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Cognitive Neurodynamics of Listening Effort

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Abstract

In recent years, there has been a rising interest in finding a method to estimate listening effort objectively. The benefit of such a measure would be a support for the hearing aid fitting procedures to reduce the listening effort in hearing impaired persons by an adequate adaption of the personal hearing aids.

In this thesis, two new approaches are presented to extract listening effort correlates using electroencephalographic data. Both methods are based on the distribution of the instantaneous phase. It is hypothesised that for a non effortful listening environment the phase is rather uniformly distributed on the unit circle, compared to a demanding condition. For the latter, it is assumed that the phase is clustered due to an increased auditory attention to the interesting auditory signal.

This concept was firstly tested by using auditory late responses. For this, the wavelet phase synchronisation stability was investigated as a feasible measure of listening effort. To evoke auditory late responses, syllabic paradigms with a different degree of difficulty were generated. Due to the varying task demand, it was expected that the subjects would require a measurable difference in the amount of effort to solve the paradigms. This method was tested in young as well as in middle-aged normal hearing subjects and in people with a different degree of hearing loss. Thus, a possible age and/or hearing loss related effect on the neural correlates of listening effort could be investigated.

The second part of the thesis deals with the extraction of listening effort correlates from the ongoing oscillatory activity. The ongoing EEG activity was investigated with respect to a future hearing aid fitting procedure. Here, in comparison to evoked potentials, the auditory stimulation is not limited to signals of short duration, like tones, syllables or single words. For the analysis of the phase distribution, directional statistics was applied. Additionally, the behavior of the objective listening effort over the measurement time was investigated. This was done to unveil time-varying effects like fatigue, which are not represented in an overall measure.

To test this technique, EEG data from experienced hearing aid users (moderate hearing loss) were recorded, wearing differently configured hearing aids. This part should also give information about the hearing aid settings with reference to a reduction of the subject's listening effort in noisy environments.

It is concluded that listening effort can objectively be estimated by analysing the instantaneous phase of the evoked as well as of the ongoing EEG. With respect to a prospective hearing aid fitting, the proposed method to extract the listening effort from the ongoing EEG activity is more suited as it can be applied in more natural hearing environments. However, the objective estimate of listening effort extracted from the ongoing EEG has to be evaluated by a larger population of subjects.

Zusammenfassung

In den vergangenen Jahren stieg das Interesse an der Entwicklung einer objektiven Methode zur Bestimmung der Höranstrengung. Der Benefit wäre, die Hörgeräteanpassung so zu unterstützen, dass durch eine adäquate Anpassung der Hörsysteme die Höranstrengung Schwerhöriger reduziert wird.

In der vorliegenden Thesis werden zwei Ansätze zur Extraktion von Höranstrengungskorrelaten aus elektroenzephalografischen Daten vorgestellt. Beide Verfahren basieren auf der Verteilung der EEG-Momentanphase. Es wird angenommen, dass für eine nicht anspruchsvolle Hörsituation die Momentanphase auf dem Einheitskreis eher uniform verteilt ist. Für eine anspruchsvolle Hörsituation hingegen wird vermutet, dass aufgrund einer erhöhten auditorischen Aufmerksamkeitslenkung zum interessierenden auditorischen Signal hin, die Phase eher gruppiert vorliegt.

Diese These wurde zuerst mithilfe von auditorisch evozierten Potentialen überprüft. Hierzu wurde die Wavelet-Phasensynchronisationsstabilität untersucht. Durch die Generierung von Silbenparadigmen mit unterschiedlichem Schwierigkeitsgrad wurde erwartet, dass die Probanden auch einen unterschiedlichen messbaren Grad an Höranstrengung zum Lösen der auditorischen Aufgabe aufwenden müssen. Dies wurde in jungen Probanden und Personen mittleren Alters als auch in Personen mit unterschiedlichem Hörverlust getestet. Dadurch sollte ein eventueller alters- oder hörschädigungsbedingter Einfluss auf die Korrelate der Höranstrengung untersucht werden.

Der zweite Teil der Arbeit beschäftigt sich mit der Extraktion von Höranstrengungskorrelaten in der kontinuierlichen EEG-Aktivität. Diese wurde im Hinblick auf eine spätere Anwendung zur Unterstützung von Hörgeräteanpassungen untersucht da hier, im Vergleich zur evozierten Aktivität, die Stimulation nicht begrenzt auf auditorische Kurzzeittestsignale (z.B. Töne, Silben) ist und dadurch eine realere Hörsituation abgebildet werden kann.

Zur Analyse der Phasenverteilung wurde hier der Rayleigh-Test verwendet. Zusätzlich wurde der Verlauf der Höranstrengung über der Messdauer untersucht um zeitlich variierende Effekte (z.B. Ermüdungserscheinungen) zu untersuchen, die nicht in einem umfassenden Maß darstellbar sind.

Um dieses Verfahren zu testen, wurden EEG-Daten von erfahrenen Hörgeräteträgern (mittelgradiger Hörverlust) aufgezeichnet die unterschiedlich konfigurierte Hörgeräte trugen. In diesem Teil sollten auch die einzelnen Hörgeräteeinstellungen bezüglich der Reduktion der Höranstrengung im Störgeräusch untersucht werden.

Es wird gefolgert, dass die Höranstrengung durch Betrachtung der Momentanphase der kontinuierlichen, als auch der evozierten EEG-Aktivität objektiv bestimmt werden kann. In Bezug auf eine spätere Anwendung des Verfahrens zur Unterstützung der Hörgeräteanpassung wird gefolgert, dass aufgrund der besseren Möglichkeiten der auditorischen Stimulation, die kontinuierliche EEG-Aktivität besser geeignet ist. Allerdings muss das objektive Höranstrengungsmaß noch durch Vergrößerung der Probandenpopulation evaluiert werden.

Abbreviations

act	active
AER	auditory evoked response
Ag/AgCl	silver/silver chloride
ALR	auditory late (evoked) response
ANOVA	analysis of variance
BHA	binaurally coupled hearing aid setting
crw	correctly repeated words
CWI	correlation waveform index
CWT	continuous wavelet transform
dB	decibel
DM	directional microphone setting
d'prime	performance accuracy
DSE	directional speech enhancement
DSL[i/o]	desired sensation level [input/output] formula
DSEmed	DSE in a medium setting
DSEoff	DSE turned off
DSEstr	DSE in a strong setting
DSP	difficult syllabic paradigm
ECoG	electrocorticography
EEG	electroencephalography
ERP	event-related potential
ESP	easy syllabic paradigm
f	frequency
fMRI	functional magnetic resonance imaging
G	gain
HL	hearing level

Hz	hertz
ISI	interstimulus interval
ISTS	international speech test signal
k	kilo
L_{AFmax}	continuous sound level with A-weighting filter, fast time constant
L_{Aeq}	equivalent continuous sound level with A-weighting filter
L_{eq}	equivalent continuous sound level
LE	listening effort
LPP	late positive potential
manh	middle-aged normal hearing subjects
MEG	magnetoencephalography
MGB	medial geniculate body
MMGB	medial subnucleus of the medial geniculate body
μ	micro
mild	middle-aged subjects with mild hearing loss
mod	middle-aged subjects with moderate hearing loss
ms	milliseconds
N	noise
n	number of subjects
NAL- R	national acoustic laboratories - revised formula
ODM	omni directional microphone setting
Ω	ohm
OLEosc	objective listening effort (oscillatory)
OLEev	objective listening effort (evoked)
OlSa	Oldenburger sentence test (Oldenburger Satztest)
pas	passive
PC	personal computer
POGO	prescription of gain and maximum output formula
PTA	pure tone average
RT	reaction time
s	seconds
S	signal
SI	speech intelligibility
SNR	signal-to-noise-ratio
SPL	sound pressure level

TRN	thalamic reticular nucleus
V	volt
VMGB	ventral subnucleus of the medial geniculate body
WPSS	wavelet phase synchronisation stability
ynh	young normal hearing subjects

Contents

1. Introduction

1.1. Listening effort and attention

In recent years, the topic "listening effort" attracted more and more attention in the field of rehabilitative audiology. Although a standardised definition of listening effort is not yet available, it can be described as the exertion listeners experience by processing naturally occurring auditory signals in demanding environments (Pichora-Fuller and Singh, 2006; Gosselin and Gagné, 2010; McGarrigle et al., 2014). These processes include speech comprehension, listening to music and localisation of sound sources. This definition can be complemented by looking closely at the first part of the term "listening effort", which gives information about the involved processes. With reference to Kiessling et al. (2003), "listening" can be characterised as the process of hearing with intention and attention. Compared to the pure physiological, passive process of hearing which enables access to the auditory system, listening requires mental effort and the allocation of attentional as well as cognitive resources (Hicks and Tharpe, 2002b; Kiessling et al., 2003; Pichora-Fuller and Singh, 2006; Gosselin and Gagné, 2010; Picou et al., 2013; Hornsby, 2013). Moreover, this goal-directed attentional effort can be considered as support to optimise cognitive processes (Sarter et al., 2006).
Regarding auditory processing in a complex auditory scene, an acoustic target or auditory stream can be selected out of concurrent streams by stimulus salience (bottom-up processes) and focused attention on the target via top-down processes (Shinn-Cunningham and Best, 2008). For example, a single speaker in the presence of others can be singled out in a noisy background either based on the physical attributes of the auditory signal or by the individuals goals, like focusing attention on that single speaker. In these cases the listeners have to exert effort to attend to

relevant and irrelevant sounds to inhibit unwanted sound sources (Dhamani et al., 2013). The described endogenous or controlled processes are, compared to the exogenous automatic processes, limited in capacity, require attention and exert effort (Kahneman, 1973).

1.2. Factors influencing listening effort

Several authors assume that the cognitive effort can be influenced by two factors: First, by signal degradation, which can either result from environmental variances, like an increase of background noises, or from biologic deviations. These deviations include in case of a hearing loss elevated hearing thresholds and a reduced frequency resolution. The second influencing factor relates to age-related variations in cognition (Pichora-Fuller and Singh, 2006; Akeroyd, 2008). In these two cases, the listener has to increase the top-down processes to perceive the attended speech signal correctly.

Regarding the first aspect, people with hearing loss are particularly affected by signal degradation and require much more processing effort to identify sounds compared to those with normal hearing (Downs, 1982; Arlinger, 2003; Shinn-Cunningham and Best, 2008; Kramer, 2008). This increase of cognitive effort is a result from a greater *perceptual filling in* of inaudible fragments to make sense of the auditory information. The *perceptual filling in* includes the use of word, context and spectrotemporal information as well as in the case of phonemic restoration the use of phonetic, semantic and linguistic content of audible fragments (Shinn-Cunningham and Best, 2008). Especially in noisy environments, people with hearing loss have difficulties to shift their attention to the interesting auditory signal to extract the meaning out of the audible speech tokens. Thus, for listeners with hearing loss the endogenous processing is increased. Due to these facts, normal hearing listeners and listeners with hearing loss may have a similar speech understanding, but require a different degree of effort (Edwards, 2007; Shinn-Cunningham and Best, 2008).

Another aspect which can also lead to a diminished speech perception in complex listening situations, is an age-related decline in attention, a slowing of cognitive functions and a decrease in working memory capacity (Pichora-Fuller and Singh, 2006; Desjardins and Doherty, 2013).

Moreover, in his review Arlinger (2003) emphasised and supported by citation of some studies the hypothesis that there is a significant connection between a decrease of cognitive functions and an uncorrected hearing loss. Nevertheless, a positive proof that hearing loss causes a reduction of cognitive functions does still not exist (Arlinger, 2003). However, both are in correlation to each other. On one hand a slowing down of the cognitive functions impedes the uptake of auditory information (Pichora-Fuller and Singh, 2006) and on the other hand hearing loss causes an incomplete uptake of auditory information resulting in a slowing down of cognitive processes (Arlinger, 2003).

The consequences of an increased listening effort are mental fatigue and distress (Edwards, 2007; Kramer, 2008; Mackersie and Cones, 2011; Wascher et al., 2014). Particularly, with respect to the working population, employees with hearing loss have, compared to their normal hearing counterparts, a higher number of sick-leaves due to these stress related health problems (Kramer et al., 2006; Kramer, 2008).

Regarding the determination of listening effort, it is also important to take factors like the individuals motivation to maintain performance during the auditory task into account. Sarter et al. (2006) emphasised that the top-down regularisation of attentional effort is not only a function of the task demand but is likewise dependent on the motivation of the participant. If the subject gets tired due to an increased effort to solve the task, there will be a noticeable decline in attention. However, this effect is less prominent if the participant has a high motivation to continue the task (Boksem et al., 2006). Thus, such effects have likewise to be considered in regards to the development of an objective estimate of listening effort.

1.3. Hearing aid techniques and listening effort

The main objective of hearing aids is to amplify inaudible auditory signals. For this purpose, the hearing aids are fitted traditionally via prescriptive formulas (e.g., NAL-R, POGO, DSL[i/o], half-gain rule) which are based on the individuals hearing loss profile depicted by for instance pure tone audiograms or most comfortable and uncomfortable loudness levels (Dillon, 2001).

Concerning the present hearing aid development, a central topic is the improvement of the signal processing in noisy environments. This stems from the fact that most of the hearing aid users reported a lack of benefit in environmental noise (Tremblay and Miller, 2014).

For such challenging multiple stream environments, the currently available digital hearing aids have mainly two techniques to enhance the interesting auditory signal while suppressing concurrent streams (Shinn-Cunningham and Best, 2008). These speech enhancement and noise reduction techniques are based on noise reduction algorithms and the usage of directional microphone techniques. The noise reduction algorithms basically analyse the spectrotemporal properties of the incoming signals to separate the speech signal from the noise floor followed by a partial suppression of noisy signals (Lunner et al., 2009; Luts et al., 2010). Problems may occur if the noisy signals are non-stationary as in a multitalker conversation (Shinn-Cunningham and Best, 2008). Then, the classification of the signals in speech and noise signals in short time intervals (range of milliseconds) is aggravated, see Lunner et al. (2009). The second technique to filter out unwanted noise is a setting using directional microphones. Here, the cancellation of noisy signals is dependent on the direction where the hearing aid user is turned towards (Shinn-Cunningham and Best, 2008). The microphone has an anterior and a posterior port. The incoming signal of the posterior port is internally delayed by a few milliseconds and then subtracted from the incoming signal at the anterior port. Modifying these delay times leads to different angular directions of the interference cancellation which can be displayed in polar patterns. If the delay time equals the propagation time of the signals, noise signals coming from the rear side will be canceled out maximally (Chung, 2004; Fischer et al., 2012). When the speech signal has a different direction as the noise signals, for example a signal which is coming from the front, it will only be attenuated partially (Puder, 2009). This works well in listening conditions with only one talker, but in multitalker conversations the hearing aid wearer has to switch between the talkers to face the current speaker (Shinn-Cunningham and Best, 2008). Nevertheless, these system reduce the interfering background noises to a specific amount. Thus, it is assumed that these techniques ease the speech understanding in complex environments. As a result, the exerted effort to selectively attend to the interesting stream (Shinn-Cunningham and Best, 2008) and the amount of required cognitive resources should be reduced (Lunner et al., 2009).

