A method to analyse and test the automatic selection of hearing aid programs

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Digital hearing aids usually provide different hearing aid programs. This means different settings can be selected to adapt the signal processing to different hearing situations. Furthermore, advanced devices often include a classification algorithm that continuously analyses the acoustic environment and automatically selects a hearing aid program accordingly. However, there exists no method to analyse this adaptive feature. Therefore, we present a possibility to analyse and test which hearing aid program is active in a specific hearing situation. To proof the concept, hearing aids of two different manufacturers are analysed. These results give insights into the differences between classification strategies and classification quality among hearing aid manufacturers. Moreover, it shows that some signals, which humans can easily classify, are difficult to classify for hearing aids. Furthermore, the result of one device is compared with the classification entries of the data logging feature, which shows good agreement and verifies the new method. In addition, this comparison shows that the new method allows for a more comprehensive analysis so that using the data logging is no reasonable alternative.

INTRODUCTION

Digital hearing aids usually provide different hearing aid programs. This means the hearing aid can store different set of parameters, defining the signal processing, which is useful to adapt the signal processing to different hearing situations (Schaub, 2008; Husstedt, 2016). For some devices, the user manually selects the desired hearing aid program (see Fig. 1a). For more advanced devices, a classification algorithm continuously analyses the acoustic environment and selects a hearing aid program accordingly (see Fig. 1b). However, neither for the hearing aid user nor for the hearing aid professional is it clear what program the hearing aid actually selects in a specific hearing situation. Manufacturers pursuing different strategies so that different hearing aids may classify the same hearing situation differently. Furthermore, the hearing aid does not always select the proper hearing aid program, since the classification of hearing situations is still a difficult task (Tchorz *et al.*, 2016). A false classification causes an improper signal processing and thus may reduce speech intelligibility, comfort, and user satisfaction.

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In this work, we present a method that allows one to analyse and test the automatic selection of hearing aid programs. With this method, measurement results show which hearing aid program is active in a specific hearing situation. This gives insights into the classification strategy of hearing aid manufacturers and helps to evaluate the performance and quality of the applied classification algorithms. The rest of the paper is organized as follows. First, the measurement procedure is explained in detail. Then, it is demonstrated how the method is applied to two state-of-the-art hearing aids. Moreover, to verify the new method, the result of one device is compared with the entries of the data logging feature. Finally, the results are summarized and a conclusion is drawn.

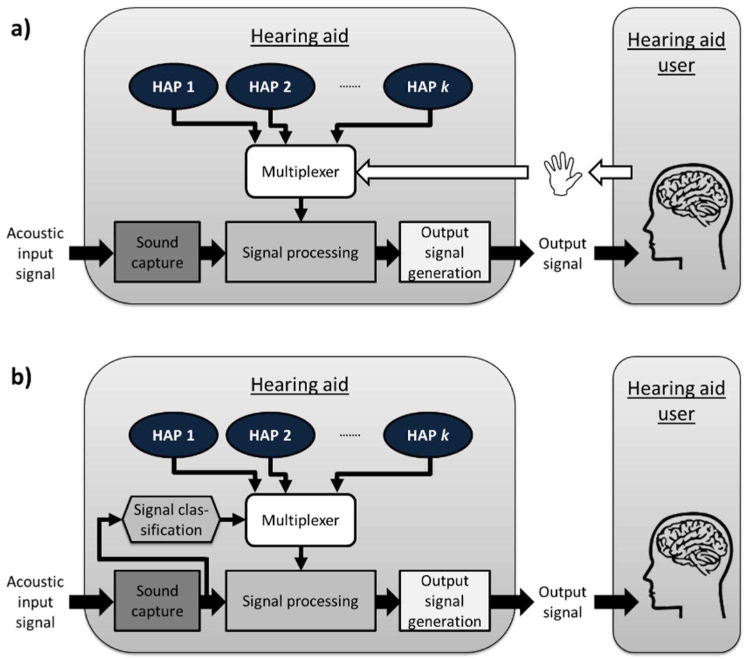

Fig. 1: Visualisation of the manual (a) and automatic (b) selection of hearing aid programs (HAP).

