HUMAN COCHLEAR MODEL: A SIMULINK IMPLEMENTATION

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Abstract

Models of human auditory pathway are commonly used as a part of audio coding systems or algorithms for objective evaluation of sound quality. In order to simulate properly the nonlinear, signal-dependent behavior of the cochlea, physiologically accurate models are being developed. In the paper, a SIMULINK implementation of a hardware active nonlinear model of human cochlea is presented. The model was designed to reproduce satisfactorily the masking properties, frequency selectivity and sensitivity of human inner ear.

1 Introduction

In the past, various attempts to model the effects of human cochlea were made. The most common ways of modelling its frequency selectivity employ either a certain form of Fourier transform (FFT, DCT, MDCT etc.) or a linear filterbank [4]. Both approaches fail to mimic properly some fundamental properties of cochlear frequency transformation: non-linearity, level-dependent response and assymetry. For this reason, more accurate models are being sought, including non-linear filterbanks [2, 3] and models based on electro-mechanical analogy [5, 7]. In this paper, a SIMULINK implementation of a hardware nonlinear cochlear model according to [7] is addressed.

2 The Cochlea

2.1 Physiology

Cochlea is a fluid-filled tube divided longitudinally in three canals; *scala media* forms an elastic partition between the two remaining scalae, *scala vestibuli* and *scala tympani* (see Fig. 1). Scala tympani and scala vestibuli are connected at the tip of the cochlea (*helicotrema*), to enable the flow of the fluids. Scala media, on the other hand, is separated completely from the two other canals. At the base of the cochlea, scala vestibuli opens to the middle ear cavity with a membraneous *oval window* representing the input to the cochlea. The oval window is connected to the ear drum via the middle ear ossicles. The auditory receptor itself, the *organ of Corti*, is placed inside scala media, along the *basilar membrane* (BM). It contains two sets of sensory cells: the *outer* and *inner hair cells* (see Fig. 2).

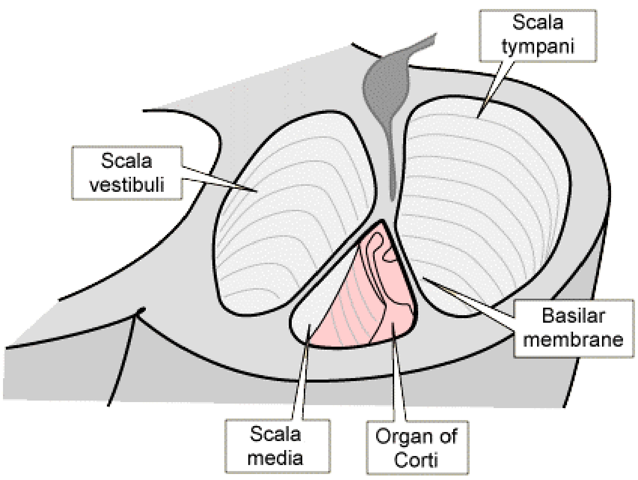

Figure 1: Cochlea, cross-section.

The vibrations of the middle ear bones are transmitted to the cochlear liquids via the oval window. Pressure changes in the liquid result in basilar membrane motion which is registered by the hair cells. The outer hair cells (OHC) act as a saturating amplifier, adaptively amplifying vibrations of the basilar membrane. The inner hair cells (IHC) act as the very receptor, translating the vibrations to the series of neural impulses.