In current literature, there are some studies examining the effects of the use of hearing aid on listening or perceived effort (Downs, 1982; Gatehouse and Gordon, 1990; Sarampalis et al., 2009; Ahlstrom et al., 2013; Hornsby, 2013; Picou et al., 2013; Brons et al., 2013; Picou et al., 2014). In these studies, a pure hearing aid amplification effect (Downs, 1982; Gatehouse and Gordon, 1990; Picou et al., 2013;

Ahlstrom et al., 2013), the introduction of additional features like noise reduction algorithms (Bentler et al., 2008; Sarampalis et al., 2009; Brons et al., 2013; Hornsby, 2013), the effects of an automatic microphone setting (Hornsby, 2013) or different directional processing strategies (Picou et al., 2014) were investigated. The quantification of listening effort happened using so called dual-task paradigms or rating scales, see Section 1.4 for a detailed description of these methods. The general outcome of these studies was that due to the amplification of the relevant auditory information, the audibility of the speech signal was improved resulting in a decreased listening effort. With respect to the hearing aid features, only small differences regarding the listening effort were observed. The authors argued that these small differences could be related to limitations in the experimental design (Hornsby, 2013), the sensitivity of the measure (Picou et al., 2014) or to individual differences of the subjects (Brons et al., 2013).

The next step in the development of hearing aid processing schemes should be the design of ”intelligent” digital hearing aids by taking, besides the physical attributes of the incoming signals, also the individuals cognitive abilities like stream segregation, selective attention or the working memory capacity into account (e.g. ”cognition-driven hearing aid signal processing” (Lunner et al. (2009), p. 401); Pichora-Fuller and Singh (2006); Edwards (2007); Pichora-Fuller and Levitt (2012); Tremblay and Miller (2014)). Regarding these aspects, there is also a need for a stronger individualisation of the hearing aid fitting (Kalluri and Humes, 2012) as well as for rehabilitative training (Pichora-Fuller and Singh, 2006; Edwards, 2007; Pichora-Fuller and Levitt, 2012). For example, if a person with hearing loss is able to select a target stream out of a mixture of signals, this person may not need a strong noise reduction setting compared to a person without this ability. Information about these underlying processes to rehabilitate auditory functions cannot be determined by physiological measures like audiograms (Gosselin and Gagné, 2010; Bertoli and Bodmer, 2014). Particularly an estimate of listening effort could give a more complete understanding of the individuals hearing disability (McGarrigle et al., 2014).

1.4. Quantification of listening effort

The next section gives a short overview of the existing subjective and objective methods to estimate listening effort as well as an overview of new approaches for the estimation by electroencephalographic data.

1.4.1. Existing subjective and objective methods

Until now, mainly subjective procedures, like questionnaires (Gatehouse and Noble, 2004; Gatehouse and Akeroyd, 2006; Ahlstrom et al., 2013), rating scales (Humes, 1999) or self-reports are used to estimate listening effort in hearing aid fitting procedures or in studies related to the assessment of listening effort. The subjective procedures give some indication of the individuals' perceived listening effort, but it is still uncertain if the subjective data reflect the real experienced effort (Zekveld et al., 2010). Discrepancies between subjective and objective results indicate that especially older participants tend to overestimate their abilities or their degree of hearing loss. However, differences between self-perceived effort and objective measures of listening effort were also found in children with hearing loss (Hicks and Tharpe, 2002a).
Besides the problem of the internal reference scale of effort, which complicates the comparison of group results, some subjects also confound task performance accuracy/difficulty with the experienced listening effort (McGarrigle et al., 2014). Furthermore, Brons et al. (2013) noticed a ceiling effect of the subjective listening effort scale for positive signal to noise ratios (SNRs). They concluded that a more suitable measure is needed. It should be noted that it is not possible to directly monitor the perceived effort during a listening situation using subjective methods because listening effort can only be determined after the listening task and not during the speech comprehension task (Gosselin and Gagné, 2010; Hornsby, 2013).
Apart from these subjective methods to estimate listening effort, there have also been different approaches to assess this effort objectively. One of the methodologies are dual-task paradigms (e.g. Downs (1982); Rakerd et al. (1996); Sarampalis et al. (2009) and Gosselin and Gagné (2010) for a review), which are based on a limited capacity theory of cognitive resources (Kahneman, 1973). The participants have to perform two competing tasks: a primary listening task and a secondary task which is mostly visual or memory related. The assumption is, that there is a

competition for single limited resources, so that the performance of the secondary task decreases with increasing difficulty level of the primary task. The reduction in the secondary task efficiency serves as a measure of listening effort. However, the described method requires considerable cooperation of the subject and can be influenced by non-listening effort related factors. This is a result from the mutual and crossmodal relationship between the two tasks, which means that the secondary task can also influence the assessment of the primary task (Strauss et al., 2010; Bertoli and Bodmer, 2014). Furthermore, it is difficult to monitor the listening effort over a specific period of time to uncover effects like fatigue. A time dependent decrease of the exerted listening effort can only be determined by (sub-)averaged information gained from a block design (Hornsby, 2013). Thus, the time resolution is dependent on the block size.

The pupil response (Goldwater, 1972; Kramer et al., 1997; Zekveld et al., 2010) and the galvanic skin response (Mackersie and Cones, 2011) were also part of the investigation concerning listening effort. Kahneman (1973) showed in a thought experiment, that the dilation of the pupil is correlated with mental effort. The same outcome was found by Kramer et al. (1997) and Zekveld et al. (2010) in two consecutive studies in the auditory domain. In these studies, sentences embedded in background noise at several signal-to-noise ratios had to be repeated by the subjects. The increasing pupil diameter was interpreted as an increase of listening effort.

Mackersie and Cones (2011) considered different psychophysiological measures, such as heart rate, skin conductance, skin temperature and electromyographic activity as possible indices for listening effort. The psychophysiological data were recorded during a speech recognition task. As the task demand increased, the electromyographic activity as well as the skin response were enhanced. They concluded that "the skin conductance appears to be the most promising measure" for listening effort (Mackersie and Cones (2011), p. 113). Nonetheless, the skin response is not only influenced by mental stressors (Mackersie and Cones, 2011), but also by other factors, like body and ambient temperature (Arunodaya and Taly, 1995; Kucera et al., 2004), additional stimuli like deep breathing, a cough or loud noises (Vetrugno et al., 2003).

However, despite all these different approaches, a widely accepted method to evaluate listening effort objectively in clinical procedures is still not available.

1.4.2. Extraction of listening effort correlates by means of electroencephalographic data

Estimating listening effort by means of electroencephalographic (EEG) data might have some advantages over conventional methods like dual task paradigms, rating scales or questionnaires. The listening effort could be measured directly during the listening task without the need of a secondary task and due to the objectiveness, the measure would be independent of an internal reference scale. Furthermore, if a minimum electrode configuration is used (e.g. 3 electrodes), the EEG can be easily measured. Hence, the method could be applied to support the hearing aid fitting procedures in clinical settings.

In the following sections, two new approaches to estimate listening effort by means of electroencephalographic data will be introduced. One is based on auditory evoked responses, the other one focuses on the analysis of the ongoing oscillatory EEG activities as well as on the resolution of the objective listening effort measure over a specific time period.

1.4.2.1. Auditory evoked responses

An auditory evoked response (AER) or potential can be defined as "activity within the auditory system (the ear, the auditory system or auditory regions of the brain) that is produced or stimulated ("evoked") by sounds [...]."(Hall (1992), p.3). The AER can be classified according to the time of occurrence poststimulus (latency) in early, middle and late evoked responses and can be assigned to specific areas of the auditory system (Hall, 1992). The focus of this part of the work is on the auditory late evoked responses (ALR), which are also known in literature as event-related potentials (ERPs). Here, the main components presumably arise from the auditory cortex (Hall, 1992) and reflect cognitive processes (Fonaryova Key et al., 2005) and the thalamocortical interaction (Strauss et al., 2010; Haab et al., 2011).

In Fig. 1.1 the components of an ALR are illustrated. The potentials are named after the voltage polarity of the responses (recorded from the vertex) and their latency and order of occurrence P1/P100 (positive voltage, 50-80ms), N1/N100 (negative voltage, 100-150ms), P2/P200 (150-200ms) (Hall, 1992). Further components of the auditory late response, which are not illustrated in Fig. 1.1, include the P3 or P300 wave and later components like the N400 wave.

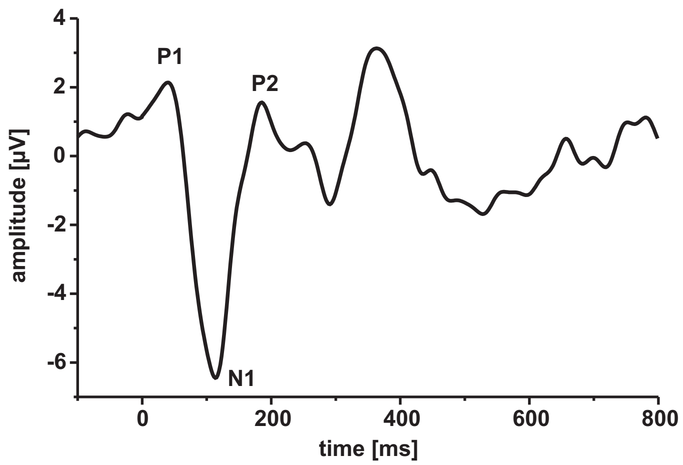

Figure 1.1.: Illustration of an auditory late (evoked) response. The components are named after the voltage polarity of the components (recorded from the vertex) and their appearance poststimulus (P1/P100 (positive voltage, 50-80ms), N1/N100 (negative voltage, 100-150ms), P2/P200 (150-200ms) (Hall, 1992)).

A number of authors investigated the effects of hearing loss and aging on speech/ stimulus processing by measuring auditory evoked responses (Goodin et al. (1978); Oates et al. (2002); Tremblay et al. (2003); Korczak et al. (2005); Martin et al. (2008) and Friedmann (2010) for a review). To date, the main objectives of these studies are not linked to listening or cognitive effort.
Lately, Bertoli and Bodmer (2014) conducted a study related to physiological measures of listening effort. The focus of this study was mainly on the P300 wave, particularly the $P3_2$ or late positive potential (LPP), but also the N1 and N2 components were investigated. The ALRs were stimulated by novelty sounds, like the noise of an animal, while the subjects had to solve an auditory paradigm which was either a tone discrimination task or a speech intelligibility test. The results showed a decrease of the N1 and N2 amplitudes to the novelty sounds as the task difficulty increased. This was interpreted in terms of divided attention on the basis of a limited capacity model of attention. As the task demand increased, the resources available for the novelty sounds were reduced. Furthermore, the LPP was enhanced with increasing task difficulty. The authors interpretation of this result was that the LPP captures the amount of stress of a listener during a specific task.
Mulert et al. (2008) investigated the effects of effort related decision making on the N1 and the P300 wave in a combined EEG-fMRI (functional magnetic resonance

imaging) study during a tone discrimination task. Their results revealed significant effects on the N1 component between the high and low effort condition, but not on the P300 component. Thus, the authors assume a relationship between the mental effort spent by the subject and the N1 potential.
As mentioned in Section 1.1, listening requires the effortful allocation of attentional resources to perceive the presented acoustic information correctly (Hicks and Tharpe, 2002b; Kiessling et al., 2003). In compliance with this fact, the presented method in this thesis to estimate listening effort is based on early stages of selective attention. These stages are endogenously modulated, i.e. they depend on attentional resources and require the aforementioned (see Section 1.1) higher-order cognitive effort (Kahneman, 1973).
In Strauss et al. (2008a, 2010) and Corona-Strauss et al. (2011) a probabilistic model of auditory stream selection was developed. In Fig. 1.2 a sketch of this simplified

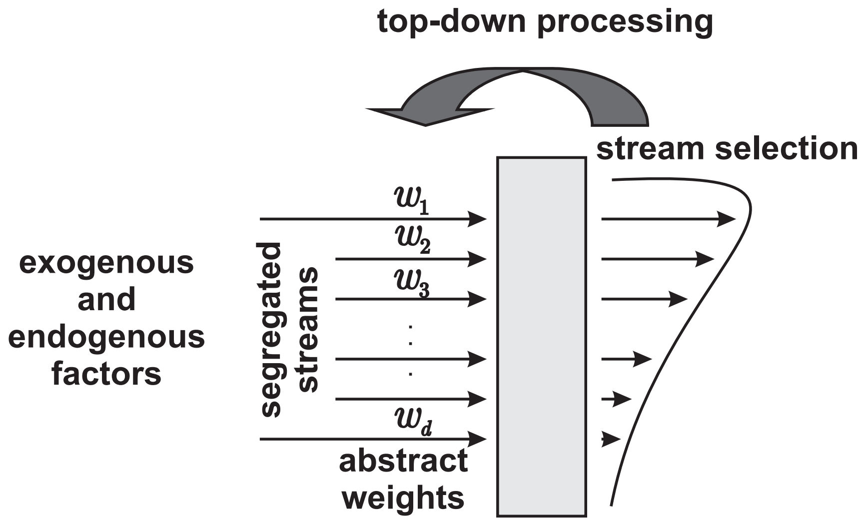

Figure 1.2.: Simplified probabilistic model of the auditory stream selection (adapted from Strauss et al. (2010)).

stream selection model is illustrated. On the left side, the incoming segregated auditory streams, which are influenced by the previously mentioned exogenous and endogenous factors, are depicted. In the model, it is assumed that the probability of an auditory stream to be selected depends on these exogenous and endogenous factors and is represented by the d–dimensional vector $\mathbf{w} = (w_1, w_2, \ldots, w_d)$ in the figure. Here, the vector $\mathbf{w}$ indicates abstract weights of the auditory streams according to their selection probability, which are sorted from highest to lowest probability. With respect to Kiessling et al. (2003) and his definition of listening as process of hearing with intention and attention, it is presumed in this sketch that the listening process is represented by an effortful endogenous modulation of the weights $\mathbf{w}$. Some of these effortful processes, like phonemic restoration, are described in Section 1.2. On the right side of Fig. 1.2, the stream selection is illustrated showing the modulated auditory streams according to their selection probability (from top to bottom).

