AMSE
IIETA International Information and Engineering Technology Association
Advancing the World of Information and Engineering

Modelling, Measurement and Control C
Vol. 79, No. 4, December, 2018, pp. 179-184
Journal homepage: http://iieta.org/Journals/MMC/MMC_C

Cochlear implant: On the number of channels

P.A. Cucis[1,2,3*], C. Berger-Vachon[1,2,4], S. Gallego[1,5], E. Truy[1,2,3]

[1] University Claude-Bernard Lyon1, 43 bd du 11 Novembre, Villeurbanne-Cedex 69622, France
[2] Lyon neuroscience Research Center, (CRNL), INSERM U1028, 95 bd Pinel, Bron-Cedex 69675, France
[3] ENT Department, Edouard-Herriot Hospital, 5 place d'Arsonval 69437 Lyon cedex 03, France
[4] IFSTTAR, 25 Avenue François Mitterrand, 69500 Bron, France
[5] Sensory and Cognitive Neuroscience Laboratory (LNSC), CNRS UMR 7260, Université d'Aix-Marseille, France

Corresponding Author Email: cucis.pa@gmail.com

https://doi.org/10.18280/mmc_c.790405

Received: 28 September 2018
Accepted: 31 October 2018

Keywords:
cochlear implant, channel number, sound coding strategies, channel interaction, simulation, behavioral and physiological measurements

ABSTRACT

Cochlear implant efficiency is linked to the number of spectral channels. There is a good spectral resolution when many channels are used but, there are also interactions between those channels and it limits speech understanding. Several studies have been conducted on the subject but there is no consensus on the choice of a sound coding strategy. The efficiency of a sound coding strategy depends on the external conditions and on the patient's physiology. Channel interaction is measurable and it is worthy to study relying on the state of the art. This article considers some papers on the topic and suggests a new approach based on physiology and on simulations. This is the starting point of a new research project.

1. INTRODUCTION

Sound coding strategies are always under investigation in order to provide the best service to the patients particularly regarding the perception of speech in noise. The number of open channels seems to be linked to the amount of channel interaction between the electrodes. In this paper, we will focus on this issue using the literature.

Considering modern techniques, it is worthy to explore this topic. First, we need to do a state of the art.

Our team (CRNL, HEH, LNIA) suggested that increasing the number of open channels increases channel interaction at the expense of speech understanding. A good approach is selecting relatively few channels [1]. Nevertheless, when there is background noise, a high number of channels is needed to improve speech understanding [2]. Our team already addressed the impact of coding strategies in noisy environments and the interconnection with channel interaction. It is worthy to consider channel interaction during cochlear implant fitting process [3, 4].

We recently started an experiment which aims at putting channel interaction through speech understanding. The idea comes from one of our previous studies [5]. The results suggested that, in noise, vocal audiometry scores are better when the patients use all the available channels [6-7]. Results were very heterogeneous and we suggested that channel interaction could be the cause. Our goal is to gear the number of open channels to the external conditions.

2. MULTIELECTRODE COCHLEAR IMPLANT AND CHANNEL INTERACTION

2.1 Cochlear Implant

A cochlear implant is a medical device designed for people with profound deafness. It is an interface between the sounds of the environment and the neural fibers inside the cochlea. A cochlear implant is an acoustic-electric transducer that artificially excites the auditory nerve and replaces the non-functional inner hair cells.

Charles Eyriès and André Djouno described in 1957 the electrical stimulation of the auditory nerve [8]. With only one electrode, it enabled the patient to detect speech rhythms. Then, in the 70's, Prof C.H. Chouard and his team developed the first French multi-electrode cochlear implant (Chorimac) that enabled the patients to have access to a wider frequency range which is essential to understand speech [9].

A cochlear implant has two parts. One part is implanted and composed of the implant body and the electrode array (inserted inside the cochlea). The external part is composed of the sound processor and the antenna (Fig. 1).

Cochlear implants work as follow: first, the microphones capture sounds and the chip analyzes and transforms them into electric impulses. Then, the antenna transmits the information through the skin via electromagnetic waves to the internal part. The internal part receives the information: first, the body of the implant sends the signal to the electrode array and then the electrodes stimulate the neural fibers of the cochlea. The nervous signal induced by the stimulation travels through the auditory nerve into the brain which interprets them as sounds [10].