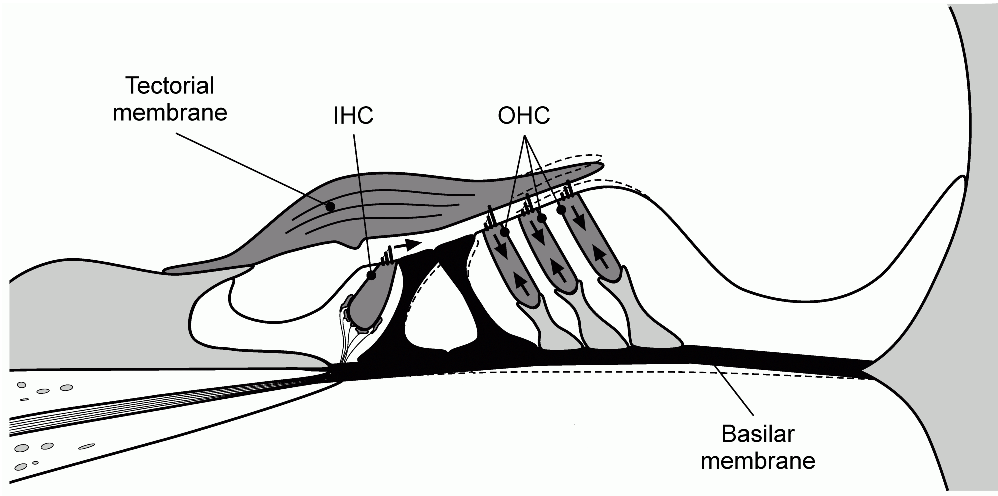

Figure 2: The organ of Corti, cross-section.

The cochlea provides a certain kind of frequency-to-place transformation: each sound frequency causes excitation of a different region of the BM. Higher frequencies are resolved at the base of the cochlea, low frequencies excite regions near the apex. Therefore, a unique *characteristic frequency* (CF) can be assigned to a given position along the BM. This principle evokes the structure of a filterbank. However, positioning of the filters along the frequency axis is nonlinear and is often expressed in terms of so called critical-band, or Bark, scale. The tonotopic assignment follows Eq. 1:

$$CF = A\left(10^{x} - k\right) \tag{1}$$

where CF is expressed in kHz, x is the relative distance from apex, $x \in \langle 0,1 \rangle$, k is constant [6].

2.2 Function

Humans are capable of perceiving sounds with intensities varying in the range of approximately 120dB. To reduce the amount of information advancing to the brainstem, our hearing system exhibits certain compressive behavior. The cochlea plays an important role in this “data rate” reduction: the amplitudes of BM vibration fall into the range of only 60dB [6]. Moreover, weak stimuli presented simultaneously with a loud sound of similar frequency are not perceived (this phenomenon is called masking) [8].

The cochlea is not a passive system. The active amplification of BM vibrations is provided by the outer hair cells which significantly broaden the range of audible intensity levels, lowering the hearing threshold in quiet by as much as 50dB. The effect of OHC is particularly pronounced for low level stimuli and becomes less prominent with increasing sound intensity. As a consequence of the active non-linear amplification, phenomena like cubic difference tones or otoacoustic emissions occur. Another consequence is the dependence of sharpness of the cochlear filter on the stimulus level. For low level sounds, the BM vibrations are sharply tuned, displaying a high peak at the characteristic frequency. As the stimulus level increases, the vibrations become more broadly tuned and the peak amplitude shifts towards higher CF (towards the cochlear base) [6].

3 The Model

Nonlinear, level-dependent response, active amplification and assymetry of masking patterns are difficult to represent by Fourier transform or a filterbank. The presented model [5, 7] is based on electro-mechanical analogy, with a simplified presumption of basilar membrane acting as a one-dimensional system of resonators. In the model, the entire length of cochlea is divided into electrically coupled sections with the resolution of eight sections per critical band (Bark). The hydromechanics and motion of the BM are simulated by a cascade of resonators, tuned to the particular characteristic frequencies. Such an arrangement corresponds to the frequency-to-place transformation (see Fig. 3).