Moreover, in Strauss et al. (2008a, 2010), this auditory stream selection scheme was coupled with a neurophysical framework of corticothalamic feedback dynamics for the simulation of auditory late responses influenced by exogenous and endogenous effortful corticofugal modulations (see also Trenado et al. (2008) and Appendix A for a detailed description of the model). It is assumed that the abstract weights $\mathbf{w}$ of the auditory streams modulate three corticofugal gains G1, G2 and G3, which can be mapped to different structures of the auditory pathway. This mathematical model, which builds the theoretical basis for the listening effort estimation, predicted auditory attention effects on the instantaneous phase in the time interval of the N1 component (Low et al., 2007; Trenado et al., 2008; Strauss et al., 2008a, 2010).

Regarding auditory evoked potentials, selective attention effects were observed in an endogenous modulation of the N1 component (Hillyard et al., 1973; Hink et al., 1977; Hillyard et al., 1998; Kauramäki et al., 2007; Rao et al., 2010). In these studies, the N1 amplitude was enhanced in attended versus unattended conditions due to an increased attentional effort. Rao et al. (2010) investigated the cognitive effort in interfering sound sources. As the subjects had to classify the pitch of filtered noise, the N1 potential was magnified. This was interpreted as the N1 component was "triggered by greater attentional effort" (Rao et al. (2010), p. 130).

Besides the traditional averaging procedures to analyse the amplitude and latency characteristics of evoked potentials in the time domain, recent approaches investigated the phase characteristics of single-trial responses (Brockhaus-Dumke et al., 2008; Ponjavic-Conte et al., 2012; Bruns, 2004). A measure for the phase locking or synchronisation of single trials can be calculated in the time-frequency space. Different terms for this measure have been proposed: phase locking (value/index), phase synchronisation, phase consistency (Bruns, 2004) or intertrial phase locking (Ponjavic-Conte et al., 2012). Recently, Brockhaus-Dumke et al. (2008) showed in a study related to auditory gating in schizophrenia, that the phase of single-trials in the lower EEG frequency range, namely in the alpha and theta band, provides more

information on auditory information processing as the averaged evoked potentials. Furthermore, they showed that the phase synchronisation in the latency range of the N1 component was reduced in schizophrenia patients compared to the control group.
Ponjavic-Conte et al. (2012) observed attention effects on the instantaneous phase in the time range of the N1 wave. As soon as a concurrent auditory stimulus was introduced, the intertrial phase locking in the theta band was reduced. Regarding the limited capacity model of (auditory) attention (Kahneman, 1973), the distractor consumed attentional resources previously allocated to the target stream.
Kerlin et al. (2010) noticed that auditory selective attention enhances the representation of the auditory speech signal in the auditory cortex within the theta band (4-8 Hz). They assume that selective attention acts like a gain mechanisms on the lower frequencies resulting in an increased auditory cortical activity. Also Klimesch et al. (2007) discussed in a review that the amplitudes of evoked responses reflect frequency characteristics of alpha as well as theta activity.
The relation between the electroencephalographic phase reset due to auditory attention was investigated in Low and Strauss (2009). It is hypothesised that modifying the phase at the alpha-theta border modulates the level of attentiveness and is reflected by a change of the N1 amplitude. In Strauss et al. (2010), it was shown that artificially stabilising the phase of auditory evoked potentials resulted in an increase of the N1 negativity. Thus, it is assumed that analysing the phase of the EEG signals directly by calculating the phase stability reveals a more sensitive correlate of listening effort as the traditional N1 amplitude measures. Furthermore, it was shown in Strauss et al. (2008a, 2010) that the instantaneous phase of the N1 component could serve as an index of the amount of listening effort needed to detect an auditory event, like a target syllable/toneburst. The instantaneous phase was extracted by applying the wavelet transform to the auditory late responses. This way, the wavelet phase synchronisation stability (WPSS, see Section 2.1.5.2 for details) could be calculated. It is assumed that for a fixed frequency, a higher synchronisation of the phase reflects a higher cognitive effort to solve an auditory task (Strauss et al., 2008a, 2010).
For the first part of this thesis, namely the investigation of the extraction of listening effort correlates by means of auditory evoked potentials, a pilot test was conducted (Bernarding et al., 2010a). There, two syllabic stimulation paradigms with a different level of difficulty were tested in young normal hearing subjects (a

total of 21 young subjects; 12 male/9 female, aged 20 to 34 years, mean age 25±3.52 years). The stimulation paradigms were termed as difficult paradigm (DSPnoise) and easy paradigm (ESPnoise). In the difficult paradigm, the syllables had the same vowel; in the easy paradigm, different vowels were used. Thus, it was expected that the differentiation of the syllables in the DSPnoise requires more effort from the subjects. Additionally, the stimuli were embedded in multitalker babble noise at a SNR of +5dB.

Fig. 1.3 shows the result of the WPSS together with the mean value of the WPSS in the range of the N1 wave which was termed as objective listening effort (evoked) $OLEev$, see Section 2.1.5.2 for details. In the upper figure on the left, the grand

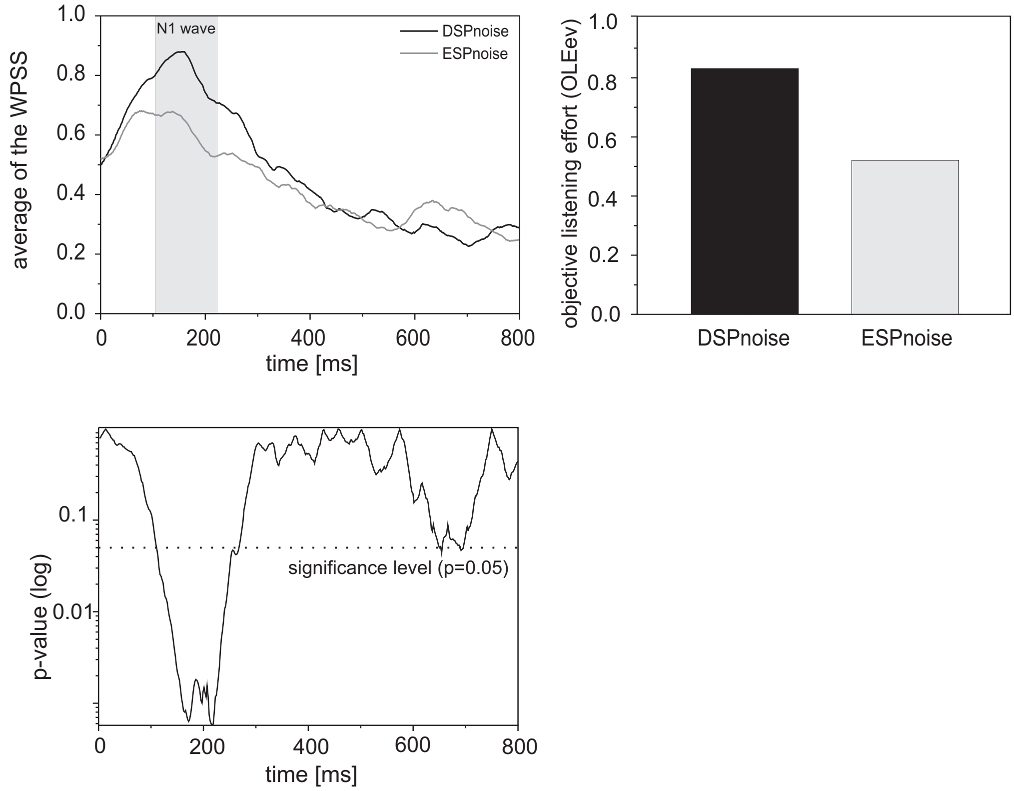

Figure 1.3.: Results of a pilot test. Left (upper figure): Grand normalised average of the WPSS (over 18 subjects) for the two paradigms. Left (lower figure):Results for the time resolved ANOVA (DSPnoise vs. ESPnoise) in logarithmic scale (DSPnoise and ESPnoise) embedded in babble noise (SNR: +5dB). Right: LE-level (mean WPSS in the range of the N1 wave).

normalised average of the WPSS (over 18 included young subjects, scale $a = 40$, f=6.4 Hz, EEG theta band) for the DSPnoise and the ESPnoise paradigm is illustrated. The figure below shows the result of the time-resolved one-way analysis

of variance (ANOVA). On the right side, the corresponding *OLEev*, which is the mean of the WPSS in the range of the N1 wave, is depicted. It is noticeable, that the WPSS and the *OLEev* are much larger for the paradigm with the higher task demand (DSPnoise) as it is for the paradigm with the lower task demand (ESP-noise). Furthermore, a significant difference is evident between the two tests in the time interval of the N1-P2 complex.
The first part of this work deals with the extraction of listening effort correlates using the previously proposed estimate of it, namely the phase synchronisation stability. In this part it was particularly examined if the WPSS reflects the listening effort in young normal hearing subjects and in middle-aged persons with a normal hearing sensitivity as well as two different degrees of hearing loss (mild and moderate hearing loss). Furthermore, behavioral data, namely reaction time (RT) and performance accuracy (d'prime), were taken into account to complement the electrophysiological data.

1.4.2.2. Ongoing oscillatory EEG activity

There are some limitations in the study of auditory evoked responses regarding the design of stimulation paradigms. A central issue is the limitation of the auditory stimulation to signals of short duration, like tone bursts, syllables and short words (Hall (2007), pp. 490ff.). Therefore, the AERs cannot be analysed during longer listening periods, for instance during a speech intelligibility test. The AERs are not only dependent on endogenous, but also on exogenous effects, in other words, they are influenced by physical stimulus characteristics (Hall, 2007). Thus, the exogenous effects have to be minimised, which constrains the comparability of the obtained results. This means that the different noise types, SNRs or hearing aid settings, which always modify the incoming auditory signal, cannot be compared directly to each other.
To overcome these problems, the second part of this work deals with the ongoing oscillatory activity. Here, the EEG can be analysed during longer listening periods. Thus, the listening effort could be extracted by using noise embedded sentences or during a sentence recognition test. These degrees of freedom in the design of the auditory stimulation are essential requirements for a possible prospective EEG-aided hearing aid adjustment in clinical settings.
A widely debated topic in the EEG (Kerlin et al., 2010; Ng et al., 2012), electro-

corticographic (ECoG) (Zion Golumbic et al., 2013; Mesgarani and Chang, 2012) and magnetoencephalographic (MEG) (Peelle et al., 2013; Ding and Simon, 2012) research is the phase entrainment of cortical oscillations. In literature two main hypotheses regarding the functional role of cortical entrainment are under discussion: (1) The cortical entrainment emerges due to physical characteristics of the external stimuli; (2) the phase locking is a modulatory effect on the cortical response triggered by top-down cognitive functions (Ding and Simon, 2014).

The first theory is supported due to the effect that theta oscillations in the auditory cortex entrain to the envelope of sound (Ng et al., 2012; Kerlin et al., 2010; Weisz and Obleser, 2014). This low-frequency activity can be seen as a reflection of the fluctuations of the speech envelope (Zion Golumbic et al., 2013). It is discussed that such an entrainment to the speech rhythm increases the sensitivity to acoustic information and supports the processing of speech (Peelle et al., 2013).

The second aspect deals with a modulatory effect on the phase via top-down processes. Here, the synchronisation of the phase in auditory processing regions acts like a mechanism of attentional selection (Peelle et al., 2013). Moreover, Zion Golumbic et al. (2013) mentioned a "selective entrainment hypothesis" (p. 981), which states that an endogenous attentional modulation causes the neuronal oscillators to phase-lock to the attended speech stream in order to form an internal representation of this stream while unattended streams are excluded. It is hypothesised that this effect arises from attentional sampling, where hierarchically higher regions modulate the neuronal activity so that attended stimuli fall into an optimal phase of an oscillation (Weisz and Obleser, 2014).

This theory of an attentional modulation of the neural oscillations is supported by studies in the auditory as well es in the visual domain. Recently, Kerlin et al. (2010) analysed if selective attention to speech acts like a gain control. The results showed an attentional enhancement of the cortical activity in the area of the auditory cortex at lower frequencies (4-8 Hz). In a MEG study, Peelle et al. (2013) investigated the effects of a possible internal modulation using noise vocoded sentences. They found, besides the phase-locked effect to the speech envelope (slow amplitude modulations, approx. 4-7 Hz), an additional top-down modulatory effect on the cortical response which contributes to the speech comprehension process. There were similar findings in the visual domain. For instance, Busch and Van Rullen (2010) ascertained that the detection performance of a target was highly correlated with the phase of ongoing oscillations around 7 Hz.

Regarding such a possible attentional (effortful) modulation of the neural responses via phase locking or synchronisation, it can be concluded that the proposed method for the extraction of listening effort correlates relies on the instantaneous phase information of the ongoing electroencephalographic activity. The presented method to analyse correlates of listening effort is based on the distribution of the instantaneous phase of the EEG signals. The hypothesis is that for a non effortful listening environment the phase is more uniformly distributed on the unit circle than for a demanding condition. For the latter, it is assumed that the phase is clustered on the unit circle due to an endogenous effortful modulation caused by an increased auditory attention to the interesting auditory signal. To prove if the phase is uniformly distributed around the unit circle or if it departs from uniformity, the Rayleigh Test was applied to the instantaneous phase of the EEG. This test was also used by Babiloni et al. (2002) to investigate the significance of (non) phase-locked components in EEG single-trials after electrical stimulation.

To extract the possible oscillatory listening effort correlates, two studies were designed. The objectives of the first study were to identify such correlates and to investigate the effects of different hearing aid settings on the listening effort and its correlates, respectively. These hearing aid settings included a directional microphone configuration combined with different levels of a noise reduction algorithm as well as a setting using omnidirectional microphones. For this, two listening conditions were generated to imitate an effort demanding environment. In one condition, the speech material consisted of sentences embedded in background noise taken from a German sentence test (Oldenburger sentence test, Wagener et al. (1999)) and the task of the subject was to repeat the sentence directly after the tone signalisation. In the other condition, an audiobook which was also embedded in background noises was presented. Here, the subjects had to answer easy short story related questions. In the second study, different hearing aid settings were also tested. Here, a directional microphone configuration and a new binaurally coupled hearing aid technique were compared to a fitting, using omnidirectional microphones. Again, the sentences from the German sentence test were taken as test material and the background noises were increased by adding additional noises and sound sources. Due to these aggravated listening conditions, the differences concerning the listening effort were minor. Thus, it was necessary to obtain a temporal resolution of the objective listening effort to unveil underlying processes such as fatigue effects or a cessation to spend attentional effort (Bernarding et al., 2015a).

1.5. Contribution of this work

The main contribution of this work was the development of an objective method to estimate listening effort by means of electroencephalographic activity for a prospective application of this measure in hearing aid fitting procedures. For this, two new approaches were introduced which are based on the phase synchronisation effects of the electroencephalographic data. Both approaches were tested in subjects with hearing loss.

The first method relies on the analysis of auditory evoked responses. The listening effort was assessed by the investigation of the inter sweep wavelet phase synchronisation stability as measure for phase locking effects of the evoked electroencephalographic activity. For this purpose two specific auditory stimulation paradigms were designed, which exert a different degree of listening effort from the subjects while minimising the exogenous influences on the evoked responses.