Nowadays in France, two main sound coding schemes are used by four manufacturers. On one side, "CIS like" (Continuous Interleaved Sampling) sound coding strategies are implemented in Med-El® and Advanced Bionics® sound processors. On the other side, "NofM" (for N out of M) are implemented in Cochlear® and Oticon Medical® sound processors.

All coding strategies are now based on the principle of the vocoder [11]. The signal is split into spectral bands by a digital filter bank. Each band is associated with a specific electrode corresponding to the tonotopic organization of the cochlea. In that respect, a low-frequency sound activates one or more electrodes at the top of the array and a high-frequency sound activates electrodes at the base of the array. The main difference between “CIS like” and “NofM” strategies lies in the number of electrodes activated in the same calculation run. “NofM” strategies activate the most energetic channels among the total number of channel while “CIS like” strategies activate all the electrodes available in the same run [12].

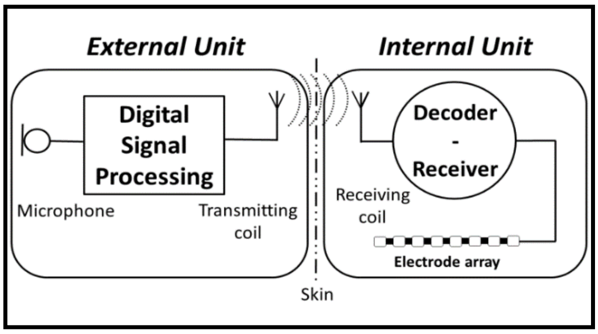

Figure 1. Schematic representation of a cochlear implant

2.2 Channel interaction

The arrival of multielectrode cochlear implant enhanced considerably speech understanding for cochlear implant users. Nevertheless, channel interaction is an inherent problem of multielectrode stimulation. Indeed, the overlap of electrical fields stimulates a large number of nerve fibers and can create an overlap among the “neural channels”. Depending on the overlap degree, signals, information, neural integration, and neural processing can be very degraded. Two types of interactions can be identified: the electrical-field overlap and the neural-interaction.

In the best case, there is a low interaction degree: neurons are well preserved and the electrode array is close to the cochlear cells, each electrode stimulates a reduced number of auditory neurons. In an intermediate case, the population of activated neurons overlaps the territory of the neighbor electrodes. In the worst case, all electrodes stimulate the same neuron population like a “pseudo-monoelectrode”.

For the three cases, it is assumed that neurons are homogeneously reparted and that the current spread is regular along the electrode array. In reality, this is not what happens [13].

2.3 Measurement of channel interaction.

Channel interaction can be measured using psychophysical tests, for example:

- The modification of the loudness induced by the overlap between electrical fields when two electrodes are activated in phase or out of phase.
- The refractory period of nerve cells which changes the detection thresholds following a previous stimulation. [14].

Objective measurements can also highlight channel interaction.

At the electrical level, the Electric Field Imaging (EFI) method can measure directly the overlap. The electrical stimulus itself is recorded along the electrode array [15].

At the neural level, there is the Electrically Evoked Compound Action Potential (ECAP). The ECAP shows a synchronization of cochlear fibers electrically stimulated, which is equivalent to the Wave I of the auditory brainstem response (ABR) (Fig. 2) [16]. For cochlear implant users, an electrical stimulation by an electrode induces a neural potential recorded by another electrode in the array [17].

At a higher neural level, ABR curves reflect the electrical response of the auditory nerves after a stimulation. Five waves are measurable and they correspond to five generators inside the brainstem. Wave I is generated by the auditory structures close to the cochlea, Wave II is generated when entering in the cochlear nucleus, Wave III comes from the superior olivary complex, Wave IV comes from the lateral lemniscus and finally, Wave V comes from the inferior colliculus of the midbrain. ABR waveforms are illustrated in figure 2 [18].