MEASUREMENT PROCEDURE

Preliminary part

In order to analyse the automatic selection of hearing aid programs, it is necessary to choose test signals representative for every hearing situation considered (e.g., music for the music program). Then, every of the n hearing aid programs is configured in an arbitrary way as reference, e.g., with reference test gain (RTG). However, it is not important to have equal configurations for different hearing aid programs. In a next step, every of the test signals is successively presented to the hearing aid and each time the output signal is measured and saved as reference (see upper left part of Fig. 2).

Measurement part

During the measurement part, n measurement cycles are performed where a marker is set just to one hearing aid program at a time. In this context, setting a marker means changing the signal processing of the corresponding hearing aid program so that a change of the output signal is noticeable. For instance, setting a marker could be realized by simply changing the gain for the corresponding hearing aid program. During each measurement cycle, all test signals are presented to the hearing aid, and each time the output signal is measured (see Fig. 2). These signals are then compared with the reference signals saved during the preliminary part. The comparison clearly shows which signal is affected by the marker. In the ideal case, a difference only occurs for that signal where a maker is set to the corresponding hearing aid program. If for example a marker is set to the speech program, a difference should only occur for speech signals. However, if we consider that the output signal is affected when speech is present and a marker is set to the music program, one can conclude that the speech signal is classified as music.

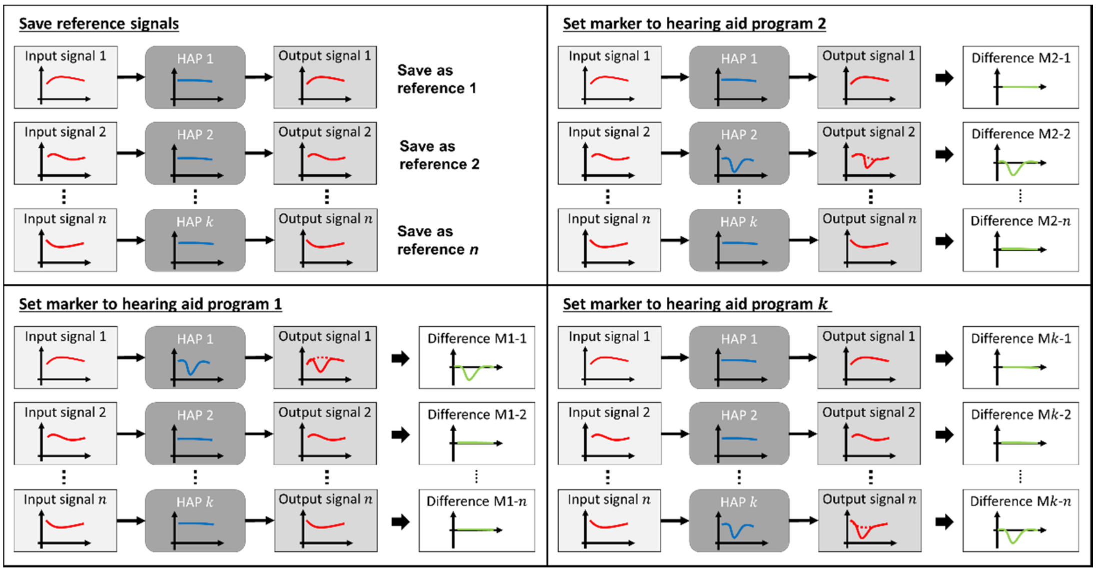

Fig. 2: Visualisation of the preliminary measurement and saving of the reference signals (upper left part). The other parts of the figure visualise the measurement cycles where the maker is set to hearing aid program 1, 2, and k. In this visualisation, only one input signal is presented for each hearing aid program (HAP) so that the number of test signals n is equal to the number of HAPs k. Moreover, the input signal and the corresponding HAP have the same index – e.g., if input signal 1 represents a speech signal, HAP 1 is the speech program.

MEASURMENT

Test signals

A comparison of several hearing aid manufacturers shows that most of the upper class models can at least distinguish between the following listening situations: speech, speech in noise, music, and noise. Thus, these listening situations are considered in the following (see Table 1). Furthermore, it is important to mention that many devices can also classify the situation “quiet”. However, without any input signal, the method is not applicable so that this situation is not considered. Nevertheless, this is no significant limitation, because in quiet, there is no external input signal that can be processed so that it is of minor interest for user what signal processing is active.