Another aspect of this part was to test the proposed measure in young and middle-aged subjects, in normal hearing subjects as well as in listeners with a mild and a moderate hearing loss.

To develop the final study design, some preliminary studies were conducted. These studies were necessary to fix the particular parameters of the stimulation paradigms. Here, different noise types and noise onsets (Corona-Strauss et al., 2011), and syllabic stimuli (Bernarding et al., 2010a), were tested. Furthermore, non-listening effort related factors on the wavelet phase synchronisation stability in time intervals of the N1 and the P3 waves were investigated (Bernarding et al., 2010b). Additionally, some preparatory work was done to test the proposed measure in young and middle-aged subjects as well as in listeners with normal hearing and with hearing loss. In a first approximation to test the proposed measure in hearing impaired subjects, a hearing loss and a simple amplification scheme of a hearing aid were simulated (Bernarding et al., 2011b). After this, the new approach to measure listening effort using the phase synchronisation stability was tested in small subject populations investigating the effects of age (Bernarding et al., 2011a), and the effects of hearing loss (Bernarding et al., 2011c).

For this purpose, young and middle-aged subjects as well as subjects with mild to moderate hearing loss participated in these studies. In this thesis, the final study is presented whose design was influenced by the outcomes of the aforementioned preliminary work (Bernarding et al., 2013b). It was shown that the phase syn-

chronisation stability could be used to assess objective listening effort correlates. Furthermore, it was shown in this part that, due to the fact that exogenous variables were kept to a minimum, this phase clustering was solely effected by an attentional, effortful endogenous modulation.
The second approach focuses on the phase characteristics of the ongoing oscillatory activity as a prospective objective marker for listening effort. The ongoing EEG activity was taken into account as the evoked activity has some limitations regarding the design of the auditory stimulation. These limitations are related to the strong dependency of the evoked potentials on exogenous factors or due to limitations of the stimuli to auditory signals of short duration. Thus, specific stimulation paradigms to test persons with hearing loss and to investigate the effects of differently configured hearing aids could be generated. In a first step, the concept of entropy was investigated to describe the distribution of the instantaneous phase (Bernarding et al., 2012, 2013a). These results were encouraging, because listening situations which require a different amount of attentional effort could be distinguished objectively. However, the entropy differences between the listening situations were small, more precisely in the range of 10^{-4}. Thus, a more appropriate test to determine the distribution of the instantaneous phase of the EEG data was needed.
In a next step, the Rayleigh-test, taken from the field of circular statistics, was applied to investigate if a uniform or a non-uniform phase distribution dominates. The outcome of this Rayleigh-test was examined as an objective measure of listening effort and was evaluated by using traditional subjective rating scales. A further issue of this part was the investigation of different hearing aid configurations. Here, the main objective was finding a hearing aid setting which reduces the listening effort in specific listening situations. For this, two studies were conducted: In study 1, the focus of the hearing aid settings was on speech enhancement and noise suppression algorithms. Preliminary results were published in Bernarding et al. (2014) and the complete findings will be formulated in Bernarding et al. (2015b). In study 2, a traditional directional microphone setting was compared to a new binaurally coupled hearing aid technique and a fitting using omnidirectional microphones. In the second study, due to aggravated listening conditions, the differences of the listening effort between the hearing aid settings were minor. Thus, it was necessary to obtain a temporal resolution of the objective listening effort to unveil underlying processes such as fatigue effects or a ceasing to spend attentional effort. For this purpose, a

time-resolved form of the objective listening effort estimate was developed.

Organisation of the work: This thesis is organised as follows: In the next chapter, Chapter 2, the design and generation of the specific auditory stimulation paradigms together with the calibration procedure is explained. Furthermore, details about the measurement setup as well as the data acquisition and the included subjects are provided. In a separate section of this chapter, the two methods for the extraction of the neural correlates of listening effort are explained precisely. This information is given separately for the two examined methods, for the study of listening effort correlates extracted from the auditory evoked activity as well as for the ongoing oscillatory activity. In Chapter 3, the results of these two approaches are presented. Here, the behavioral results together with the phase synchronisation measures as objective listening effort measures are shown for each subject group. This chapter is followed by a detailed discussion of the reported findings as well as information about future works. Finally, in the last chapter, conclusions are given.

2. Materials and Methods

In this chapter, information is provided about the experimental design including the construction of the auditory paradigms, the selection of appropriate auditory stimuli, their calibration and the task of the participants. Subsequently, details are given about the participants and the measurement setup to collect the EEG data. Concerning the data processing, the pre-processing of the EEG data, the extraction of the instantaneous phase by the wavelet transform and the analysis of the phase distribution is described.

As mentioned in the previous chapter, two approaches to extract listening effort correlates were investigated. One is based on the evoked EEG activity, the other one focuses on the analysis of the ongoing oscillatory EEG activity. Thus, the aforementioned information is given separately: Part 1 deals with the evoked activity and part 2 describes the methods used to analyse the ongoing oscillatory activity. In part 2, also details about the hearing aid fitting procedures are provided.

2.1. Part 1: Auditory evoked potentials

2.1.1. Stimulus materials

The consonant-vowel syllables /ba/, /da/, /pa/, /de/ and /bi/ were used as auditory stimuli. The consonants /b/ and /d/ were chosen because of their different place of articulation. Thus, they are difficult to discriminate for persons with hearing loss (Oates et al., 2002). Also the bilabials /p/ and /b/ are difficult to distinguish, based on their different voice onset times (Tremblay et al., 2003). Due to the combination of the syllables it was expected that the discrimination of them

should yield a difficult task.
Each syllable was spoken by a female person and digitally recorded with a sampling frequency of 16 kHz. After the collection of the syllables, the stimuli were shortened to a total duration of 200 ms and their amplitudes were normalised. To minimise a sudden onset of the syllables, a window was applied. This window was build by rise and fall times, which were the first and second halves of a Gaussian window (duration: 50 ms), and a plateau part (duration: 150 ms) with a flat amplitude of 1.

Calibration of the auditory stimuli
The syllables were calibrated according to the norms (International Organization for Standardization, 2009). For this, a hand-held sound level metre (type 2250, Brüel & Kjær, Denmark) connected to a pre-polarised free field 1/2" microphone (type 4189, Brüel & Kjær, Denmark) was used. The microphone was connected to an artificial ear (type 4153, Brüel & Kjær) which was coupled to circumaural headphones (HDA200, Sennheiser, Germany). To assess the fluctuating noise levels of the speech material, the "equivalent continuous sound level" (L_{eq}) was selected (Brüel & Kjær, 2013; International Organization for Standardization, 2009). Furthermore, for the determination of the speech level, defined as L_{eq} with frequency weighting C, silent intervals were removed (International Organization for Standardization, 2009).

2.1.2. Experimental design

To test objective differences regarding the exerted listening effort, two paradigms with a different degree of difficulty were constructed:

The *Difficult Syllabic Paradigm (DSP)* consisted of the syllables /pa/, /da/ and /ba/. This paradigm was expected to be difficult to solve because the syllables have the same vowel /a/ and the consonants are hard to distinguish (see Section 2.1.1).
The *Easy Syllabic Paradigm (ESP)* was composed of the syllables /pa/, /de/ and /bi/. This paradigm was anticipated to be easier to solve, because different vowels were presented.

Both paradigms had the same target syllable /pa/, which had to be detected by the subjects. Furthermore, only ALR sequences evoked by the target syllable were analysed. Thus, it was assumed that the exogenous effects on the ALR components were minimised as these evoked potentials are also modulated by physical attributes of the stimuli, like intensity or frequency, see Hall (1992). In both paradigms, each syllable was randomly presented 100 times using a randomised interstimulus interval ranging from 1-2 s. The randomised order was used to maximise the entropy of the experiment so that the solution requires an effortful task. The subjects were instructed to pay attention to the stimulus and to press a button each time they heard the target syllable. Furthermore, the subjects were asked to keep their eyes closed and to minimise movements to avoid muscular artifacts. The auditory stimuli were monaurally presented over headphones (HDA200, Sennheiser, Germany) at a fixed intensity of 65 dB sound pressure level (SPL) to the subjects with normal hearing sensitivity and mild hearing loss.

A loudness level adjustment was performed for the subjects with moderate hearing loss to ensure an equal audibility of the syllables. This loudness adjustment was performed in two steps: First, a recorded sentence (*In German: Bitte gleichen Sie die Lautstärke von [...] an die des Satzes an. In English: Please adjust the loudness of [...] to the loudness of the sentence.*) was presented and adjusted to a comfortable loudness level by the subject. In the second step, the gap *[...]* was filled with each syllable (/ba/, /da/, /pa/, /de/, /bi/). Then, the loudness of each syllable was gradually adjusted to the loudness of the sentence. After this, a short sequence of each paradigm was generated and presented to every subject to guarantee that each syllable could be perceived and distinguished. Note that the target syllable had always the same intensity in both testing conditions. The selected intensities of the loudness adjustment for each syllable are shown in Tab. 2.1.

Table 2.1.: Mean and standard deviation values of the individually adjusted intensities for each syllable for the 19 included subjects with moderate hearing impairment.

Syllable	**/ba/**	**/da/**	**/pa/**	**/de/**	**/bi/**
Intensity [dB SPL]	74.4±4.5	74±3.4	73.8±4.1	74.7±5.1	75.6±4.5

The experiment lasted 30 minutes and a short break was made between each paradigm to avoid fatigue. After a detailed explanation of the procedure, all subjects signed a consent form. All measurements were conducted in accordance to the Declaration of Helsinki.

2.1.3. Participants and inclusion criteria

Participants

A total of 94 subjects participated in this study:

a) 24 young subjects (13 male/11 female) with normal hearing levels (ynh; aged 21 to 35 years, mean age: 25.25±4.01 years),

b) 22 middle-aged subjects (9 male/13 female) with normal hearing levels (manh; aged 40 to 60 years, mean age: 51.15±5.64 years),

c) 25 middle-aged subjects (15 male/10 female) with mild hearing loss (mild; aged 45 to 63 years, mean age: 51.87±5.82 years) and

d) 24 middle-aged subjects (9 male/15 female) with moderate hearing loss (mod; aged 43 to 57 years, mean age: 51.12±5.53 years).

The young subjects were student volunteers from the Saarland University and Saarland University of Applied Sciences, the middle-aged subjects with normal hearing sensitivity were relatives or acquaintances of them, while the subjects with hearing loss were volunteers from the rehabilitation centre MediClin Bosenberg.

Inclusion criteria

For an objective waveform analysis, a correlation waveform index (CWI) was defined. The CWI was used to support objectively the visual detection of a clearly identifiable waveform of the N1-P2 complex. For this, the EEG responses to each stimulus, namely the ALR single sweeps, were separated into two matrices, containing either all odd or all even sweeps. Then, the matrices were separately averaged and the resulting vectors x (average of the even sweeps) and y (average of the odd sweeps) were saved. After this, the correlation coefficient $\rho \in [-1, 1]$ between the resulting waveforms x and y was calculated (Duda et al., 2001).

$$\rho := \frac{\sum_{i=1}^{N}(x_i - \overline{x})(y_i - \overline{y})}{\sqrt{\sum_{i=1}^{N}(x_i - \overline{x})^2}\sqrt{\sum_{i=1}^{N}(y_i - \overline{y})^2}}, \tag{2.1.1}$$

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 achieved in our studies by comparing different ALR waveforms with the corresponding correlation coefficients. A value $\geq$ 0.4, in the range of the N1 wave (50-200 ms), corresponds to a clearly identifiable ALR waveform according to a visual analysis by an expert.

The subjects inclusion criteria for the study were: (i) they had an identifiable waveform of the N1-P2 complex in the ALRs and fulfilled the CWI, and (ii) they detected correctly at least 80 % of the target syllables, which served as a control of the cooperation of the subject.

Included subjects

Due to these inclusion criteria, some subjects had to be excluded. Finally, each group consisted of the following number of tested subjects:

a) 20 included subjects (ynh): four subjects were excluded due to criterion (i),

b) 20 included subjects (manh): two subjects were excluded due to criterion (i),

c) 20 included subjects (mild): five subjects were excluded due to criterion (i),

d) 19 included subjects (mod): three subjects were excluded due to criterion (i) and two subjects were excluded due to criterion (ii).

The degree of hearing impairment was defined according to the definition of the European Commission as the pure tone average (PTA^{4f}) of the hearing threshold at 0.5 kHz, 1 kHz, 2 kHz, 4 kHz (Shield, 2009). According to this definition, a PTA^{4f} of 20 dB hearing level (HL) or less is classified as normal hearing sensitivity, a mild hearing loss is defined with thresholds in the range of 21 to 39 dB HL and a moderate hearing loss is defined as a PTA^{4f} from 40 to 69 dB HL (Shield, 2009). Furthermore, most of the subjects with hearing loss had a high-frequency or sloping hearing loss.
Fig. 2.1 depicts the mean pure tone audiograms (top) and the corresponding standard deviations (bottom) of the included subjects for the different groups.

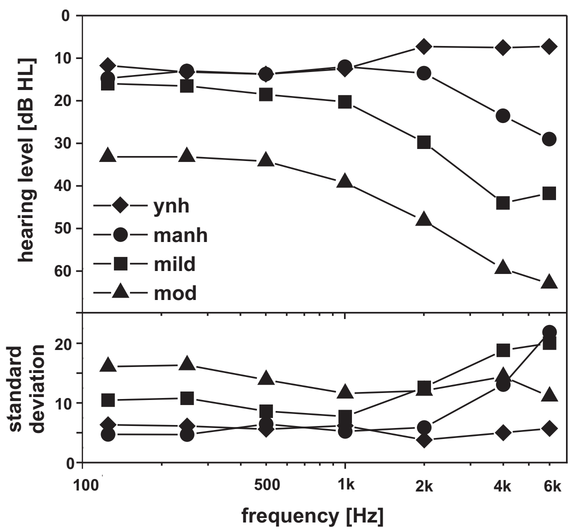

Figure 2.1.: Mean pure tone audiograms (top) and corresponding standard deviations (bottom) for each subject group (ynh=young normal hearing subjects, manh=middle-aged normal hearing subjects, mild=middle-aged subjects with mild hearing loss, mod=middle-aged subjects with moderate hearing loss).

2.1.4. Measurement setup and data acquisition

Fig. 2.2 illustrates the measurement setup for the collection of the electroencephalographic data together with the components for the auditory stimulation. The auditory stimulation and the trigger signal are generated on the personal computer 1 (PC1). The PC1 controls also the programmable attenuator to regulate the intensity levels of the auditory stimulation. The auditory stimulation is monaurally presented via headphones (HDA200, Sennheiser, Germany) to the right or to the left ear of the subject. The trigger signal was transferred to a multimodal trigger conditioner box (g.Trigbox, g.tec, Austria) which modified the voltage level of the incoming signal. Thus, the trigger signal could be acquired together with the electroencaphalographic activity by the biosignal amplifier (g.USBamp, g.tec, Austria). The biosignal amplifier was connected to and controlled by the personal computer 2 (PC2).