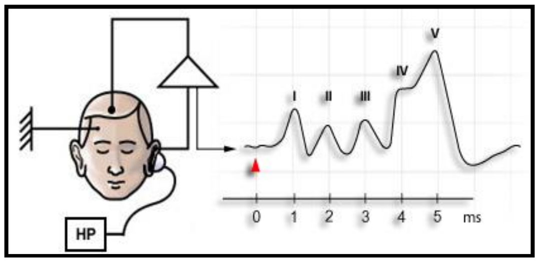

Figure 2. ABR waveforms and recoding-electrodes placement [18]

Guevara et al. developed a new method to evaluate channel interaction using eABR (e stands for electrical) with cochlear implant users. They compared the Wave V amplitude in response to a stimulation of five electrodes together, to the sum of the amplitudes of Waves V after a one-by-one stimulation. They define the Monaural Interaction Component (MIC) as the ratio of those amplitudes. Figure 1 shows the ABR recordings obtained using their method [19]. If there is no channel interaction, the MIC ratio equals 1. On the contrary, if there is channel interaction, the MIC ratio is superior to 1.

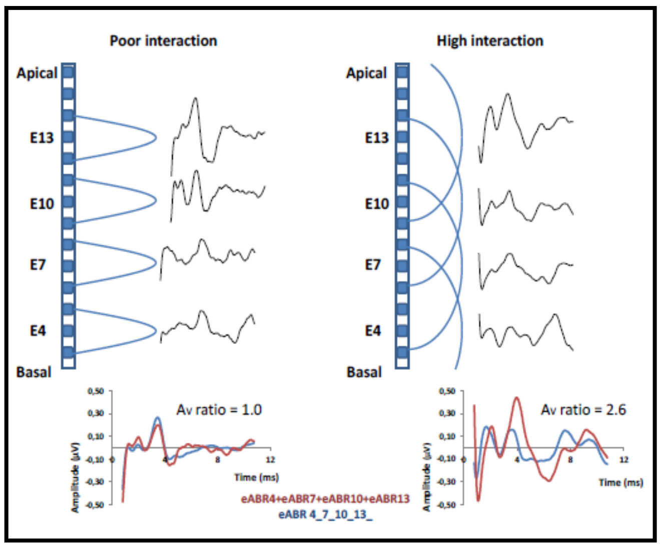

Figure 3. Schematic representation of the electrical interaction assessment method using eABR [19]

2.4 Overcoming channel interaction

Audiologists and engineers developed methods to address the problem of channel interaction.

2.4.1 Stimulation modes

There are several modes of stimulation which vary the current pathways, based on bipolar and monopolar stimulations (figure 4).

Cochlear implants can work on bipolar mode. The current flows between two electrodes on the array. When the electrodes are side by side, the stimulation is focused and when the distance between the electrodes is larger, the stimulation covers a larger surface. However, the bipolar mode is not hyper-selective and a very focused stimulation can activate a larger population of neural fiber than expected.

Cochlear implants can also work on monopolar mode. The active electrode is on the electrode array inside the cochlea while the ground electrode is inside the mastoid or in the implant body. Monopolar mode usually stimulates more nerve fibers than the bipolar mode, nevertheless the monopolar mode requires lower stimulation levels to obtain a perception threshold than the bipolar mode.

The spread of stimulation in monopolar mode is a problem only when there are simultaneous stimulations which exacerbate channel interaction [13, 20-21]. There are other stimulation modes like the common-ground mode or the tripolar mode but they are less common than bipolar and monopolar.

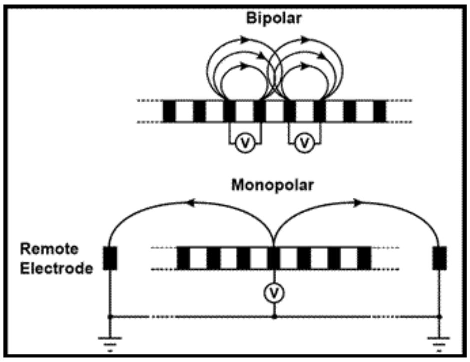

Figure 4. Bipolar and monopolar mode graphic representation. From Clark GM. Cochlear implants Fundamental and application [22]

2.4.2 Coding strategies

Another way to lower channel interaction is to use sequential stimulation. The CIS strategy was the first to integrate this approach. Now, most modern coding strategies integrate sequential stimulations in order to lower the electrical field overlap [23]. Nevertheless, sequential stimulations can limit the stimulation rate so, some manufacturers use NofM coding strategies (for N out of M) which activate only the N most energetic electrodes (N<M). The number of active electrodes in each stimulation sequence is reduced and so it lowers the overlap (but also the spectral range).

The order of electrode activation is also important. It is possible to stimulate from apex to base but it is not optimal to reduce channel interaction. An optimal sequence can be used in order to maximize the distance between two successive stimulations. For example, with a 6-channel implant, electrodes can be activated in this order: 6-3-5-2-4-1 [24].