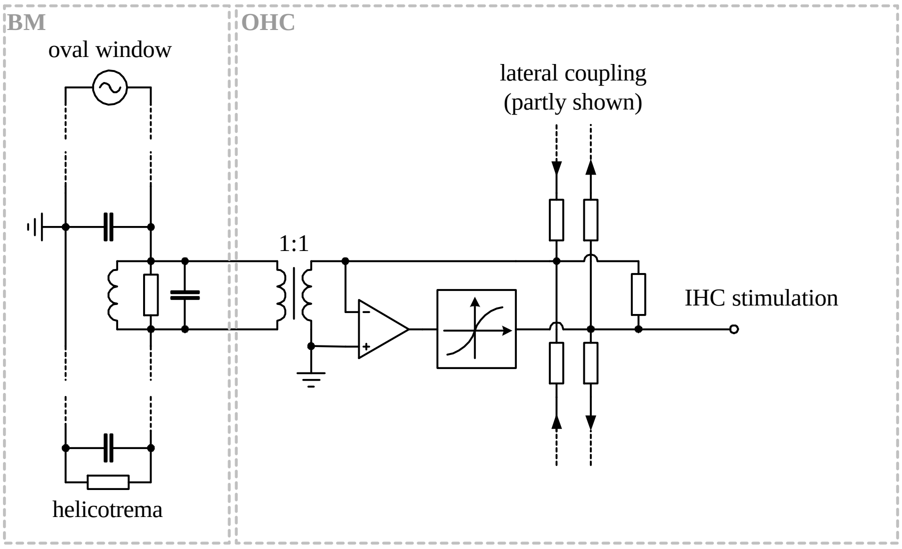

Figure 3: One section of the cochlear model. Basilar membrane and OHC circuits outlined.

The active amplification of the OHC is taken into consideration by adding the lateral electrical networks connected to individual resonators. These circuits comprise an ideal amplifier plus a nonlinear element, providing the desired amplification of low level signals as well as the observed nonlinearity. The output of the OHC circuit serves as the driving signal for an IHC model, and is also fed back to the basilar membrane circuit to influence its vibrations. The nonlinearity follows Eq. 2:

$$y = \mathrm{sgn}(x)\left|K x\right|^{0,4} \tag{2}$$

where x is the input signal (BM velocity), y is the output signal (IHC stimulation), and K is the gain of the amplifier.

In the continuous system of cochlea, even a pure tone excites a *region* of BM. Moreover, the neighbouring hair cells are not totally electrically isolated and influence each other to a certain extent. This fact is reflected in the model by means of lateral coupling of the neighbouring OHC sections, which decreases with the relative distance of the coupled sections.

4 Implementation

4.1 Top level

The above described electrical circuit was implemented in the SIMULINK environment, utilizing predominantly the SimPowerSystems blockset. Fig. 4 depicts the top level of the model, including the input and the first four critical bands, each containing eight model sections.

Each critical band block can be connected to adjacent blocks, the **`IN`** & **`OUT`** ports interconnect the resonators (basilar membrane circuits), the **`In1`** to **`Out4`** ports interconnect the OHC circuits (see also Fig. 3, the lateral coupling). The **`K`** input can be used to set the gain of the nonlinear amplifier. Finally, the **`IHC`** port is the multiplexed output of the eight model sections that is to be lead to subsequent stages of the auditory pathway model (eg. inner hair cell model).

Standard generator blocks were used for testing the model, eg. the chirp signal generator. To gather the outputs or to check a single model block, a multimeter block was used to measure the desired quantity. If needed, a zero-order hold block plus a spectrum-scope was used.