The continuous EEG was recorded by the biosignal amplifier using a sampling frequency of 512 Hz (bandpassfilter: 1-30 Hz). The Ag/AgCl-electrodes (Schwarzer GmbH, Germany) were attached as follows: ipsilateral to the stimulus at the right or left mastoid (A1/A2), common reference at the vertex (Cz) and ground at the upper forehead (Fpz). Electrodes impedances were kept below 10 kΩ in all measurements.

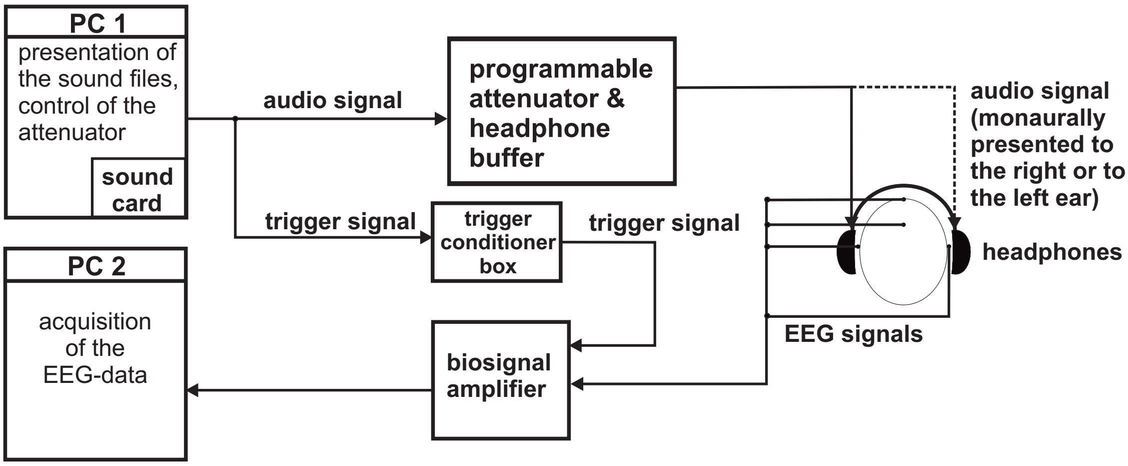

Figure 2.2.: Measurement setup for the collection of the EEG data together with the components for the auditory stimulation. On PC1 (i) the auditory stimulation and the trigger signal are generated; (ii) the programmable attenuator is controlled in order to regulate the intensity levels of the auditory stimulation. The auditory stimulation is presented to the subjects via headphones. The voltage level of the trigger signal is adapted by the trigger conditioner box. The PC2 acquires the EEG data and the trigger signal coming from the biosignal amplifier.

Artifacts were rejected by an amplitude threshold of 50 μV. The ALR single sweeps were segmented from the continuous electroencephalographic data according to the trigger signal in a time window of 900 ms starting 100 ms prestimulus. The data acquisition and the post processing of the EEG data was realised using software for technical computing (MATLAB and Simulink, MathWorks Inc., USA).

2.1.5. Data analysis

2.1.5.1. Behavioral measures

To complement the electrophysiological data, the behavioral data, namely the reaction time (RT) and the performance accuracy (d'prime), proposed by the theory of signal detection (Yanz, 1984), were taken into account. These two measures were also applied in other event-related potential studies (e.g. Oates et al. (2002); Korczak et al. (2005); Martin et al. (1999)) to support the gained results. The median reaction time can be interpreted as a possible index for the processing of the incoming stimuli (Gatehouse and Gordon, 1990), whereas the d' prime reflects the discrimination ability and the perception of information. Opposed to percent correct scores, also the false alarm rate of the subjects response is included (Yanz, 1984). Note that two different subjects could have an equal performance, but a different amount of listening effort to understand speech (Edwards, 2007). The performance accuracy can be calculated as $d' = z(false\ alarm\ rate) - z(hit\ rate)$ (Yanz, 1984; Stanislaw and Todorov, 1999), where z is the standard score. The hit rate is defined as the probability to detect a target syllable correctly and the false alarm rate corresponds to the probability to respond to a distractor.

2.1.5.2. Wavelet phase synchronisation stability and listening effort

In the following section, a short description of the wavelet transform (see Daubechies (1992); Aguiar and Soares (2011) for details) and the calculation of the phase synchronisation stability is provided.

Continuous wavelet transform
Let us consider a given function $\psi(\cdot) \in L^2(\mathbb{R})$, where $L^2(\mathbb{R})$ denotes a set of square integrable functions, i.e. satisfying $\int_{-\infty}^{\infty} |x(t)|^2 dt < \infty$. Note that due to the norm of x, which represents the energy of x and is defined as $||x||^2 = \int_{-\infty}^{\infty} |x(t)|^2 dt < \infty$, the space $L^2(\mathbb{R})$ can also be considered as a "space of finite energy functions" (Aguiar and Soares (2011), p.6).
The *admissibility condition*, which states the minimum prerequisite of a function to be used as *(mother) wavelet*, is given by

$$0 < \int_{\mathbb{R}} \frac{|\Psi(\omega)|^2}{|\omega|} \mathrm{d}\omega < \infty, \tag{2.1.2}$$

where $\Psi(\omega)$ is the Fourier transform of the wavelet. This condition involves that $\Psi(0) = \int_{\mathbb{R}} \psi(t)dt = 0$, which means that the wavelet oscillates around the time axis and has zero mean. A family of wavelet functions can be obtained by translation and dilation (scaling) of the the mother wavelet ψ:

$$\psi_{a,b}(\cdot) = |a|^{-1/2} \psi \left(\frac{\cdot - b}{a} \right), \tag{2.1.3}$$

where $a, b \in \mathbb{R}$, $a \neq 0$. The scale a regulates the width of the wavelet (stretching or compression), whereas the translation factor b shifts the wavelet in time, specifically, it controls the location (see Aguiar and Soares (2011)).

The continuous wavelet transform, which maps a signal $x(t) \in L^2(\mathbb{R})$ in the time-scale or time-frequency domain, respectively, can be given by the inner L^2-product

$$(\mathcal{W}_\psi x)(a,b) = \langle x, \psi_{a,b} \rangle_{L^2} = \int_{\mathbb{R}} x(t)\psi^*_{a,b}(t)dt, \tag{2.1.4}$$

where $*$ denotes a complex conjugation.

(Wavelet) Phase synchronisation stability

For the analysis of single sweep sequences, the wavelet phase synchronisation stability (WPSS), that was introduced in Strauss et al. (2005, 2008b) to assess attention correlates in ALR sequences, was used. The stability of the instantaneous phase, which solely regards the phasing of the sweeps (Bruns, 2004), can be applied to monitor auditory attention (Low et al., 2007).

The synchronisation stability $\Gamma_{a,b}$ of a sequence $\mathcal{X} = \{x_m \in L^2(\mathbb{R}) : m = 1, \ldots, M\}$ of M ALR sweeps can be defined by

$$\Gamma_{a,b}(\mathcal{X}) := \frac{1}{M} \left| \sum_{m=1}^{M} e^{\imath \arg\left((\mathcal{W}_\psi x_m)(a,b)\right)} \right|. \tag{2.1.5}$$

Note that (2.1.5) yields a value in $[0, 1]$. A value equal to one means that phases are perfectly coherent, whereas a value close to zero represents a more non-phase locked activity. In literature, the phase synchronisation stability is also known as intertrial phase-locking value or index (Ponjavic-Conte et al., 2012), as phase consistency (Bruns, 2004) or as mean resultant vector (Mardia and Jupp, 2000).

In Strauss et al. (2008a, 2010) it was shown that the WPSS in the time interval of the N1 wave, which is modulated by auditory attention (Hillyard et al., 1973), can serve as a possible objective measure for listening effort. It is assumed that a higher synchronisation stability (a value close to 1) represents an augmented cognitive effort, resulting from an increased attention to solve an auditory task.

For a fixed a and b and a suitable experimental paradigm, it was suggested in Strauss et al. (2008a) that

$$\text{objective listening effort} \propto \Gamma_{a,b}(\mathcal{X}) \tag{2.1.6}$$

in the time interval of the N1 wave.

To compare different listening conditions, the objective listening effort estimate $OLEev$ was defined. This measure, calculated for a specific scale and wavelet type,

represents the mean of the WPSS in the time range of the N1 interval. The limits of the N1 interval were taken from the minimum and maximum latency values of all measured subjects. This means that in this study, the interval boundaries were set to 70 and 160 ms.
Furthermore, the difference between the $OLEev$ of both experimental paradigms (DSP and ESP) was calculated to show possible intergroup changes. The WPSS or more precisely the $OLEev$ of the ESP served as a "baseline" condition. Thus, exogenous effects resulting from elevated hearing thresholds should be minimised. In this way, the analysis of more endogenous components which are rather linked to listening effort should be optimised.

2.2. Part 2: Ongoing oscillatory EEG activity

2.2.1. Study 1: Comprehensive extraction of listening effort correlates

2.2.1.1. Hearing aid fitting

Commercially available behind-the-ear hearing aids connected to double ear-tips (double domes) were tested. The devices were fitted according to the hearing loss of the participant using a basic fitting.
The effects of the hearing aid setting *directional speech enhancement* (DSE) on the participants listening effort were examined. The DSE setting is a combination of a directional microphone technique and a Wiener filter noise reduction.
To observe objective differences regarding the listening effort, four conditions were investigated. For this, the devices were fitted with the noise reduction and the directional microphone setting was turned on and the DSE feature was set in

a) a strong (DSEstr) level and

b) a medium setting (DSEmed).

c) The DSE feature turned off (DSEoff) and as a control condition with

d) omnidirectional microphones (ODM).

The last condition was selected because it was assumed that this feature requires the largest amount of listening effort compared to the other three settings. As the outcomes between the other three hearing aid settings were not yet investigated, the ODM condition was used to make sure that at least a differentiation between a hard and an easy solvable condition could be obtained. Thus, the ODM condition was considered as a control condition.

2.2.1.2. Stimulus materials

The hearing aid settings were tested in two conditions. In the active part, the subjects had to perform a task immediately after each stimulus presentation. The speech material was taken from a German sentence test (Oldenburg sentence test (OlSa), Wagener et al. (1999)), which is primarily applied in clinical settings for the detection of the speech intelligibility threshold. Each sentence is spoken by a male voice and consists of the following structure: subject - verb - numeral - adjective - object (e.g. Peter buys three red cups). Additionally, there is no predictability on the context of the sentences.
In the passive part, the subjects had to complete the task after the presentation of the speech material. In this part, the speech materials were short stories taken from a German listening comprehension test and also recorded by a male speaker ("Der Taubenfütterer und andere Geschichten" (Thoma, 2007), level B1 according to the Common European Framework of Reference for Languages: Learning, Teaching, Assessment (Modern Language Division, 2007)).
For both cases, the speech material was embedded in multitalker babble noise (International speech test signal (ISTS), Holube et al. (2010)) composed of international speech tokens naturally produced by female voices. Additionally, a cafeteria noise was added to the audio signals consisting of clattering dishes and cutlery (downloaded from a data base of auditory signals, Data Base: AudioMicro (2013)). Furthermore, for the passive condition, the intensity of the cafeteria and the multitalker babble noise varied between two intensity levels in random time intervals between 5 and 15 s.

Calibration of the auditory stimuli

The auditory stimuli were calibrated using a hand-held sound level metre (type 2250, Brüel & Kjær, Denmark) connected to a pre-polarised free field 1/2" microphone (type 4189, Brüel & Kjær, Denmark). To measure a single sound source (signal or noise), the loudspeaker for the calibration was placed 1 m in front of the sound level metre at the level of the subject's head. Overlapping sound sources were measured in a distance of 1 m in the centre of the loudspeakers.
To assess the fluctuating noise levels of the speech material, the "equivalent continuous sound level" (L_{eq}) was selected (Brüel & Kjær, 2013). Furthermore, an A-weighting filter was applied as it is commonly used for the calibration of test stimuli for the sound field audiometry (BSA Education Committee, 2008). The

calibrated intensities were set to the following values: The intensities of the OlSa and the short stories were fixed at 65 dB L_{Aeq}. For the active condition, the ISTS noise had a level of 60 dB L_{Aeq} and the cafeteria noise was set to 67 dB L_{AFmax}. The ISTS noise used in the "passive condition" fluctuated between 64 and 66 dB L_{Aeq}, likewise the cafeteria noise changed dynamically either at 64 and at 66 dB L_{AFmax}.

2.2.1.3. Experimental design

Before the experimental session started, the subjects performed an adaptive speech intelligibility test to guarantee that the subjects were able to discriminate at least 80 % of the presented speech material. For this, the subject's hearing aids were fitted with the DSE medium setting and an adaptive OlSa was performed to achieve the speech intelligibility threshold. Thereby, the individual's intelligibility level at 50 % could be determined using the loudspeaker configuration S0N0. Finally, the 80 % intelligibility level was calculated using the stated discrimination function in the manual of the Oldenburger sentence test. To test the DSE feature, four loudspeakers were used. The speakers were positioned in a distance of 1 m from the subject's head at 0°, 135°, 180° and 225°.
As introduced in Section 2.2.1.2, two listening conditions were generated in order to extract the possible listening effort correlates: an "active" and a "passive" condition.

Active condition
For this part, 50 OlSa sentences together with the ISTS noise were presented at the frontal loudspeaker at 0°. Additionally, distracting noises, which were composed of two time-delayed ISTS and cafeteria noise sequences, were played behind the subject at the positions 135°, 180° and 225°. During the experiment, the task was to understand and to repeat words of the sentences presented at S0. A sinusoidal tone (1 kHz, duration: 40 ms) was added after each sentence to indicate the point of time where the subject's response was expected, followed by a silent gap with a duration of 5 s. The responses were written down by the experimenter.

Passive condition
In this part, the audiobook taken from the German listening comprehension test was played through the frontal loudspeaker at 0°. The loudspeakers at the rear side (at the positions 135°, 180° and 225°) presented simultaneously the two time-delayed

ISTS noise sequences plus the cafeteria noise. The subject's task was to answer easy short story related questions after the complete presentation of the audiobook. Thus, as the subjects' response was demanded after the listening condition, this part was termed as "passive" condition.

