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Simulations of Hearing Loss and Hearing Aid: Effects on Electrophysiological Correlates of Listening Effort

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Abstract— **In the last decades, many investigations were done to examine the effects of sensorineural hearing loss on the speech perception ability. Besides testing hearing impaired persons, there is also the possibility to simulate the hearing loss. Therefore, some electrophysiological as well as speech recognition studies were performed in normal hearing subjects using techniques to model the sensorineural hearing loss. Thus, the effects of peripheral hearing loss without central auditory pathologies can be examined.**

In previous studies, we have shown, that the wavelet phase synchronization stability (WPSS) of auditory late responses could serve as a possible indicator of listening effort. Now, the aims of this present study were to explore the effects on the WPSS by using two different simulations of hearing loss and a simulated hearing aid. The preliminary results showed, that in case of a simultaneous simulation of hearing loss by noise masking and a hearing aid, an objective discrimination between an easy and a difficult listening situation can be achieved. Furthermore, the WPSS reflected also a good discrimination by using the filtered and attenuated paradigms.

I. INTRODUCTION

Individuals with sensorineural hearing loss have often a poor speech perception ability. This diminished speech perception can be caused by various factors, like a reduction in frequency selectivity, intensity or temporal resolution and recruitment [1], [2], [3]. In the last years, in order to test this peripheral hearing loss without central auditory pathologies, some electrophysiological [4], [5] as well as speech recognition studies [1], [2] have been performed in normal hearing subjects using techniques to model the sensorineural hearing loss. This simulation of the hearing loss can either be achieved by spectrally shaped noise masking or by filtering the speech signal (see [3] for a review). The first method can be used to increase the hearing thresholds in selected spectral regions and could also be used to simulate a recruitment effect, whereas filtering can be applied to simulate the effects of sensorineural hearing loss on consonant feature recognition [2].

A further problem of hearing impairment is listening effort. This perceptual effort can occur, when more attentional and cognitive resources are needed to understand speech in adverse listening situations [6].

Today's hearing aids have already some features, which could reduce the listening effort. However, an objective measure of the listening effort to fit the hearing aids to the individual needs remains an unsolved problem. Thus, the focus of our ongoing research is the objective estimation of listening effort.

In previous studies [7], [8], [9], we have shown a feasible method to extract electrophysiological correlates of listening effort in young subjects by determining the wavelet phase synchronization stability (WPSS) of auditory late responses by using tonebursts and noise embedded syllabic paradigms. The WPSS, which can be calculated after the extraction of the phase of ALR sequences, serves as an indicator of the cognitive effort, which the subject requires to solve an auditory paradigm.

The objective of this ongoing study was to investigate the effects of decreased audibility on the WPSS by using two different hearing loss simulation techniques. Thus, the syllabic stimulation paradigms were filtered or a spectrally shaped masking noise was applied. Furthermore, we wanted to examine the effect of a hearing aid on the WPSS. Therefore, the syllabic paradigms were adapted by a prescriptive formula (NAL-R [10]) to achieve the required gains.

II. MATERIALS AND METHODS

A. Subjects and Data Acquisition

Fourteen young normal hearing subjects participated in the study (26± 4.11 years, 8M/ 6F). The subjects had normal hearing thresholds (<15dB (HL)) and showed no history of hearing problems. After a detailed explanation of the procedure, all subjects signed a consent form, followed by the acquisition of the audiogram and electrodes placement. Ag/AgCl-electrodes were attached at: the right mastoid (ipsilateral to the stimulus), vertex (common reference) and upper forehead (ground). Electrodes impedances were always below 5kΩ. The electroencephalographic activity was collected by means of a biosignal amplifier (gUSBamp, g.tec, Austria) with a sampling frequency of 512Hz. Further, the signals were bandpass-filtered in the range of 1-30Hz, and movement artifacts were removed by an amplitude threshold of 50μV.

B. Construction and Processing of the Syllabic Paradigms

The speech stimuli were consonant-vowel syllables spoken by a female speaker using a sampling frequency of 16kHz.

