \\ &+ \exp\big(i(k*z - \omega*t)\big) \\ &+ \exp\big(-i(k^* * z - \omega*t)\big) \\ &+ \exp\big(-i(k^* * z^* - \omega*t)\big)]\end{aligned} \tag{2}$$

Where A_3 is the amplitude of the wave, $k = k_r + ik_i$ is the wave number and z the complex variable: z = x +iy.

Mathematical equations developed in [9] lead to the following relation between membrane height displacement and auditory wave:

$$\frac{\partial h(x,t)}{\partial t} = -\frac{\partial \varphi_3(x,y,t)}{\partial y} \tag{3}$$

Solving this equation permits to obtain the basilar membrane height displacement h(x,t) (spatial and transient shape of the basilar membrane displacement is given in Figure 3)

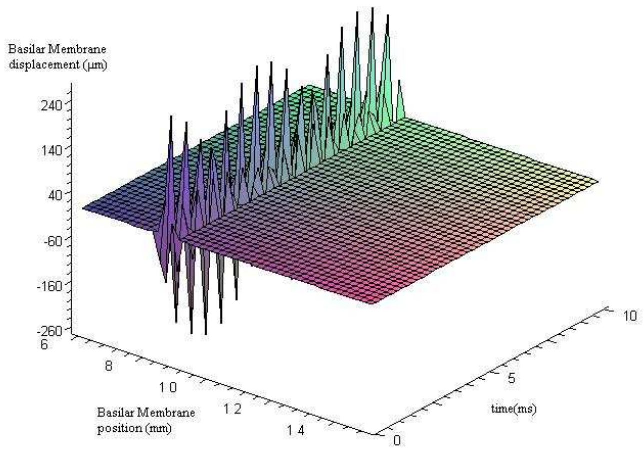

Figure 3: Basilar membrane displacement (μm) with respect of the distance from the cochlea base (x in μm) and time (ms) for a 600Hz input acoustic wave

One can observe that the BM displacement is a function of the distance from the cochlea base and the amplitude of the auditory wave. The auditory nerve stimulation is proportional to the BM displacement and this phenomenon has to be recreated artificially with electrodes stimulation.

IV. ELECTRODES ARRAY STIMULATION BASED ON THE COCHLEA PHYSICS

The width of the BM displacement greatly depends the position x as shown in Fig. 4.a, when the BM is excited by a 600Hz sound wave and by a 300Hz sound wave which make it to resonate at different places and with different widths. The detection of a BM displacement permits to obtain the width of this excitement:

$$W(t) = \int_{X_W} h(x,t)dx$$
$$with\ X_W = \{x \mid abs(h(x,t) \geq threshold\} \quad (4)$$

The basilar membrane movement compresses the organ of Corti resulting in a voltage variation in the corresponding auditory fiber. If this voltage variation is greater than a threshold voltage around 30mV for the human auditory nerve [13], the nerve fiber is excited. We implemented the mechanical model of the organ of Corti found in [14] (we simplified it by neglecting the Tectorial Membrane influence over the Reticular Lamina) and after converting the voltage threshold into the corresponding membrane height, we obtained the basilar height threshold value around 20μm.
Extraction of the width W for acoustic waves of different frequencies varying from 60Hz to 20KHz (hence resulting in different place of excitation of the BM) is shown in Fig. 4.b.