In both parts, the four different hearing aid configurations (a) DSEstr, b) DSEmed, c) DSEoff, d) ODM) were tested in a randomised order. Note that also the presentation of the active and the passive condition was randomised.
In each case, the subjects were asked to rate their perceived effort after each tested hearing aid setting using a seven point scale (LE-Scale: no effort - very little effort - little effort - moderate effort - considerable effort - much effort - extreme effort adapted from Schulte et al. (2009)) and their experienced speech intelligibility (SI-Scale: excellent - very good - good - satisfactory - sufficient - unsatisfactory - insufficient, Volberg et al. (2001)). Additionally, the subjects were asked to determine their preferred hearing aid setting for a listening situation like the presented one after the completion of each part. During both conditions, the continuous EEG was recorded from the persons with hearing loss.

2.2.1.4. Participants and inclusion criteria

A total of 14 experienced hearing aid users with a moderate hearing loss participated in this study. All 14 subjects attended in the active part of this study (mean age: 65.64 ± 7.93 years, 7 male/7 female). In the passive part a total of 12 subjects (mean age: 66.25 ± 7.74 years, 7 male/5 female) participated. The subjects were included when they had at least 80 % artifact free EEG data.
At the end, 13 subjects were included for the active part (mean age: 65.54 ± 8.24 years, 7 male/6 female). One subject was excluded due to EEG artifacts. For the passive part, a total of 10 subjects were included (mean age: 67.1 ± 7.92 years, 6 male/4 female). Here, one subject could not solve a part of the auditory task and the other one was excluded due to artifactual EEG data.
Fig. 2.3 depicts the mean pure tone audiograms (top) and the corresponding standard deviations (bottom) of the included subjects for both parts of the study.

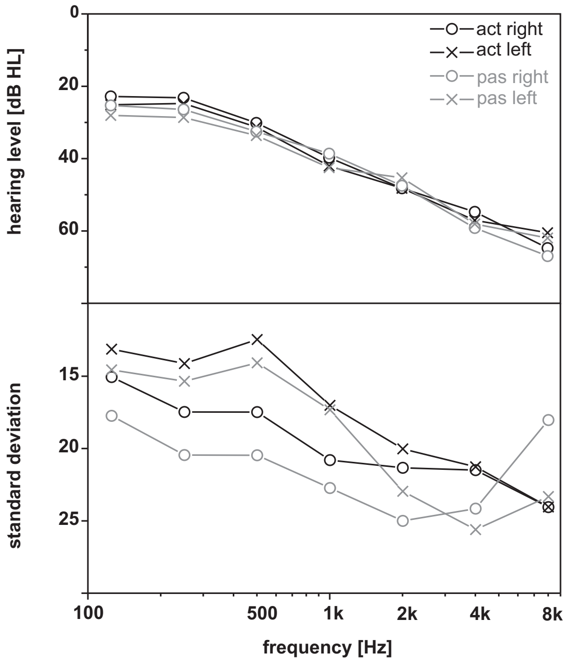

Figure 2.3.: Mean pure tone audiograms (top) and corresponding standard deviations (bottom) of the included subjects for both conditions of the study (active = black colour, passive = grey colour).

2.2.1.5. Measurement setup and data acquisition

Fig. 2.4 depicts the measurement setup for the collection of the electroencephalographic data together with the components for the auditory stimulation. The personal computer 1 (PC1) controlled the external sound card (PCMCIA Hammerfall, RME, Germany) using the software Adobe Audition. Herewith, the auditory stimulation as well as a trigger signal for the active condition were generated. The trigger signal indicated the onset of each OlSa sentence and was necessary for the segmentation of the EEG data (see Section 2.2.1.6). Furthermore, the intensity levels of the auditory signals were adjusted in the PC1. The auditory stimulation was sent from the external sound card via an audio amplifier (2 x Renkforce SAP-702) to each loudspeaker (4 x JBL Control One). The trigger signal was transferred to a multimodal trigger conditioner box (g.Trigbox, g.tec, Austria) which modified the voltage level of the incoming signal. Thus, the trigger signal could be acquired together with the electroencaphalographic activity by the biosignal amplifier (g.USBamp, g.tec, Austria). The biosignal amplifier was connected to and controlled by the personal computer 2 (PC2).

The EEG was recorded by sixteen active electrodes (g.LADYbird, g.tec, Austria)

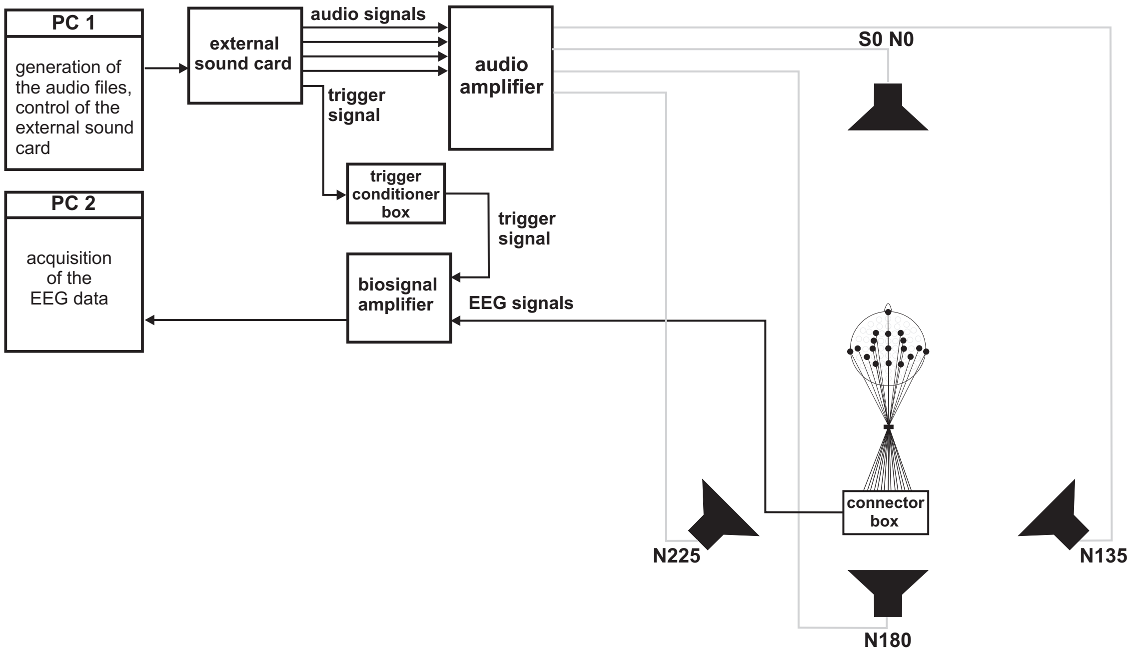

Figure 2.4.: Measurement setup for the collection of the EEG data together with the components for the auditory stimulation. On PC1 (i) the auditory stimulation and the trigger signal are generated; (ii) the external sound card is controlled; (iii) the intensity levels of the auditory signals are set. The auditory stimulation is amplified by an audio amplifier and presented via four loudspeakers. The voltage level of the trigger signal is adapted by the trigger conditioner box. The PC2 acquires the EEG data and the trigger signal coming from the biosignal amplifier.

placed according to the international 10-20 system, with Cz as reference and a ground electrode placed at the upper forehead using a sampling frequency of 512 Hz. In all measurements, the impedances were kept below 5 kΩ. The data acquisition and the post processing of the EEG data was realised using software for technical computing (MATLAB and Simulink, MathWorks Inc., USA).

2.2.1.6. Data analysis

The data was filtered offline using a bandpass filter from 0.5 to 40 Hz. For the active condition of the study, a trigger signal indicated the onset and offset of each sentence. Thus, the EEG data could be analysed during the presentation of the sentences (duration: approx. 2 s). After extraction of the EEG data for each sentence, artifacts were rejected if the maximum amplitude threshold of each EEG segment exceeded $\pm 70\mu V$. In the passive condition, artifacts were removed using a moving time window (duration: 2s) and the same artifact threshold of $\pm 70\mu V$.

For the quantification of the phase synchronisation processes of the oscillatory EEG, the distribution of the instantaneous phase on the unit circle was investigated. The instantaneous phase $\phi_{a,b}$ of each EEG channel was extracted by the application of the continuous wavelet transform (see Section 2.1.5.2 for details). In particular, the phase of a signal $x \in L^2(\mathbb{R})$ is given by the complex argument from the complex wavelet transform with the signal: $\phi_{a,b} = arg(\mathcal{W}_\psi x)(a,b)$.

In order to prove if the instantaneous phase is uniformly distributed (random process) around the unit circle or if the phase departs from uniformity and has a mean direction, the Rayleigh Test was applied to the phase data (Mardia and Jupp, 2000). For this, the mean resultant vector $\bar{R}$ of the phase values, which is also known as phase synchronisation stability (see Section 2.1.5.2), has to be calculated. Assuming there is a set of unit vectors $x_1, ..., x_M$ with the corresponding phase angles $\phi_m, m = 1, ..., M$, then the mean resultant vector can be determined by

$$\bar{R} = \frac{1}{M} \left| \sum_{m=1}^{M} e^{\imath \phi_m} \right| . \tag{2.2.7}$$

The mean resultant vector $\bar{R}$ can be interpreted as a measure of concentration of a data set. The two theoretical illustrations of Fig. 2.5 depict the phase values of a uniform (Fig. 2.5a) and a non uniform distribution (Fig. 2.5b) projected on the unit circle together with their corresponding mean resultant vector $\bar{R}$. If $\bar{R}$ is close to 0 (see Fig. 2.5a, $\bar{R} = 5.5 \cdot 10^{-17}$), then the phase values are more dispersed on the unit circle, which means that the data is distributed uniformly. Otherwise, if $\bar{R}$ is close to 1 (see Fig. 2.5b, $\bar{R} = 0.9936$), the phase is more clustered on the unit circle and has a common mean direction. The null hypothesis H_0 of the Rayleigh Test states that the data samples are uniformly distributed around the unit circle. This means it rejects uniformity when $\bar{R}$ is sufficiently large (Mardia and Jupp, 2000).

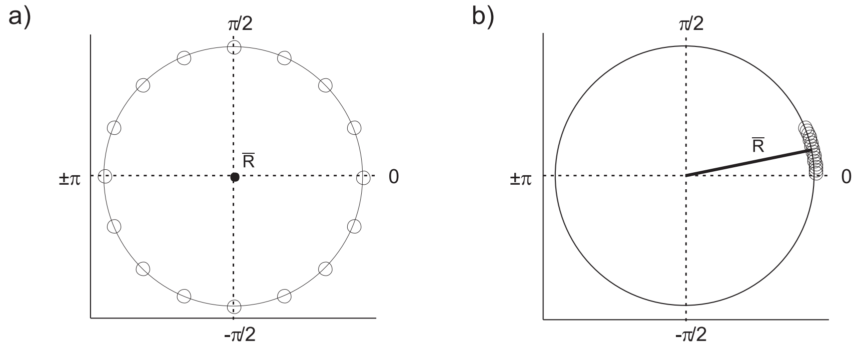

Figure 2.5.: Illustration of two data sets ($n = 16$ samples) of phase values (black circles) together with their corresponding mean resultant vector $\bar{R}$ on the unit circle showing a) a uniform distribution ($\bar{R} = 5.5 \cdot 10^{-17}$) and b) a non uniform distribution ($\bar{R} = 0.9936$).

An approximation of the probability value Pr under H_0 can be calculated (Mardia and Jupp, 2000) by

$$Pr = e^{\sqrt{1+4m+4(m^2-(m\bar{R})^2)}-(1+2m)}. \tag{2.2.8}$$

A small probability value Pr indicates that the null hypothesis is rejected, which means that the data departs from uniformity.
To facilitate the comparison between the subjectively perceived listening effort and the objective result of the Rayleigh Test on the phase values of the oscillatory EEG, it was defined that the

$$\text{objective listening effort (OLEosc)} \propto (1 - Pr) \tag{2.2.9}$$

for a specific scale a and a suitable auditory paradigm. Thus, a high value of the $OLEosc$ corresponds to a larger listening effort.

2.2.2. Study 2: Dynamic extraction of listening effort correlates

2.2.2.1. Hearing aid fitting

Behind-the-ear hearing aids connected to double domes (ear tips) were tested. The devices were fitted according to the individual's audiogram using a basic fitting. The tested hearing aid settings in this study were: a) a directional microphone setting (DM), b) a binaurally coupled hearing aid setting (BHA) and c) a setting using omnidirectional microphones (ODM). The last condition was selected since it was assumed that this feature requires the largest amount of listening effort (see Section 2.2.1.1). Thus, this condition was considered as a control condition.

2.2.2.2. Stimulus materials

The speech material were sentences taken from a German sentence test (Oldenburg sentence test (OlSa), Wagener et al. (1999)), which were also used in Study 1 in the active condition (see Section 2.2.1.2 for details). The speech material was embedded in multitalker babble noise built by shifted sequences of the speech material. Additionally, a cafeteria noise was added to the audio signals which consisted besides the environmental noises of a cafeteria, like multitalker conversations, ringing mobile phones, etc., also of clattering dishes and cutlery.

Calibration of the auditory stimuli

The auditory stimuli were calibrated as described in Section 2.2.1.2. In this setting, eight loudspeakers were used and arranged in a circle. The OlSa sentences were calibrated at loudspeaker S0 to an intensity of 68 dB L_{Aeq}. Here, the sound level metre was placed in a distance of 1 m in front of the loudspeaker. The background noises (multitalker babble noise and cafeteria noises) were presented at each of the eight loudspeakers. For the calibration of these noises, the sound level metre was placed in the centre of the eight loudspeakers with an equal distance of 1 m to each loudspeaker. Thus, the intensity of these overlapping sound sources could also be fixed to 68 dB L_{Aeq} resulting in a SNR of 0 dB.

2.2.2.3. Participants and inclusion criteria

A total of 16 experienced hearing aid users with mild to moderate hearing loss participated in this study (mean age: 61.75 ± 6.57 years, 6 male/10 female). The subjects were included when they had at least 80 % artifact free EEG data.
At the end, 15 subjects were included (mean age: 61.4 ± 6.64 years, 6 male/ 9 female). One subject was excluded due to EEG artifacts. Fig. 2.6 depicts the mean pure tone audiograms (top) and the corresponding standard deviations (bottom) of the included subjects.

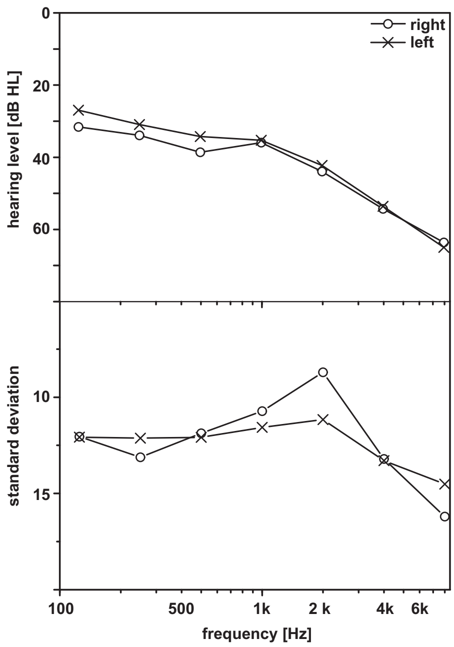

Figure 2.6.: Mean pure tone audiograms (top) and corresponding standard deviations (bottom) of the 15 included subjects.