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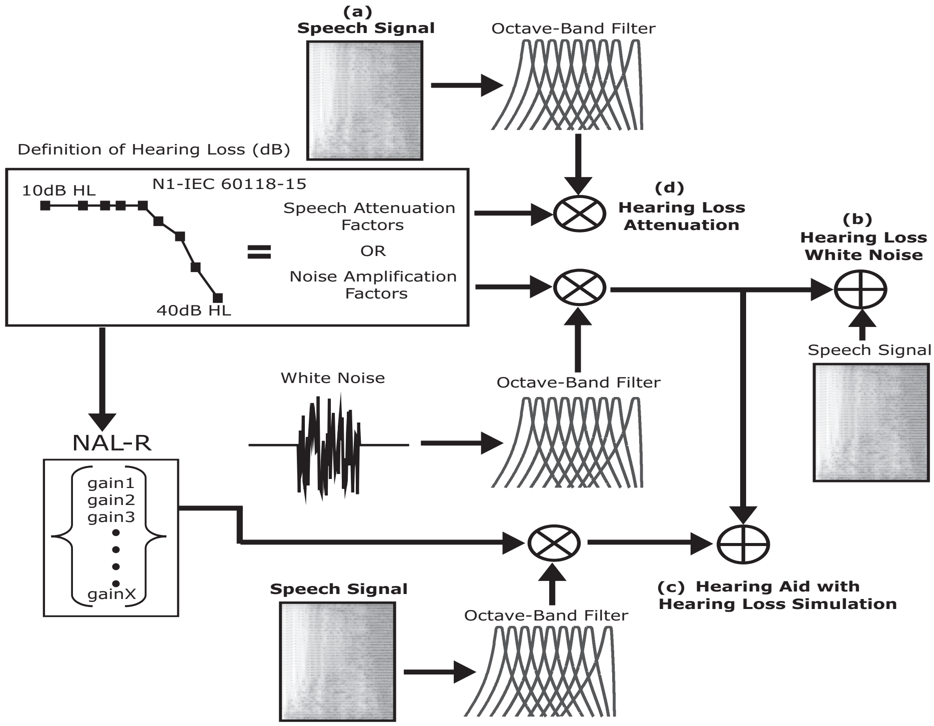

Fig. 1. Construction and processing of the syllabic paradigms. Four conditions were designed: (a) Control condition (unprocessed syllabic paradigms), (b) simulation of hearing loss by noise masking, (c) simulation of hearing loss by noise masking and simulation of a hearing aid, (d) simulation of hearing loss by attenuation.

Subsequently, the amplitudes of the syllables were normalized, and a window was applied. Such a window consisted of three major parts: a rise and a fall time, which were the first and second halves, respectively, of a Gaussian window with a total duration of 50ms; and a plateau time of 150ms with a flat amplitude of 1. Each syllable had a duration of 200ms and they were calibrated after windowing by acquiring their peak equivalent Sound Pressure Level (peSPL), for details see [11].

The study had four conditions, as illustrated in Fig. 1: (a) control condition (only syllabic stimulation), (b) syllabic stimulation plus simulation of hearing loss by noise masking, (c) syllabic stimulation plus simulation of hearing loss by noise masking and simulation of a hearing aid, (d) syllabic stimulation plus simulation of hearing loss by attenuation.

Each of these 4 conditions was tested using two different syllabic paradigms. These paradigms had a different level of complexity, which was accomplished by the combination of the syllables, resulting in:

"Difficult Syllable Paradigm (DSP)": this paradigm should be more difficult to solve because the syllables had the same vowel and different plosives.

"Easy Syllable Paradigm (ESP)": this paradigm should be easier to solve, because the syllables had different vowels and consonants.

In order to maximize the entropy of the experiment, such that their solution requires an effortful task, in all paradigms the syllables had randomized order and randomized interstimulus interval (ISI) ranging from 1-2s. As it can be seen in Fig. 1, a mild hearing loss (10dB HL @ frequencies of 0.25, 0.5 and 1kHz, 15dB HL @ 2kHz, 20dB HL @ 3kHz, 30dB HL @ 4kHz, 40dB HL @ 6kHz) was selected, defined in [12] as N1. Subsequently, a one–third octave band filter bank according to [13] was used in order to separate the different frequency bands either of the speech or noise signals. The filter consisted of 18 channels with center frequencies f_c spaning from 250Hz to 6kHz [1], [2]. The control condition (a) included only the syllabic paradigms without further postprocessing, and they were presented at an intensity level of 65 dB peSPL, which represents a conversational speech level [4]. In condition (b), the noise masking hearing loss was achieved by using the one–third octave band filter bank and white noise. The filtered white noise was amplified in the specific frequency channels, whose amplification values corresponded to the dB HL values of the hearing loss at each frequency band. Also, a correction factor for each frequency band (ranging from 4 to 6 dB SPL), which is also used to achieve the effective masking levels for clinical masking in audiometers [14] and a 10 dB SPL safety factor [14] were added to the processed noise. Missing correction factors were also calculated by using linear interpolation. After application of the amplitude factors, the intensity of the masking noise was 35 dB SPL. Condition (c) comprised the simulation of a hearing aid plus the simulation of a hearing loss, in this case we used the noise masking hearing loss constructed for condition (b). The hearing aid was simulated by using the NAL-R formula [10] in order to calculate the corresponding gain factors for the selected hearing loss type.