Index	Listening situation	Test signal
1	Speech	ISTS 65 dB
2	Speech + Noise	ISTS 65 dB + IFnoise 60 dB
3	Speech + Noise	ISTS 65 dB + IFnoise 50 dB
4	Speech	Audio book 65 dB
5	Speech + Noise	Audio book 65 dB + IFnoise 60 dB
6	Speech + Noise	Audio book 65 dB + IFnoise 50 dB
7	Music	Piano 65 dB
8	Music	Violine 65 dB
9	Noise	IFnoise 65 dB
10	Noise	Gravel sieving 65 dB

Table 1: List of test signals used for the measurements. The index indicates in which order the signals are presented to the hearing aid, and the loudness is given as sound pressure level (SPL) in decibel.

For each of the four listening situations at least two test signals are chosen (see Table 1). For speech, the International Speech Test Signal (ISTS; Holube *et al.*, 2010) and the German audio book “Abendlied” are used. For speech in noise, both speech signals are mixed with the International Female Noise (IFnoise) with signal-to-noise ratios (SNR) of +5 dB and +15 dB. The IFnoise was generated by using multiple overlapping of the speech material of the ISTS so that it has the same long-term average spectrum as the ISTS (EHIMA, 2016). As test signals for music, a piano and a violin track without any voices are used. Finally, as noise, the IFnoise and an industry noise caused by gravel sieving are considered.

Study design

Two upper class hearing aids of two different manufacturers are analysed with the new method. During the measurement, the ten test signals of Table 1 are presented in a free field, and the output signal is recorded with an ear simulator according to IEC 60318-4. To program the devices, in the fitting software, the “first fit” function is used for a hearing loss of type N3 according to IEC 60118-15. As marker, a reduction of the gain of approx. 20 dB between 1 kHz and 3 kHz is programed. For the comparison of each output signal with the reference, several measures are possible, e.g., simply comparing the overall sound pressure level (SPL). However, it turned out that a more robust and more sensitive method is using the 1/3 octave levels of both signals. Hence, the differences for all 1/3 octave levels between 500 Hz and 8 kHz are computed and then, the root mean square (RMS) of these differences is calculated. This RMS of all 1/3 octave level differences is shortly denoted as Δ with $[\Delta]$ = dB, and used for all results presented in the following. Figures 3 and 4 show the results of hearing aid I and II, respectively. In these figures, the darkness of the pixels represents the values of Δ. Repeatability measurements show that the impact of measurement tolerances on Δ is below 0.4 dB. Therefore, the darkness map begins at 0.5 dB so that all values below 0.5 dB result in white pixels. Moreover, values of Δ between 0.5 dB and 2 dB are coloured by a grey scale to indicate small differences. Values of Δ above 2 dB are represented by a black pixel, since a clear effect of the marker can be recognized.

The focus of this study is on the final result of the classification algorithms, rather than on the transient behaviour. Therefore, both hearing aids have 55 s time to adopt to a test signal, and the output signal between 55 s to 60 s is evaluated only. Furthermore, Figs. 3 and 4 also include the expected classification, which is indicated by crosses. Nevertheless, since no standardized definitions exist for hearing situations, it is not clear what signal-to-noise ratio (SNR) separates speech, speech in noise, and noise. Consequently, the crosses especially for speech, and speech in noise should not be interpreted as correct or ideal classification.

Results

If we have a look at the results of device I and II, we can notice that the same input signal triggers more than one hearing aid program. An explanation for this effect can be that the hearing aid is switching back and forth between two hearing aid programs, or that the signal processing of two hearing aid programs is superposed. A reason for superposition could be that hearing aids do not hardly switch between different hearing situations, but allow for a smooth transition. An analysis of the transient behaviour can give deeper insights, but is not the focus of this work.

If we look at the results for test signals 1 to 6 with speech and speech in noise, we see that both devices mainly detect speech or speech in noise. Device I more often detects speech and not speech in noise, e.g., if we look at the results for test signal “Audio book 65 dB + IFnoise 50 dB”. However, as explained in the foregoing, there is no clear definition of what SNR separates speech and speech in noise, so that both results can be seen as appropriate classification. However, device I classifies “ISTS 65 dB + IFnoise 60 dB”, and device II classifies “ISTS 65 dB” partly as noise. If we assume this effect to be stronger, it might be a problem for the hearing aid user, because speech is processed as noise so that may be the gain for speech is reduced. On the other hand, we see that device I partly detects the IFnoise, which has the same long-term average spectrum as the ISTS, as speech in noise. This could lead to discomfort, because the noise is processes as speech so that the gain for the noise could be elevated. As another peculiarity, device II classifies the test signal “Piano 65 dB” as speech. This is astonishing, since the test signal does not include any voices and no human would classify this track as speech.