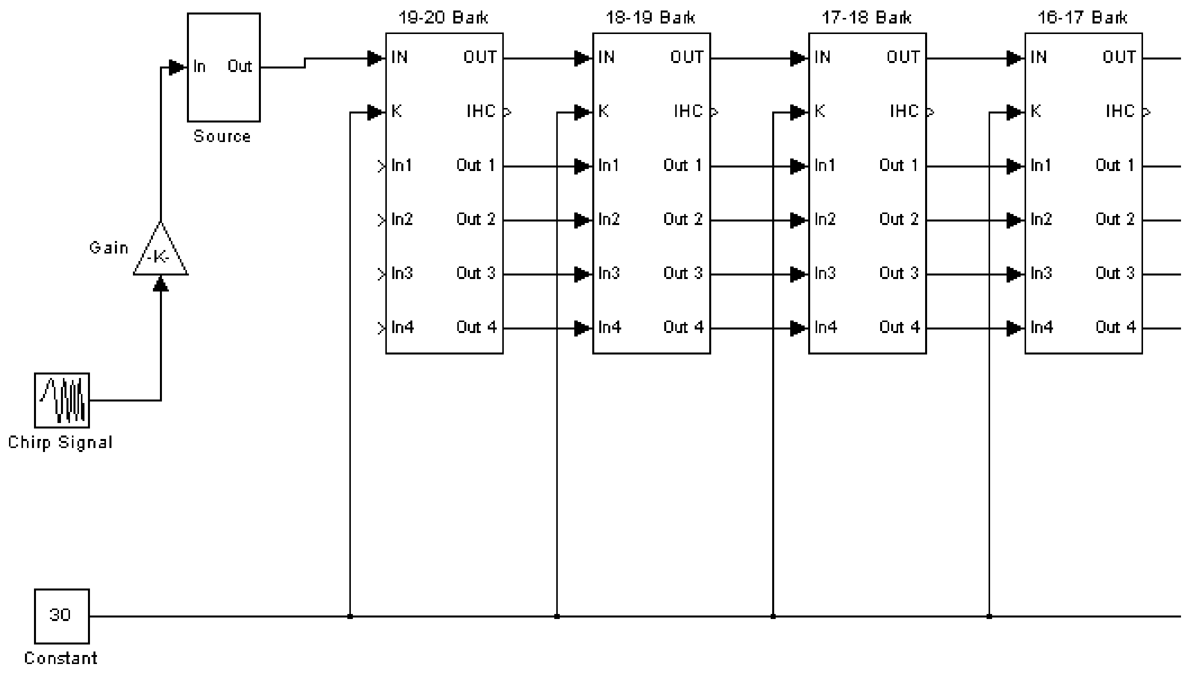

Figure 4: Top level of the cochlear model, the input plus four critical band blocks.

4.2 Input

The input to the system was implemented using a voltage source (see Fig. 5).

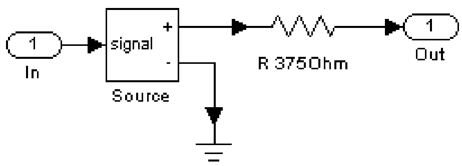

Figure 5: Input to the cochlear model.

4.3 Critical band block

Each critical band block contained eight model sections, that is eight resonators plus the corresponding OHC circuits. Fig. 7 shows the first five sections with the respective resonant frequencies noted, the input ports to the critical band block, and the multiplexed output. Fig. 6 depicts a single resonator block with its input and outputs.

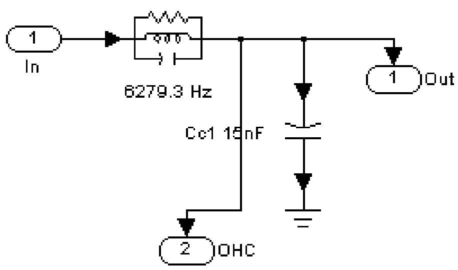

Figure 6: A single resonator included in the BM circuit.