But even when two remote electrodes are sequentially activated, high stimulation levels, short temporal delay neural and membrane residual polarization can induce channel interaction [25].

3. PSYCHOPHYSICAL TUNING CURVES

3.1 Auditory filters

Inner ear capacity to discriminate frequencies in sounds is related to the auditory filters' selectivity. In the literature, the concept of critical bands describes the frequency bandwidth of the auditory filters. Two sounds can be discriminated only if they are in two different critical bands. It is like a band-pass filter bank with varying cut-off frequencies along the cochlear ramp which depend upon the spectrum of the sound [26]. Auditory filters bandwidth can be assessed by a masking method leading to psychophysical tuning curves (PTCs).

On PTCs, each point represents the masking level needed for a specific frequency to mask a reference frequency. The sharper the PCT tip is, the more selective the cochlea is. The PTC is mainly characterized by the Q10, which is the ratio between the reference frequency and the bandwidth at 10 dB above the tip (Fig. 5) [27].

Hearing impairment changes PTCs' shape. Hearing impaired people have flat and large PTCs showing high auditory thresholds and a poor frequency selectivity [28].

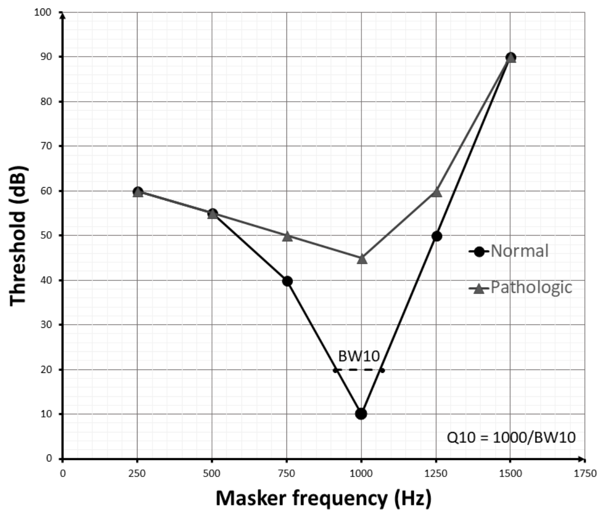

Figure 5. Normal and pathological psychophysical tuning curves for a 1000Hz reference frequency

3.2 Psychophysical tuning curves assessment

PTCs can be established using two methods: Simultaneous masking or forward masking. For both methods, it measures the minimum masking level, of a sound on another.

In simultaneous-masking, the masker and the reference sound are presented simultaneously. In forward-masking, they are presented time-shifted [29-30].

For normal-hearing subjects and for hearing-impaired people PTCs are measured with acoustical stimulations. With cochlear implant users, it can be done with acoustical stimulations through the sound processor, or with direct electrical stimulations controlling the electrode array. The paradigm of measurement is very similar between acoustical and electrical stimulation so we can compare easily PTCs from cochlear implant users and normal-hearing subjects.

With electrical stimulations, simultaneous masking needs high stimulation levels which restrains to get precise PTCs in term of frequency. Forward-masking needs lower stimulation levels than simultaneous masking. Lower stimulation levels enable to quantify frequency selectivity more accurately [31].

An algorithm can mimic the sound processing of a cochlear implant and simulate an impaired cochlea. This kind of simulation enables to test normal-hearing subjects in parallel with cochlear implant users and to compare the results. PTCs are equally flat for cochlear implant users and for normal-hearing subjects using a simulator, but there is a larger inter-individual variability with cochlear implant users [32].

Interestingly PTCs on cochlear implant users can account for channel interaction. Masking levels needed for one frequency reflect the spread of stimulation of the corresponding electrode [33].

4. ON THE NUMBER OF CHANNELS

Increasing the overall number of channels does not always improve the transfer of information. Speech understanding is often higher when there are few channels (<20). Some studies suggest that we need to choose carefully the stimulating channels to avoid the overlap of the information [1].

Speech understanding, with no background noise, is improved when the number of channels is increased up to 10. Beyond 10 channels, speech understanding remains constant and it suggests that cochlear implant users can't take advantage of the additional information. The reason for this limitation is not yet clearly identified but channel interaction seems to be a part of it [2].