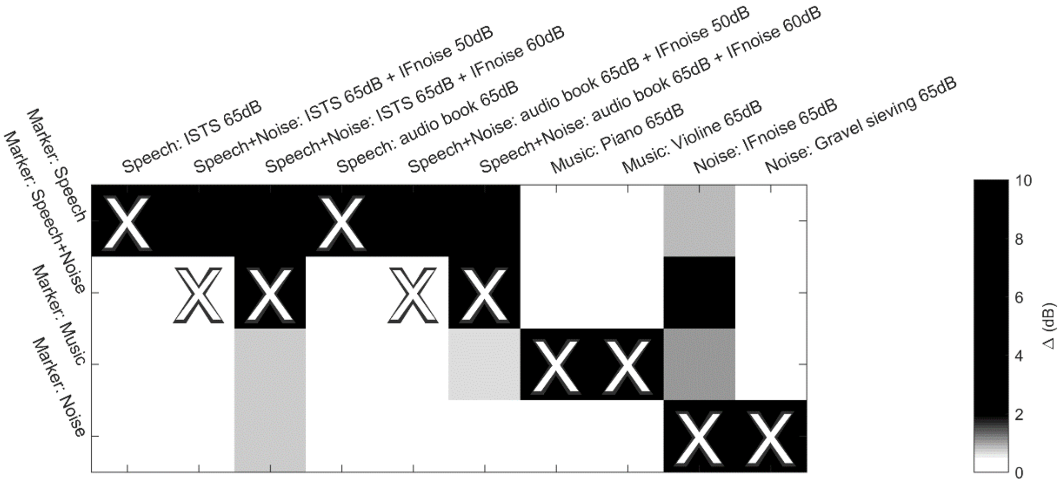

Fig. 3: Measurement results for device I evaluated in the time between 55 s to 60 s. The crosses indicate the expected classification.

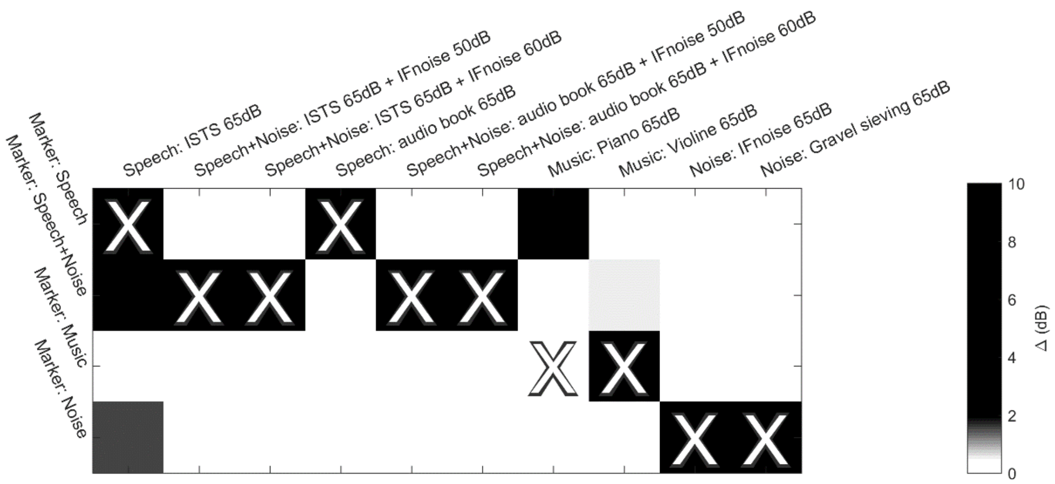

Fig. 4: Measurement results for device II evaluated in the time between 55 s to 60 s. The crosses indicate the expected classification.