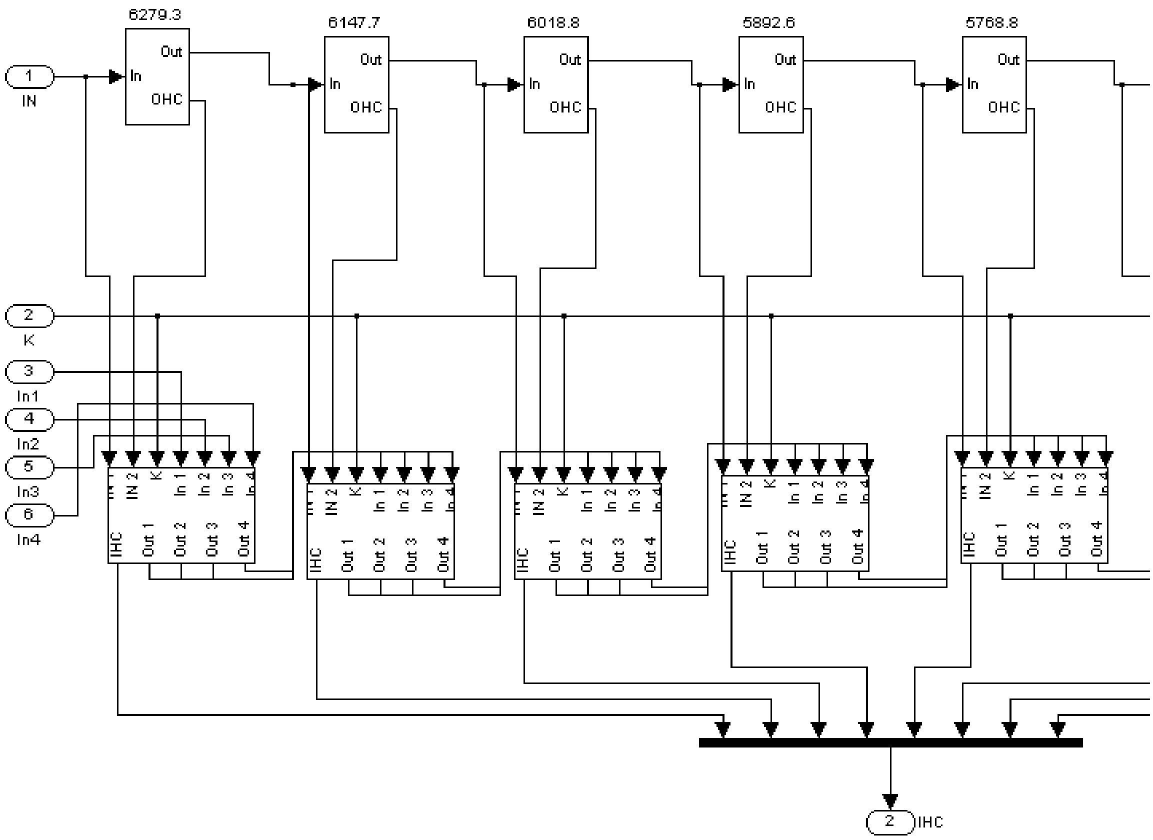

Figure 7: Critical band block (partly shown).

The top level of the OHC circuit is shown in Fig. 8. The BM resonator voltage is transmitted using an ideal 1:1 transformer, to allow bidirectional coupling. The inputs and outputs are clearly visible, note the **`K`** port and the **`In1`** to **`Out4`** ports. The detailed view of the OHC block is given in Fig. 9.

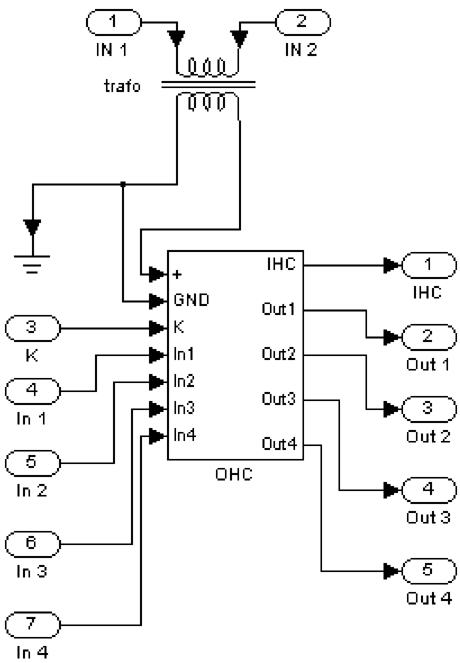

Figure 8: A single OHC circuit.