2.2.2.4. Measurement setup and data acquisition

The measurement setup for the collection of the electroencephalographic data was the same as in Study 1 (see Section 2.2.1.5 for a detailed description). Also the setup for the presentation of the auditory signals was similar to that used to conduct Study 1. Only the audio amplifier (APART PA8250, 8 channel Professional Multichannel Amplifier, Belgium) was exchanged and four additional loudspeakers were added (in total: 8 x JBL Control One). Fig. 2.7 shows the arrangement of the eight loudspeakers. At loudspeaker 0° the OlSa sentences plus the background noises (multitalker babble noise and cafeteria noises) were presented. The remaining seven loudspeakers played only the background noises.

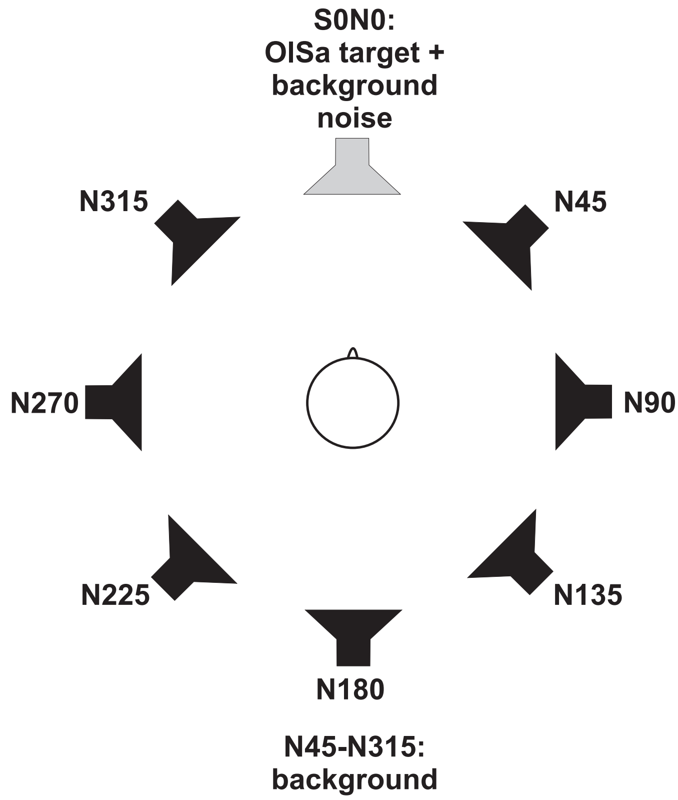

Figure 2.7.: Illustration of the loudspeaker arrangement. The auditory target (OlSa sentences) together with the background noises (multitalker babble noise plus cafeteria noises) were presented at S0N0. The remaining seven loudspeakers (N45-N315) played only the background noises.

2.2.2.5. Data analysis

The data analysis was basically the same procedure as described in Section 2.2.1.6. After offline filtering of the EEG data (bandpass: 0.5-40 Hz), the EEG signal was segmented according to the trigger signal which indicated the on- and offset of each sentence. Then, to gain the dynamical profiles of listening effort, the instantaneous phase was extracted from the EEG data using the complex argument of the wavelet transform for each presented sentence. After this, the Rayleigh Test was applied to the phase data according to the length of each sentence separately. Note that in Study 1, the instantaneous phase was extracted and the Rayleigh Test was calculated from the complete EEG channel and not, as it was done for this study, for each sentence independently (see Section 2.2.1.6). Thus, it was possible to gain dynamic listening effort profiles, which represent the $OLEosc$ (defined as $1 - Pr$-value) for each sentence, i.e. the $OLEosc$ is shown for the respective time-intervals of the sentences over the duration of the task.

Remark: Statistical analysis

In all studies, a pairwise one-way analysis of variance (ANOVA) was conducted to test if significant differences exist.

3. Results

In this chapter, the results of the two different parts of this work will be presented. First, the results for the analysis of the auditory evoked responses are shown. Then, the findings of the two studies conducted to investigate the oscillatory EEG activity are depicted.

3.1. Part 1: Auditory evoked responses

3.1.1. Behavioral analysis: Reaction time and d'prime

Fig. 3.1 depicts the means and standard deviations of the median reaction time (left y-axis, black) and d'prime values (right y-axis, grey) for the different subject groups and conditions (DSP=difficult syllabic paradigm, ESP=easy syllabic paradigm). For all middle-aged subjects, the averaged d'prime value is smaller for the DSP compared to the ESP. This difference is not noticeable for the younger subjects. Here, the d'prime showed almost the same values for both paradigms. Furthermore, an increase of the reaction time for both subject groups with hearing loss can be observed. There is also a small decrease of the reaction time for the ESP compared to their corresponding RT of the DSP in all subject groups except for the subjects with mild hearing loss. The ESP shows also a large standard deviation.

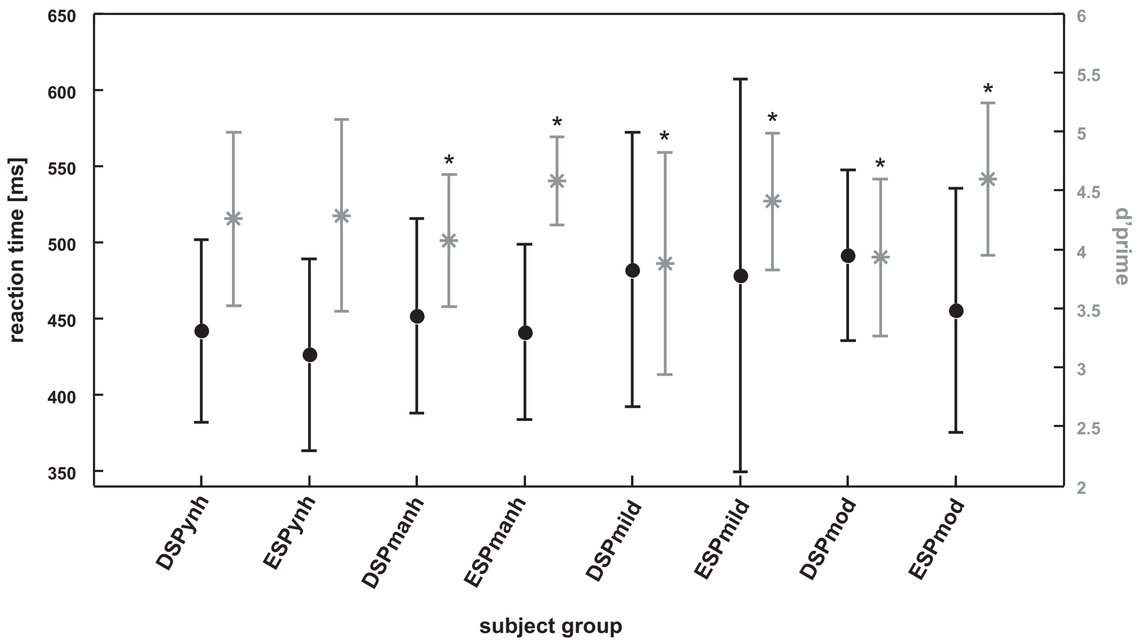

Figure 3.1.: Mean and standard deviation values of the reaction time and the d'prime for each paradigm (DSP and ESP) and subject group (ynh=young normal hearing subjects, manh=middle-aged normal hearing subjects, mild=middle-aged subjects with mild hearing loss, mod=middle-aged subjects with moderate hearing loss). Note that the intensity of the syllables was individually adjusted to each subject with moderate hearing loss.

3.1.2. Electrophysiological analysis

The mean and standard deviation values of the N1 amplitudes and latencies are shown in Tab. 3.1. In general, there is a tendency for elevated N1 amplitudes and for a decrease of N1 latencies for the DSP compared to the ESP. However, these observations were not significantly different. Furthermore, only for the subjects with mild hearing loss the latencies are larger for the ESP than for the DSP ($p<0.05$).

The WPSS was calculated for each subject and condition for the scale $a = 40$ using $M = 70$ ALR sweeps. These sweeps were evoked by correctly detected target syllables /pa/ and were free of artifacts. The scale $a = 40$ was selected due to a good temporal resolution of the WPSS in the time interval of the N1 wave as also used in Low et al. (2007); Strauss et al. (2010). This scale is equivalent to a "pseudo" frequency of 6.4 Hz in the frequency domain and is in the upper theta band.

Table 3.1.: Mean and standard deviation values of the N1 amplitudes and latencies for each subject group and condition. Note that the moderate hearing impaired subjects had individually adjusted stimulation intensities (see Section 2.1.2)

Subject group	**ynh**	**manh**	**mild**	**mod**
N1 amplitude [μV]				
DSP	-7.27±3.05	-6.14±1.78	-6.61±2.33	-6.35±1.47
ESP	-6.67±2.27	-5.58±2.14	-6.31±2.33	-5.90±1.78
N1 latency [ms]				
DSP	109.76±13.97	103.71±13.16	118.16±16.17	116.67±18.39
ESP	111.05±17.31	105.56±11.43	113.97±18.88	118.10±16.41

The wavelet used in this study was the 4th-derivative of the complex Gaussian function. This wavelet was chosen because of its symmetry, i.e. it acts as a linear phase filter and avoids phase distortions (Bradley and Wilson, 2004) and was also applied in previous studies (Strauss et al., 2008a, 2010).

In Fig. 3.2 the mean ALR waveforms for each subject group are illustrated (DSP=black line; ESP=grey line; upper panel: ynh=young normal hearing subjects (left, n=20), manh=middle-aged normal hearing subjects (right, n=20); lower panel: mild=middle-aged subjects with mild hearing loss (left, n=20), mod=middle-aged subjects with moderate hearing loss (right, n=19)).
A clear N1 component is visible for all subject groups and conditions as well as a slightly increased N1 amplitude for the DSP. Note that the mean N1 amplitude and latency values differ slightly from the values stated in Tab. 3.1 due to individual N1 latency shifts. Tab 3.1 depicts the mean values of individually analysed N1 amplitudes and latencies.

Fig. 3.3 illustrates the normalised grand averages (over all the subjects) of the WPSS (left panel) and the corresponding estimates *OLEev* (right panel, mean of the WPSS in the time interval of the N1 wave, see Section 2.1.5.2) for each subject group, from top to bottom: a) young normal hearing subjects, b) middle-aged normal hearing subjects, c) middle-aged subjects with mild hearing loss and d) middle-aged subjects with moderate hearing loss. The result of the difficult paradigm is always displayed in black, whereas the easy conditions are plotted in grey. The light grey bars in the graphics of the objective listening effort estimate represent the difference of the *OLEev* between the difficult and the easy condition (DSP-ESP).

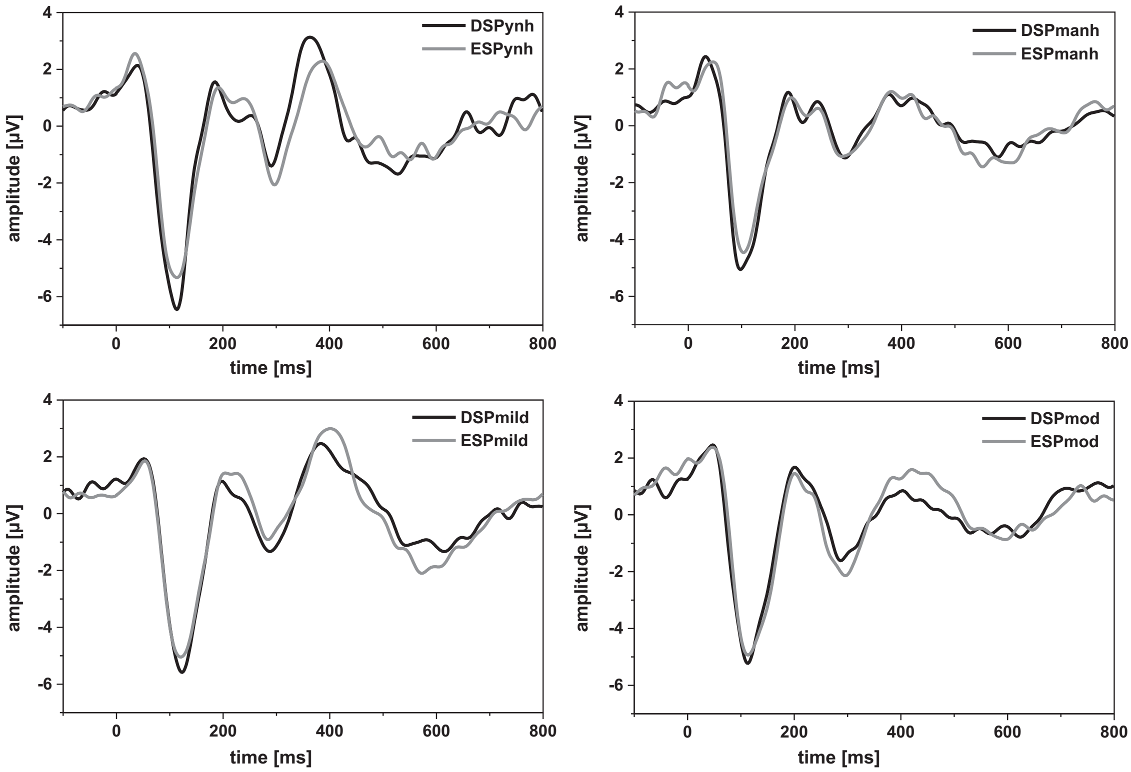

Figure 3.2.: Mean ALR waveforms for the two conditions (DSP=black line and ESP=grey line). Upper panel: ynh=young normal hearing subjects (left, n=20), manh=middle-aged normal hearing subjects (right, n=20). Lower panel: mild=middle-aged subjects with mild hearing loss (left, n=20), mod=middle-aged subjects with moderate hearing loss (right, n=19).

It is noticeable, that the WPSS in the time interval of the N1 wave (approx. 70-160 ms) is larger for the DSP as for the ESP for all middle-aged subjects ($p<0.05$, one-way ANOVA).

In the case of the young normal hearing subjects, the result of the WPSS revealed almost the same values. Furthermore, it can be seen that the difference of the $OLEev$ is slightly enhanced for the subjects with mild hearing loss compared to the coeval subject group with normal hearing sensitivity. Note that a listening effort related comparison of both subject groups with hearing loss is not appropriate due to the individual loudness adjustment of the syllables for the subjects with moderate hearing loss (see Section 2.1.2). Nevertheless, a significantly enhanced WPSS for the DSP is noticeable for the subjects with moderate hearing loss.