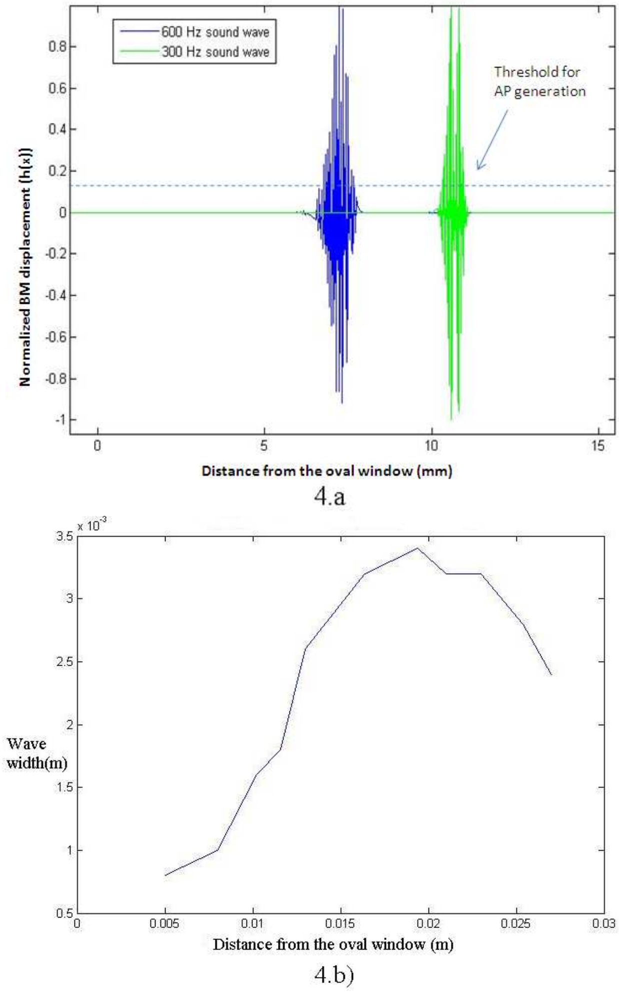

Figure 4.a: BM excitation. Figure 4.b: Width of the BM excitement.

We use CI422 device characteristics with an insertion depth of 20-25mm, a mean diameter of the electrodes around 0.35mm and a spacing between the electrodes around 0.45mm. As explained in [8], the number of nerve fibers stimulated by an electrode is a function of the power magnitude as well as the proximity of the electrodes with the BM. In fact, the electrodes array is sunk inside the scala vestibuli and due to spatial inhomogeneity of the cochlea, the distance between the electrodes and the BM may be fluctuating. For simplification purposes we supposed that the electrodes array is very close to the BM resulting in a rectangular window-like stimulation of the nerve fibers by the electrodes ($H_{implants.}$) displayed in Fig. 5.a. The selection and the power sent to the electrodes depends on the function h(x,t) as depicted in Figs. 5.b. and 5.c

The relation between the distance from the oval windows where the BM displacement is maximaland the frequency of the acoustic wave is given by the Greenwood function [15] and will be used thereafter in this paper to switch from the distance x where W(x) is maximal to the frequency of the input wave.

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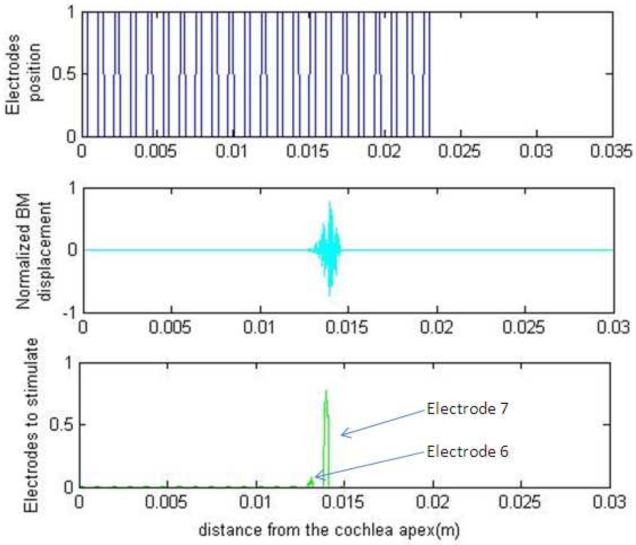

Figure 5.a: The auditory nerve fibers position stimulated by the electrodes array.
5.b: BM displacement for a 1250Hz sine wave.
5.c: Resulting electrodes stimulated by the 1250 Hz sine wave based on the BM displacement theory

As can be seen on the figure, not all the frequencies can be stimulated by the electrodes and the main limitation of this model is that the input frequency must match with the frequencies coverage by the implant, which is rarely the case for high frequencies. In consequence, an alternative model has been developed

V. COMPARAISON WITH THE FT BASED AUDITORY NERVES STIMULATION

The function $H_{implants}$ presented in Section IV selects the frequencies stimulated by the electrodes array. We chose to only compute the Fourier Transform on the frequencies which can be stimulated by the cochlear implants. Hence we obtain the following equations for the FT computation:

$$X_k = \sum_{n=0}^{N-1} x_n * e^{-\frac{i2\pi f_{central}(k)n}{N}}$$

And

$$x_n = \frac{1}{N} \sum_{f_{central}(k=0)}^{f_{central}(N-1)} X_k * e^{\frac{i2\pi f_{central}(k)n}{N}} \qquad (5)$$

Where X_k are normalized frequencies (the normalizing frequency is the sampling frequency chosen at 44KHz, similar to audio formats as .wav, .mp3, etc.) ,$f_{central}$ is the central frequency of each rectangular sub window of $H_{implants}(f)$ as one electrode can only have one voltage value and N is the number of electrodes. As the sampled audio signal contains more than N points, the sampled audio signal was cut into many windows of N samples where the previous formula was applied (Short Time Fourier Transform (STFT) [16]).