On the other side, in background noise, more than 10 channels are needed to maintain speech understanding at a good level. Some experiments, conducted with normal-hearing subjects using a cochlear implant simulator, showed that more channels are needed in background noise, but that selecting more than 12 channels out of 16 did not yield to better scores [34]. Nevertheless, other studies suggested an increase in speech understanding up to 30 channels [35]. This question remains open.

5. COCHLEAR IMPLANT SIMULATOR

5.1 The vocoder

As indicated above, modern cochlear implants work like a vocoder [36].

Because of the enhancement of signal processing through the years, scientists wanted to know how well the channels of the cochlear implant could transmit information. To do so, some of them used an algorithm mimicking the sound processing of the cochlear implant. When electric stimulations are replaced by bandpass noises or pure tones, we can simulate the auditory sensation of a cochlear implant. Dorman et al. stated that experiments conducted with such a simulator are reliable and comparable with experiments conducted with cochlear implant users. But this is an "ideal case", because the auditory cells of cochlear implant users are damaged and because cochlear implants cannot exactly reproduce a normal auditory perception [37].

5.2 Simulating channel interaction

To go further than the "ideal case" it is possible to simulate some features of hearing impairment. One of these features is channel interaction.

Several experiments have been conducted with vocoders integrating a channel interaction module [38].

One method to mimic channel interaction is to add in each channel a part of the envelope information of the neighbor channels. Using this method is like creating an overlap between the channels of the vocoder [39].

Vershuur et al. suggested that with an accurate simulation of channel interaction it is possible to recreate, with normal-hearing subjects, the results obtained in the best cases with cochlear implant users [40].

6. FUTURE EXPERIMENTS

Our team is interested in optimizing the number of channels depending on the level of background noise and the degree of channel interaction. The idea of conducting this study comes from results we obtained in a previous experiment conducted in our laboratory [41]. Our results suggest that speech recognition scores are higher when N = M and that channel interaction is a limiting factor.

First, using a vocoder we can mimic the signal processing of a cochlear implant and we can add a channel interaction simulator with different interaction degrees.

Then, channel interaction can be measured with cochlear implant users using PTCs and eABR [19].

And finally, gathering all the results could determine a way to optimize the features of cochlear implants considering channel interaction in the fitting process.

7. CONCLUSION

Channel interaction is still under investigation and some well-known tools are capable to measure it. We want to combine those reliable tools with new protocols to see how much channel interaction affects each patient and try to optimize their cochlear implant.

ACKNOWLEDGMENTS

The authors would like to thank the people and the institutions for their expertise and collaboration to this study; especially Dr. Dan Gnansia, Pr. Hung Thaï-Van and the whole team of the ORL department of the Edouard Herriot Hospital, and the Hospitals of Lyon.

REFERENCES

[1] Berger-Vachon C, Collet L, Djedou B, Morgon A. (1992). Model for understanding the influence of some parameters in cochlear implantation. Ann. Otol. Rhinol. Laryngol 101(1): 42-45.

https://doi.org/10.1177/000348949210100112

[2] Friesen LM, Shannon RV, Baskent D, Wang X. (2001). Speech recognition in noise as a function of the number of spectral channels: Comparison of acoustic hearing and cochlear implants. J. Acoust. Soc. Am 110(2): 1150-1163. https://doi.org/10.1121/1.1381538

[3] Berger-Vachon C, Cucis PA, Truy E, Van HT, Gallego S. (2018). Cochlear implants: Consequences of microphone aging on speech recognition. Complex Systems: Solutions and Challenges in Economics, Management and Engineering, Springer, Cham, pp. 497-519. https://doi.org/10.1007/978-3-319-69989-9_30

[4] Cucis PA, Berger-Vachon C, Truy E, Gallego S. (2017). Implant Cochléaire: Codage et interaction de voisinage entre les électrodes. JCJC'2017: Conference for Young Researchers, Paris, France, pp. 87-96.

[5] Cucis PA, Berger-Vachon C, Truy E, Thai-Van H, Millioz F, Gallego S. (2016). Influence de la stratégie de codage de l'implant cochléaire sur la reconnaissance des syllabes en milieu bruité. in Handicap 2016, Paris, France, pp. 39-44.

[6] Cucis PA, Berger-Vachon C, Millioz F, Truy E, Gallego S. Hearing in noise: Importance of coding strategies. normal hearing subjects and cochlear implant users. Manuscr. Submitt. Publ.