So that the intensity of the simulated hearing aid was around 87 dB SPL. For condition (d), the simulation of hearing loss by attenuation, the filtered speech signals were attenuated using the hearing loss values from the normal presentation level of 65dB peSPL.

C. Experimental Procedure, Component Identification and Inclusion Criteria

The subjects were instructed to pay attention to the syllables and press a button after the detection of the target syllable, which was in all the cases the same syllable. The whole experiment lasted 1.5 hour and the paradigms were presented in randomized order. A short pause was made between each paradigm to avoid fatigue. The N1 wave was visually identified as the most negative peak in the time interval of 50-200 ms. Furthermore, a correlation waveform index (cwi) was defined in order to obtain an objective waveform analysis. Thus, the ALR single sweep sequences were sorted into a matrix. After that, the index was achieved from each subject by separation of those single sweep matrices of every dataset in odd x_i and even sweeps y_i. Then, the correlation coefficient $\rho\epsilon[-1,1]$ was calculated [15] as defined by

$$\rho := \frac{\sum_{i=1}^{N}(x_i-\overline{x})(y_i-\overline{y})}{\sqrt{\sum_{i=1}^{N}(x_i-\overline{x})^2}\sqrt{\sum_{i=1}^{N}(y_i-\overline{y})^2}}, \qquad (1)$$

where $\overline{x} := N^{-1}\sum_{i=1}^{N} x_i$ and $\overline{y} := N^{-1}\sum_{i=1}^{N} y_i$. The cwi was set to 0.5 in the range of the N1-P2 complex as also used in [16]. The subjects inclusion criteria for the study was: (1) they had an identifiable waveform of the N1-P2 complex in the ALRs and fulfilled the cwi; and (2) they detected correctly at least 80% of the target syllables, which serves as a control of the cooperation of the subject. From the 14 measured subjects two subjects were excluded because of criterion (2).

D. Synchronization Stability and Listening Effort

For the analysis of the ALRs, we used the WPSS that was introduced in [17], [18], [19] for the quantification of auditory attention in ALR single sweeps. The larger the WPSS, the larger the effort, resulting from an increased attention to detect the target syllable. Furthermore, we defined in [9], the measure Listening Effort (LE) as the mean of the WPSS in the range of the N1 wave for a specific scale. For a more detailed explanation in the extraction of phase synchronization stability and its relation to listening effort we refer to [8]. The wavelet used in this study was the 4th–derivative of the complex Gaussian function, as in [7], [8], [18], [19].

III. RESULTS & DISCUSSION

The WPSS was calculated for each subject and condition for the scale $a = 40$ using 70 ALR sweeps, which were evoked by correctly detected target syllables. This scale was also selected in previous studies [8], because a good temporal localization of the maximum of the WPSS in the expected range of the N1-P2 complex (approx. 50 to 250 ms) can be achieved. The Fig. 2 shows the grand average of the normalized WPSS (left side) and the corresponding LE-levels (right side) for each paradigm (DSP (black line) and ESP (gray line)). From top to bottom the results of the four conditions are plotted ((a): control condition, (b): simulation of hearing loss by noise masking, (c): simulation of hearing loss by noise masking and simulation of a hearing aid, (d): simulation of hearing loss by attenuation).