The corresponding results for a 1250 Hz sine wave stimulus is shown in Fig. 6.a.

As the frequency of 1250Hz does not match with a frequency covered by one electrode, the signal power is distributed over the frequencies covered by the electrodes, causing distortion in the reconstructed signal.

This test signal stimulates only two electrodes (in the actual physical model as presented figure 5.c) or all the electrodes (in the STFT model). Transient simulations were carried on and reconstructed signal from the STFT model was significantly less distorted than the other one (Fig. 6.c). However, when the frequency of the control signal was inside the frequency range corresponding to the electrodes position shown in Fig. 5.a, both models gave approaching results (not shown in this paper). In that case the most efficient model in terms of processing resources should be selected.

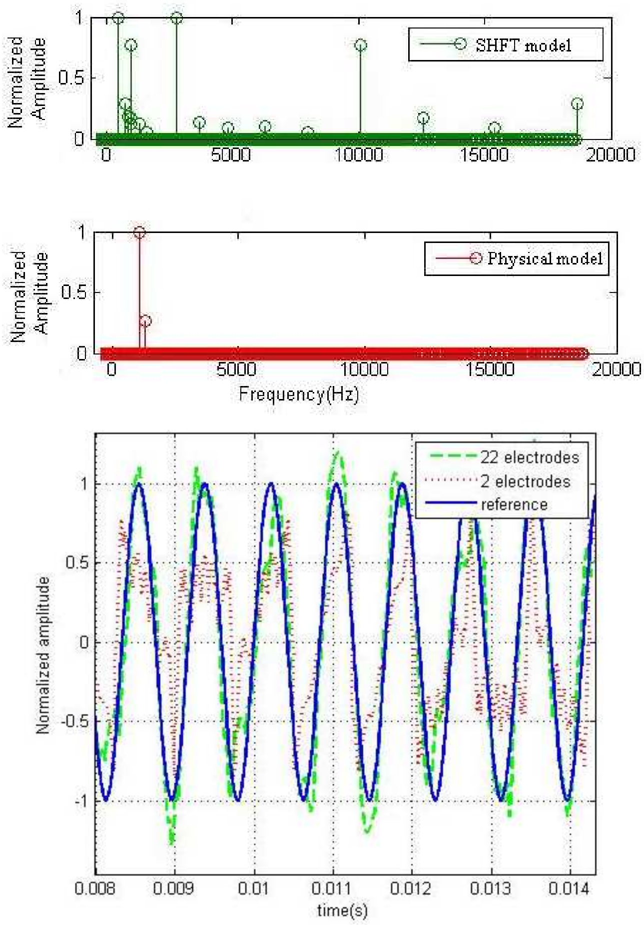

Figure 6.a: Normalized voltage sent to the electrodes for a 1250Hz input auditory wave (using the SHFT model).
6.b:Normalized voltage sent to the electrodes for a 1250Hz input auditory wave (using the BM displacement model).
6.c: Comparison between transient simulations of the reference signal, the signal reconstructed using all the electrodes and the signal reconstructed using only 2 electrodes

Testing this hypothesis may provide new insights of the hearing process. In fact if the combined models would give better results in person using cochlear implants than the algorithms implemented at present, this could mean that the brain could adapt to these new nerve stimulation. In that case our combined models, if utilized in very young patients (maximal brain plasticity and with extended learning process), could give better results than the algorithms used commercially. For instance the modification of the current algorithms to include our model is the first step, before its implementation on the cochlear implants microprocessor for real time data processing

VI. CONCLUSION

We proposed an electrodes stimulation optimization which resides on the complementary nature of physical and mathematical models, therefore, deserving further work and tests. The two models proposed in this paper (physical cochlea model and FT based model) have been compared theoretically and the corresponding results have been presented for a 1250Hz sine wave input signal. At present general knowledge in hearing processes is improving due to a large community of biomedical research, which should bring further development in the modeling of the cochlea and the mechanism of hearing. In this context the implementation of our model in upcoming cochlear implants may add hearing accuracy.

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