[7] Cucis PA, Berger-Vachon C, Truy E, Thai-Van H, Gallego S. Influence of microphone soiling on syllable recognition in cochlear implants. Simulation and recognition in noise. Acta Otolaryngol. (Stockh.), Accepted for publication.

[8] Djourno A, Eyries C. (1957). Prothese auditive par excitation electrique a distance du nerf sensoriel a laide dun bobinage inclus a demeure. Presse Med 65(63): 1417.

[9] Chouard CH. (2010). Histoire de l'implant cochléaire. Ann. Oto-Laryngol. Chir. Cervico-Faciale 127.

[10] Clark GM. (2006). The multiple-channel cochlear implant: the interface between sound and the central nervous system for hearing, speech, and language in deaf people—a personal perspective. Philos. Trans. R. Soc. B Biol. Sci 361(1469): 791-810. https://doi.org/10.1098/rstb.2005.1782

[11] Churchill TH, Kan A, Goupell MJ, Ihlefeld A, Litovsky RY. (2014). Speech perception in noise with a harmonic complex excited vocoder. J. Assoc. Res. Otolaryngol. JARO 15(2): 265-278. https://doi.org/10.1007/s10162-013-0435-7

[12] Wilson BS, Finley CC, Farmer JC, Lawson DT, Weber BA, Wolford RD, Kenan PD, White MW, Merzenich MM, Schindler RA. (1988). Comparative studies of speech processing strategies for cochlear implants. The Laryngoscope 98(10): 1069-1077. https://doi.org/10.1288/00005537-198810000-00009

[13] Shannon RV. (1983). Multichannel electrical stimulation of the auditory nerve in man. II. Channel interaction. Hear. Res 12(1): 1-16. https://doi.org/10.1016/0378-5955(83)90115-6

[14] Tang Q, Benítez R, Zeng FG. (2011). Spatial channel interactions in cochlear implants. J. Neural Eng. 8(4): 046029. https://doi.org/10.1088/1741-2560/8/4/046029

[15] Mens LHM. (2007). Advances in cochlear implant telemetry: Evoked neural responses, electrical field imaging, and technical integrity. Trends in Amplif 11(3): 143-159. https://doi.org/10.1177/1084713807304362

[16] Abbas PJ, Hughes ML, Brown CJ, Miller CA, South H. (2004). Channel interaction in cochlear implant users evaluated using the electrically evoked compound action potential. Audiol. Neurotol 9(4): 203-213. https://doi.org/10.1159/000078390

[17] Charasse B. (2003). Etude et optimisation d'un système de télémétrie des réponses neurales (NRTtm) dans la perspective d'un réglage automatisé de l'implant cochléaire. Ph.D. Thesis, Claude Berbard University Lyon 1.

[18] Audition - Oreille - Cochlée. [Online]. Available: [Accessed: 19-Jan-2018]. http://www.cochlea.eu

[19] Guevara N, Hoen M, Truy E, Gallego S. (2016). A cochlear implant performance prognostic test based on electrical field interactions evaluated by eABR (Electrical Auditory Brainstem Responses). PLOS ONE 11(5): e0155008. https://doi.org/10.1371/journal.pone.0155008

[20] Bierer JA. (2007). Threshold and channel interaction in cochlear implant users: evaluation of the tripolar electrode configuration. J. Acoust. Soc. Am 121(3): 1642–1653. https://doi.org/10.1121/1.2436712

[21] Chatterjee M, Galvin JJ, Fu QJ, Shannon RV. (2006). Effects of stimulation mode, level and location on forward-masked excitation patterns in cochlear implant patients. J. Assoc. Res. Otolaryngol 7(1): 15–25. https://doi.org/10.1007/s10162-005-0019-2

[22] Clark G. (2006). Cochlear Implants: Fundamentals and Applications. Springer Science & Business Media.