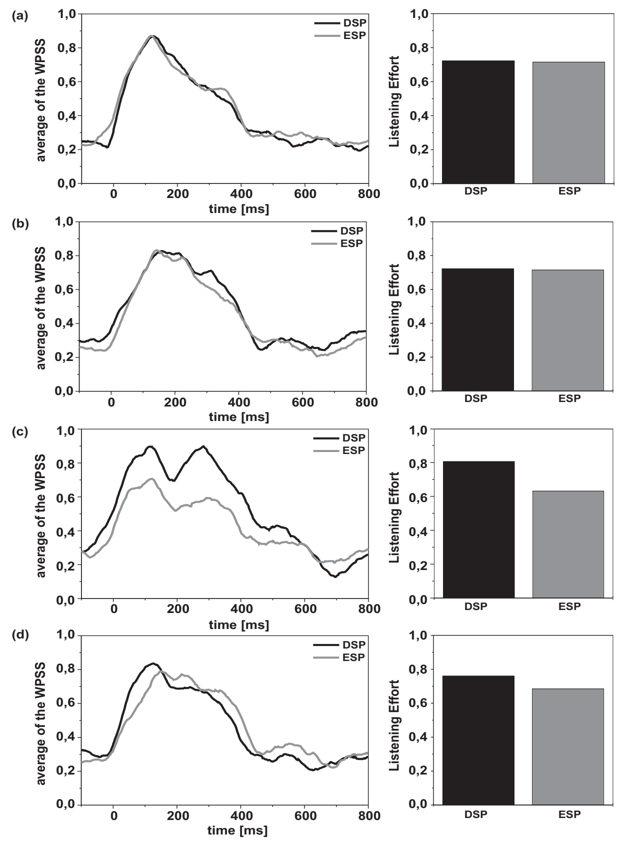

Fig. 2. Left: Grand normalized average of the WPSS (over all the 12 included subjects) for both paradigms (DSP (black) and ESP (gray)). Right: Corresponding LE-levels (mean WPSS in the interval of the N1 wave). From top to bottom are the results of the four conditions ((a): control condition, (b): simulation of hearing loss by noise masking, (c): simulation of hearing loss by noise masking and simulation of a hearing aid, (d): simulation of hearing loss by attenuation) plotted. Note, that the difference of the WPSS in condition (c) and (d) reached statistical difference ($p<0.05$).

Here, it is worth mentioning that the ESP & DSP paradigms were also tested in a previous study [8], [9]. There, the paradigms were embedded in multitalker babble noise at different signal to noise ratios (SNRs). Thus, it was difficult to detect the target syllable in the noisy surrounding. In the present study, the noise was removed, which made the differentiation of the syllables relatively easy in both cases. This can be seen in Fig. 2 for the unprocessed syllabic paradigms (condition (a)), where the WPSS of both paradigms is in the same range. In addition, similar results were achieved in a separate ongoing study [20], where the effects of hearing impairment and age were examined. There, from the three tested groups, only the younger participants displayed and reported no difference regarding the effort needed to solve both paradigms. This interpretation is also in line with the subjective ratings of the participants. Most of them reported, that both paradigms were facile to solve. The figures below show the WPSS and the LE-levels for the simulation of hearing loss by noise masking. Again, the difference of the WPSS between the paradigms is very small. Only a slight enhancement of the WPSS for the DSP (black line) can be seen around 200ms. This shift of the WPSS peak could be related to a possible increase of the N1 latency resulting from the introduced spectrally shaped noise. Furthermore, a complete masking of the syllables was not fully guaranteed due to the resultant SNR, which can be decreased in a future study. In condition (c), where the hearing aid and the hearing loss was simulated, the subjects reported due to the unusual sound of the adapted syllables, that the detection of the target syllable was difficult. This can also be observed objectively in the graph of the WPSS and the corresponding LE-levels. Both showed an increase for the DSP compared to the ESP. Furthermore, this result reached statistical significance ($p<0.05$). Here, it could also be interesting to examine possible training effects on the WPSS after some repetitions of the paradigms, so that the subjects get used to the sound of the processed syllables and perhaps less effort is required to detect the target syllable. This effect could also be useful for testing hearing aid wearers, because of their assimilation to the hearing aid.

The last condition, which is illustrated at the bottom of this figure, represents the results from the simulated hearing loss by attenuation. Here, the WPSS of the DSP is again larger in the expected time interval of the N1 wave, which also reached significance. In this case, the syllables were modified by filtering and attenuation, so that again more attention was needed to differentiate the syllables, especially in the DSP. In addition, comparing the difference of the LE-levels between the conditions (c) and (d), it can be seen that the difference is larger for the simulated hearing aid. This could be related to the different intensities. It can be interpreted, that in the hearing aid condition, were the syllabic paradigms were amplified, the paradigms awakened attention, so that the subjects payed more attention to detect the target syllable.

IV. CONCLUSIONS

In this study, the effects of a simulated hearing loss and a simulation of a hearing aid on the WPSS were examined. The results showed, that in case of a simultaneous simulation of hearing loss by noise masking and a hearing aid, an objective discrimination between an easy and a difficult listening situation can be achieved. Furthermore, the WPSS reflected also a good discrimination by using the filtered and attenuated paradigms. However, these first results are encouraging, but an improvement of the syllabic paradigms is still necessary. Therefore, a part of our future work will be to increase the degree of hearing loss in order to decrease the SNR of the noise masking conditions. Thus, the hearing thresholds of the participants will be more elevated to achieve a more effortful task. Additionally, a non-linear amplification formula can be applied to the syllabic paradigms to limit the total hearing aid gain. This will be done to avoid higher stimulation intensities above a comfortable loudness level.

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