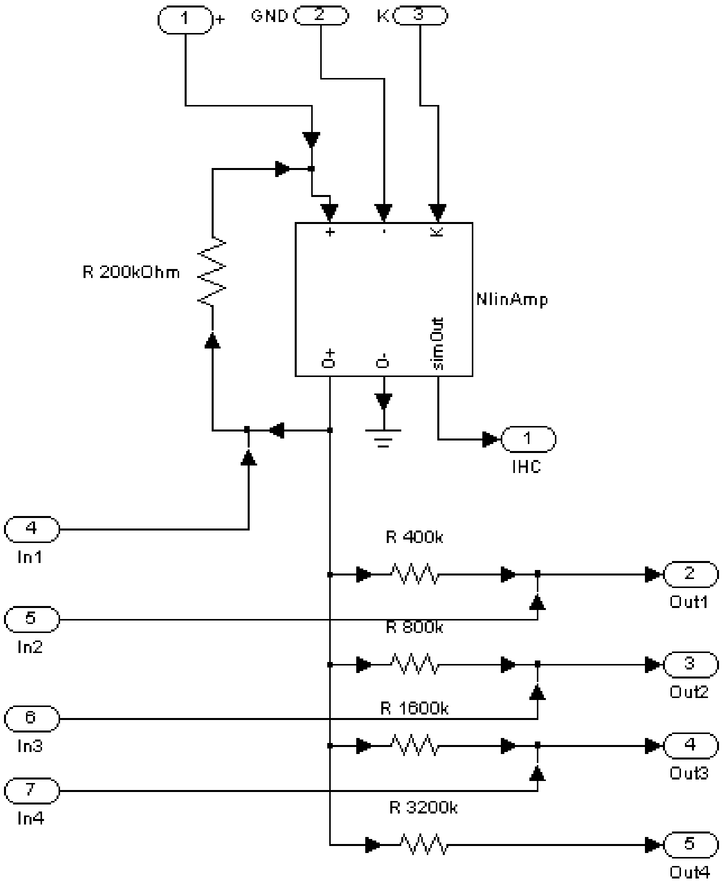

Figure 9: A single OHC circuit, detailed view.

4.4 Nonlinear amplification

The OHC model is based upon a nonlinear amplification (see also Fig. 3). To avoid any possible problems with unresolvable dependencies, the ideal amplifier was implemented using an ideal voltmeter, a SIMULINK Fcn block (see Eq. 2), and an ideal voltage source. The function of this circuit was tested separately as it mixes together the SIMULINK and SimPowerSystems blocks to provide the desired active behavior; the SimPowerSystems blockset lacks any active electrical elements.

The output of the cochlear model was taken directly from the Fcn block, thus avoiding an additional transformation of SimPowerSystems data format to SIMULINK data format. Any other solution of collecting the model outputs lead to a deadlock. The structure of the nonlinear amplifier is shown in Fig. 10.

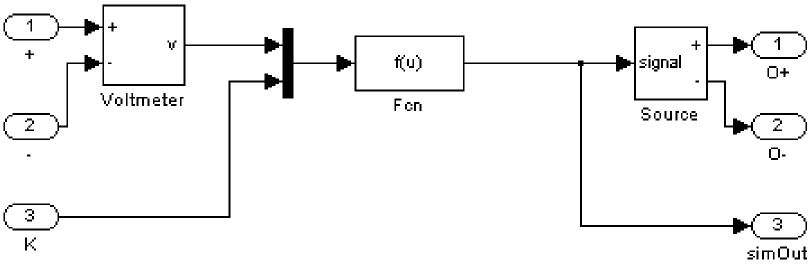

Figure 10: The nonlinear amplifier.

4.5 Helicotrema

The termination of the resonator cascade at the position of helicotrema had a form of a passive impedance, matching the output impedance of the cascade to avoid reflections (see Fig. 11).

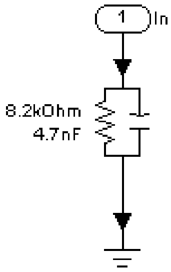

Figure 11: The terminating impedance.

5 Discussion

During the implementation of the hardware active cochlear model using SIMULINK and SimPowerSystems blockset, several difficult problems emerged. One of them was the occurence of the troublesome unresolvable loops which always lead to a substantial drop in computation speed. It was not possible, however, to remove these unresolvable dependencies by introducing an initial condition block. Finally, a solution described in this paper was found, that suffers no such loops.

To speed up the computation, the model was run in the accelerator mode which takes advantage of a C language representation of the simulation. The achieved speed was higher than that of normal mode, yet the computation times were quite long for the application to be viable.

The initial evaluation of the model revealed an unexpected resonant behavior in case of special input configuration. This behavior was caused probably by cumulative numerical errors and was never explained satisfactorily.

Despite the advantage of the modular structure, the model was quite difficult to maintain using standard SIMULINK tools. Fortunately, it was possible to alter the source code of the SIMULINK structures directly. This way, some block parameters could be modified very easily and almost simultaneously.

The SIMULINK environment offers tools for communication with MATLAB workspace. The connection of the individual auditory model stages, however, is somewhat problematic due to impossibility of running the entire model at once. The MATLAB and the SIMULINK stages have to be run successively.

For these reasons, a different approach was sought, that would avoid the above mentioned problems. Later during the development, a C language implementation of the described model was found, which utilizes the Wave Digital Filter technology [1]. Although this solution does not allow as easy modification and its principle is less obvious than that of the SIMULINK implementation, it was decided to make use of the WDF approach. The details of this alternative implementation along with the evaluation of the complete model will be published elsewhere.

Acknowledgment

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