[23] Wilson BS, Finley CC, Lawson DT, Wolford RD, Eddington DK, Rabinowitz WM. (1991). Better speech recognition with cochlear implants. Nature 352(6332): 236–238. https://doi.org/10.1038/352236a0

[24] Loizou PC. (1998). Mimicking the human ear. IEEE Signal Process. Mag 15(5): 101–130. https://doi.org/10.1109/79.708543

[25] De Balthasar C, Boëx C, Cosendai G, Valentini G, Sigrist A, Pelizzone M. (2003). Channel interactions with high-rate biphasic electrical stimulation in cochlear implant subjects. Hear. Res 182(1-2): 77-87. https://doi.org/10.1016/S0378-5955(03)00174-6

[26] Traunmüller H. (1990). Analytical expressions for the tonotopic sensory scale. J. Acoust. Soc. Am 88(1): 97-100. https://doi.org/10.1121/1.399849

[27] Stelmachowicz PG, Jesteadt W, Gorga MP, Mott J. (1985). Speech perception ability and psychophysical tuning curves in hearing-impaired listeners. J. Acoust. Soc. Am 77(2): 620-627. https://doi.org/10.1121/1.392378

[28] Nelson DA. (1991). High-level psychophysical tuning curves: Forward masking in normal-hearing and hearing-impaired listeners. J. Speech Hear. Res 34(6): 1233-1249. https://doi.org/10.1044/jshr.3406.1233

[29] Kluk K, Moore BCJ. (2005). Factors affecting psychophysical tuning curves for hearing-impaired subjects with high-frequency dead regions. Hear. Res 200(1-2): 115-131. https://doi.org/10.1016/j.heares.2004.09.003

[30] Moore BC. (1978). Psychophysical tuning curves measured in simultaneous and forward masking. J. Acoust. Soc. Am 63(2): 524–532. https://doi.org/10.1121/1.381752

[31] Nelson DA, Donaldson GS, Kreft H. (2008). Forward-masked spatial tuning curves in cochlear implant users.

J. Acoust. Soc. Am 123(3): 1522-1543. https://doi.org/10.1121/1.2836786
[32] Langner F, Jürgens T. (2016). Forward-masked frequency selectivity improvements in simulated and actual cochlear implant users using a preprocessing algorithm. Trends Hear 20. https://doi.org/10.1177/2331216516659632
[33] Kirby VM, Nelson DA, Soli SD, Fortune TW. (1987). Channel interactions measured by forward-masked 'place' tuning curves with multichannel electrical stimulation. J. Acoust. Soc. Am 82(S1): S72-S72. https://doi.org/10.1121/1.2024956
[34] Shannon RV, Fu QJ, Galvin J. (2004). The number of spectral channels required for speech recognition depends on the difficulty of the listening situation. Acta Oto-Laryngol. Suppl (552): 50-54. https://doi.org/10.1080/03655230410017562
[35] Loizou PC, Dorman MF, Tu Z, Fitzke J. (2000). Recognition of sentences in noise by normal-hearing listeners using simulations of speak-type cochlear implant signal processors. Ann. Otol. Rhinol. Laryngol. Suppl 185: 67-68. https://doi.org/10.1016/S0165-5876(00)00419-5
[36] Loizou PC. (2006). Speech processing in vocoder-centric cochlear implants. Adv. Otorhinolaryngol 64: 109-143. https://doi.org/10.1159/000094648
[37] Dorman MF, Loizou PC, Fitzke J, Tu Z. (2000). Recognition of monosyllabic words by cochlear implant patients and by normal-hearing subjects listening to words processed through cochlear implant signal processing strategies. Ann. Otol. Rhinol. Laryngol. Suppl 185: 64-66. https://doi.org/10.1016/S0165-5876(00)00419-5
[38] Gnansia D, Péan V, Meyer B, Lorenzi C. (2009). Effects of spectral smearing and temporal fine structure degradation on speech masking release. J. Acoust. Soc. Am 125(6): 4023-4033. https://doi.org/10.1121/1.3126344
[39] Crew JD, Galvin JJ, Fu QJ. (2012). Channel interaction limits melodic pitch perception in simulated cochlear implants. J. Acoust. Soc. Am 132(5): EL429-435. https://doi.org/10.1121/1.4758770
[40] Verschuur C. (2009). Modeling the effect of channel number and interaction on consonant recognition in a cochlear implant peak-picking strategy. J. Acoust. Soc. Am 125(3): 1723-1736. https://doi.org/10.1121/1.3075554
[41] Cucis PA, Berger-Vachon C, Truy E, Thai-Van H, Millioz F, Gallego S. (2016). Cochlear implants : Influence of the coding strategy on syllable recognition in noise. AMSE J.-AMSE IFRATH Publ. –2016 Model. C 77(2): 84-97.