COMPARISON WITH RESULTS OF THE DATA LOGGING FEATURE

Most modern hearing aids provide a feature commonly denoted as data logging. This feature shows the hearing aid professional information about the use of the hearing aid so that the fitting can better be adapted to the individual needs of the patient. As one type of information, many hearing aids log the hearing situation experienced by the user. In the fitting software this results is often depicted as relative time data in percentage, e.g., 30 % of the time the user experienced noise, etc.

Exactly these data are used to verify the new method. To this end, one test signal is presented to device II for 1 h, and afterwards, each time the result of the data logging is read out. A long presentation time is necessary, since the data logging does not store signals presented for a few minutes only. Figure 5 depicts the results of the data logging feature in a format similar to the results of Fig.3 and Fig. 4. The only difference is that the colour map represents the relative time in percentage.

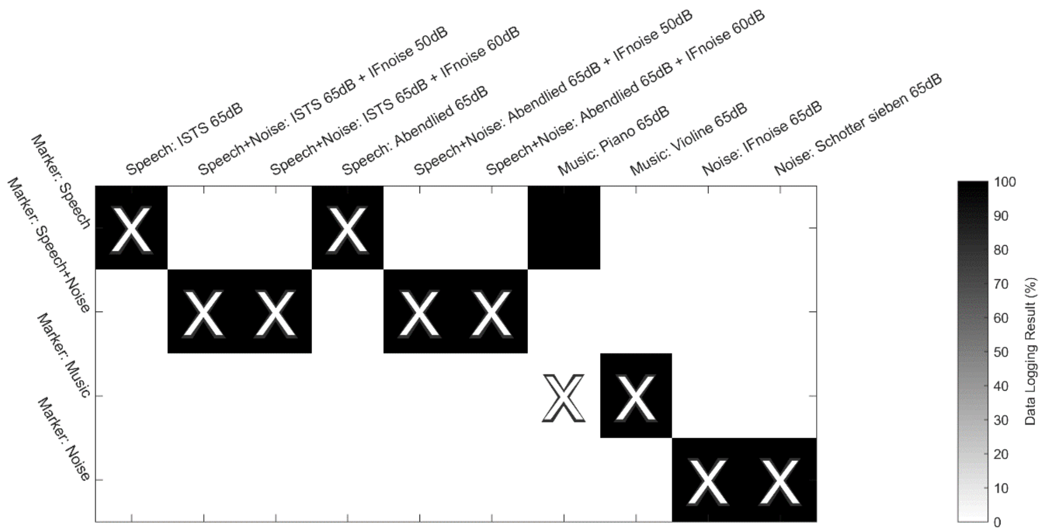

Fig. 5: Results of the data logging feature for device II (see also Fig. 4). One signal is presented for 1 h, and afterwards, each time the result of the data logging is read out.

If we compare the results of Fig. 4 with the results of Fig. 5, we see almost no difference. Only the result for the test signal “ISTS 65 dB” does not completely agree. Both figures show a classification as speech whereas in Fig. 4 the signal is additionally classified as speech in noise, and noise. There are multiple possible reasons for this difference, e.g., the classification has not reached the steady state after 55 s as in Fig. 4 so that the result is different in Fig. 5 where 1h is considered. Another reason could be that the hearing aid switches between multiple hearing situation, but speech in noise and noise is not detected often enough to be stored in the data logging feature.

CONCLUSION

The method presented allows one to analyse what hearing aid program is automatically selected by the hearing aid in a specific hearing situation. This gives insights into the classification strategy and quality among different hearing aid manufacturers – e.g., the results show that the SNR at which speech is separated from speech in noise varies for different manufacturers. Furthermore, there are some signals such as the IFnoise or the piano track, which are easy to classify for humans, but can be difficult to classify for hearing aids.

In addition, a comparison of the results of one hearing aid with results of the data logging feature shows good agreement and verifies the new method. Nevertheless, using the data logging is no reasonable alternative, because entries in the data logging are stored only after a long time (usually > 30 min). Thus, measurements take multiple hours. Moreover, the data logging only shows what hearing situation has been detected, but not if the corresponding signal processing is really active. Finally, another advantage of the new method over the data logging is that also the transient behaviour of the automatic selection of hearing aid programs can be analysed. This is very useful, since not only the reliability but also the time until a new situation has been classified is very important for the hearing aid user. Therefore, this will be subject of future work.

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