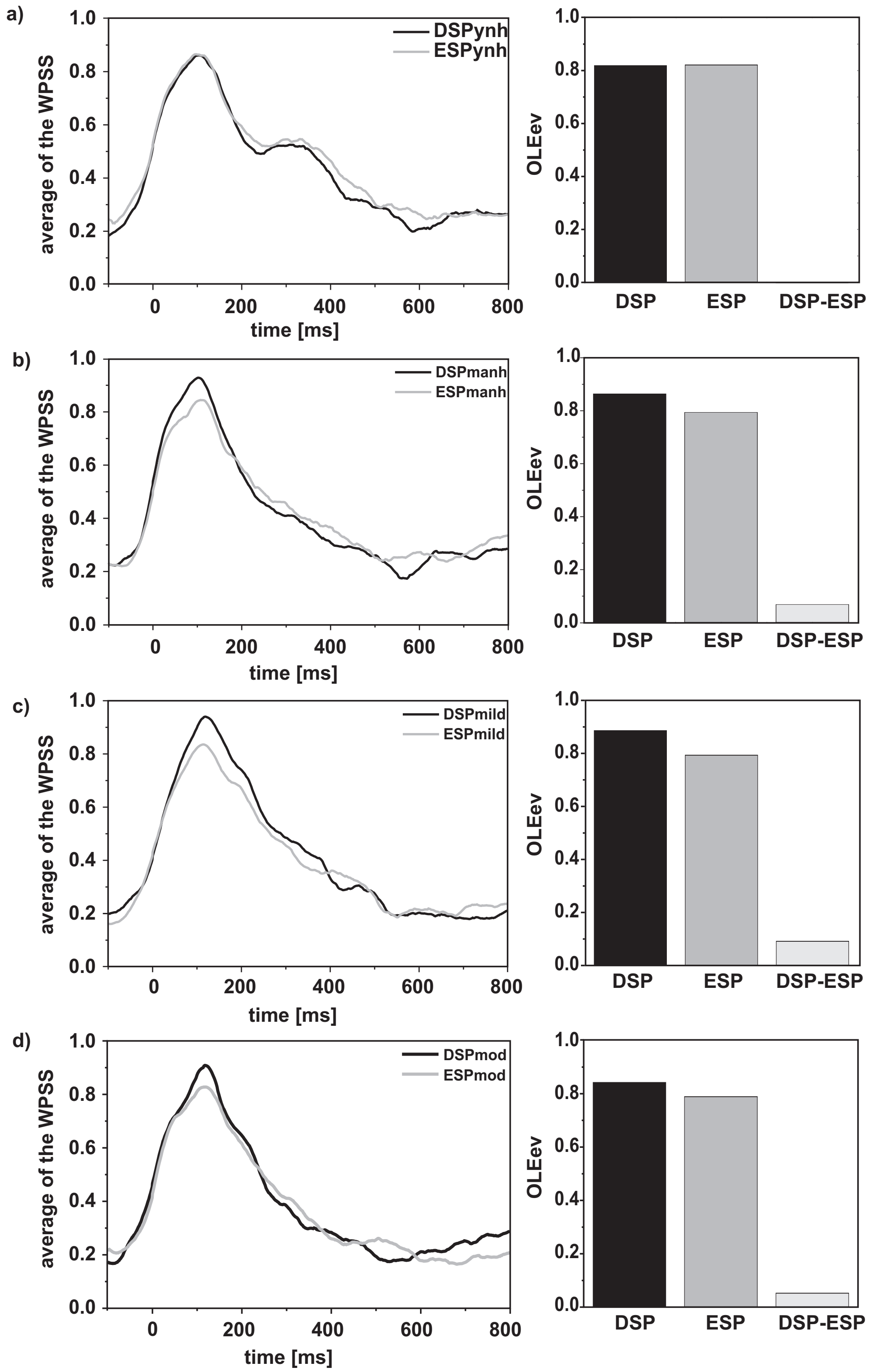

Figure 3.3.: Grand average of the WPSS (left) and the corresponding estimates *OLEev* for each paradigm (right, DSP=black and ESP=grey, difference (DSP-ESP)=light grey) and subject group (a) ynh=young normal hearing subjects, b) manh=middle-aged normal hearing subjects, c) mild=middle-aged subjects with mild hearing loss and d) mod=middle-aged subjects with moderate hearing loss and individually adjusted intensities of the syllables).

In order to make a statement about the effects of age and elevated hearing thresholds on the WPSS, the difference of the *OLEev* (see Section 2.1.5.2) was calculated for each subject of the three groups, which were tested using a fixed intensity of 65 dB SPL (young normal hearing subjects, middle-aged normal hearing subjects, subjects with mild hearing loss).
This difference was sorted by the pure tone average (PTA^{7f} of the frequencies: 125, 250, 500, 1 k, 2 k, 4 k and 6 kHz) and the age of the subject, respectively.
At the top of Fig. 3.4, the grand average of the difference of the *OLEev* divided into three PTA^{7f} ranges (4-15 dB HL: including 17 ynh and 10 manh subjects, 16-25 dB HL: including 1 ynh, 12 manh, and 7 mild subjects, 26-42 dB HL: including 13 mild subjects) is illustrated. An increment of the difference of the *OLEev* with increasing PTA^{7f} is visible. The lower part of Fig. 3.4 shows the corresponding grand average of age. A dependency between age and hearing threshold can be noticed.

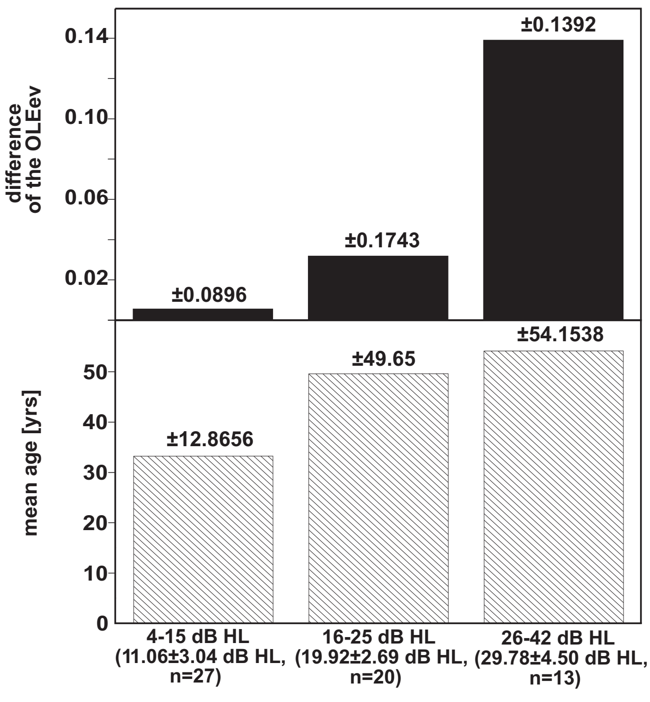

Figure 3.4.: Top: Grand average of the difference of the *OLEev* (DSP-ESP) over the three PTA^{7f} intervals. Bottom: Corresponding grand average of the age over the PTA^{7f}.

In Fig. 3.5 (top), the grand average of the difference of the *OLEev* over three age ranges (21-35 years: 20 ynh subjects, 41-50 years: 11 manh and 8 mild subjects, 51-63 years: 9 manh and 12 mild subjects) is displayed. Here, an increasing slope of the *OLEev*-difference (DPS-ESP) can be observed. The lower part of this figure shows the corresponding mean PTA^{7f} of these age ranges. An increase of the hearing threshold with increasing age is noticeable.

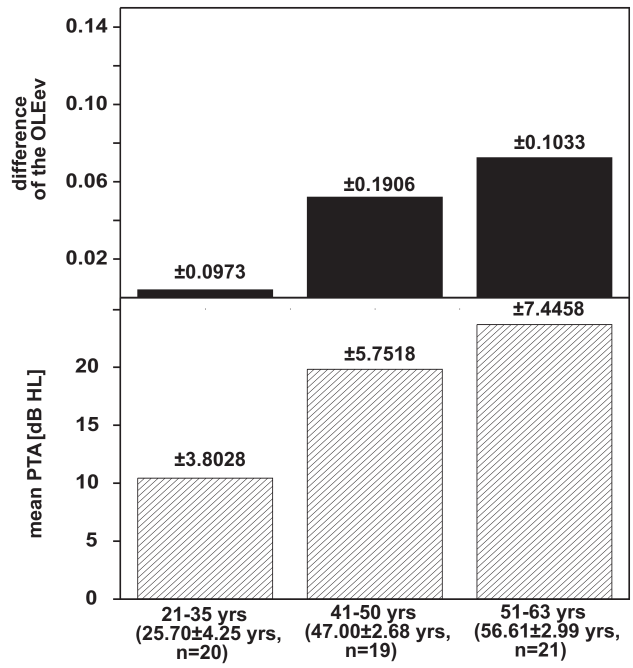

Figure 3.5.: Top: Grand average of the difference of the *OLEev* (DSP-ESP) over the three age intervals. Bottom: Corresponding grand average of the PTA^{7f} over the age.

To complement the illustrated results, Tab. 3.2 shows an overview of the results of the (one-way) ANOVA conducted on behavioral and electrophysiological data.

Table 3.2.: Results of the (one-way) ANOVA (F- and p-values)

Effect: Paradigm **Subject group**	DSP x ESP ynh	DSP x ESP manh	DSP x ESP mild	DSP x ESP mod
N1 amplitude	0.49 (p=0.48)	0.8 (p=0.3774)	0.16 (p=0.6925)	0.72 (p=0.4014)
N1 latency	0.07 (p=0.7961)	0.23 (p=0.6377)	0.57 (p=0.4558)	0.06 (p=0.8012)
WPSS	0.03 (p=0.8728)	5.93 (p=0.0197)*	5.71 (p=0.0219)*	4.5 (p=0.0408)*
d'prime	0.02 (p=0.9023)	11.08 (p=0.0019)*	4.59 (p=0.0387)*	9.81 (p=0.0034)*
RT	0.66 (p=0.42)	0.31 (p=0.5789)	0.01 (p=0.912)	2.57 (p=0.1178)
Effect: Subject group **Paradigm**	ynh x manh DSP	ynh x manh ESP	ynh x manh DSP & ESP	
N1 amplitude	2.05 (p=0.1601)	2.43 (p=0.1271)	4.48 (p=0.0374)*	
N1 latency	1.99 (p=0.1699)	1.4 (p=0.2438)	3.41 (p=0.0685)	
WPSS	4.76 (p=0.0354)*	0.24 (p=0.6246)	4.15 (0.0486)* †	
d'prime	0.77 (p=0.3852)	2.09 (p=0.1562)	0.13 (p=0.7147)	
RT	0.25 (p=0.6207)	0.61 (p=0.4401)	0.82 (p=0.3669)	
Effect: Subject group **Paradigm**	manh x mild DSP	manh x mild ESP	manh x mild DSP & ESP	
N1 amplitude	0.5 (p=0.4821)	1.05 (p=0.311)	1.55 (p=0.2168)	
N1 latency	9.59 (p=0.0037)*	2.9 (p=0.0968)	11.48 (p=0.0011)*	
WPSS	4.59 (p=0.0386)*	0.4 (p=0.5288)	1.07 (p=0.3077)†	
d'prime	0.65 (p=0.4266)	1.19 (p=0.2818)	1.4 (p=0.2401)	
RT	1.51 (p=0.2269)	1.39 (p=0.2464)	2.9 (p=0.0923)	

* Significantly different, $p<0.05$
† For the comparison of the WPSS between the subject groups,
the difference was made between the WPSS of the two syllabic paradigms.
ynh=young normal hearing subjects, manh=middle-aged normal hearing subjects,
mild=mild hearing impaired subjects, DSP=difficult syllabic paradigm,
ESP=easy syllabic paradigm, RT=reaction time

3.2. Part 2: Ongoing oscillatory EEG activity

This subsection shows the results of the two studies which were performed to analyse the phase characteristics of the ongoing oscillatory EEG activity. In these studies also the hearing aid setting which reduces the listening effort in specific listening situations was investigated. The data collected from the patients using the procedures described in Section 2.2 was processed as follows: The Rayleigh Test was performed on the instantaneous phase extracted from the EEG activity corresponding to the right mastoid electrode using the wavelet transform for a scale $a = 40$, which corresponds to a pseudo frequency of 7.68 Hz (alpha-theta border). This scale and electrode channel were used in a previous study related to the extraction of listening effort correlates. There, the correlates were gained from the evoked

EEG activity (Strauss et al., 2008a, 2010). For the analysis of the subjective listening effort scales, a number was assigned to each level of the LE-Scale (ranging from 1=very little effort to 7=extreme effort, as shown in Fig. 3.6). Then, the mean and the standard deviation of these numbers were calculated. The procedure was applied to interpret the results of the subjective speech intelligibility scales. There, the numbers assigned to each level of the SI-Scale ranged from 1=excellent to 7=insufficient, see Fig. 3.8.

3.2.1. Study 1: Comprehensive extraction of listening effort correlates

A one-way ANOVA was conducted on the $OLEosc$ values to test if differences on the listening effort regarding the applied hearing aid settings exist. Only statistical significant differences ($p<0.05$) between the $OLEosc$ values are annotated by numbers between small brackets, see Fig. 3.6.

3.2.1.1. Objective and subjective listening effort estimation

Fig. 3.6 illustrates the mean results of the objective listening effort measure (black squares; left y-axis) together with the mean results of the subjective listening effort rating (grey circles; right y-axis) over the four tested hearing aid configurations for the active part (Fig. 3.6a) and the passive part (Fig. 3.6b) of the study. Note that larger values of the $OLEosc$ indicate a larger listening effort.
The objective estimate of listening effort mirrors the subjectively perceived listening effort in all tested hearing aid settings.

In the ODM condition, which is considered as a negative control condition in this study, the subjects perceived objectively (active part: $OLEosc$=0.56±0.27; passive part: $OLEosc$=0.73±0.26) the largest listening effort and confirmed this also by subjective judgement. Here, the subjectively rated listening effort lies on the LE-Scale between considerable and extreme effort. Besides the DSEmed and DSEoff condition in the active part of the study, all other configurations were significantly different from the ODM condition.
For the active part of the study, the hearing aid setting which required the smallest listening effort was the DSEstr condition ($OLEosc$=0.26±0.26, subjective LE-scale:

moderate to considerable effort). In this case, the difference between the DSEstr and the ODM setting was significant (one-way ANOVA, p=0.008).
In the passive part, the DSEmed setting required the smallest listening effort ($OLEosc$=0.30±0.28, subjective LE-scale: moderate to considerable effort). Furthermore, significant differences were found between the ODM setting and all the other modes: DSEmed (p=0.003), DSEstr (p=0.019), DSEoff (p=0.014).

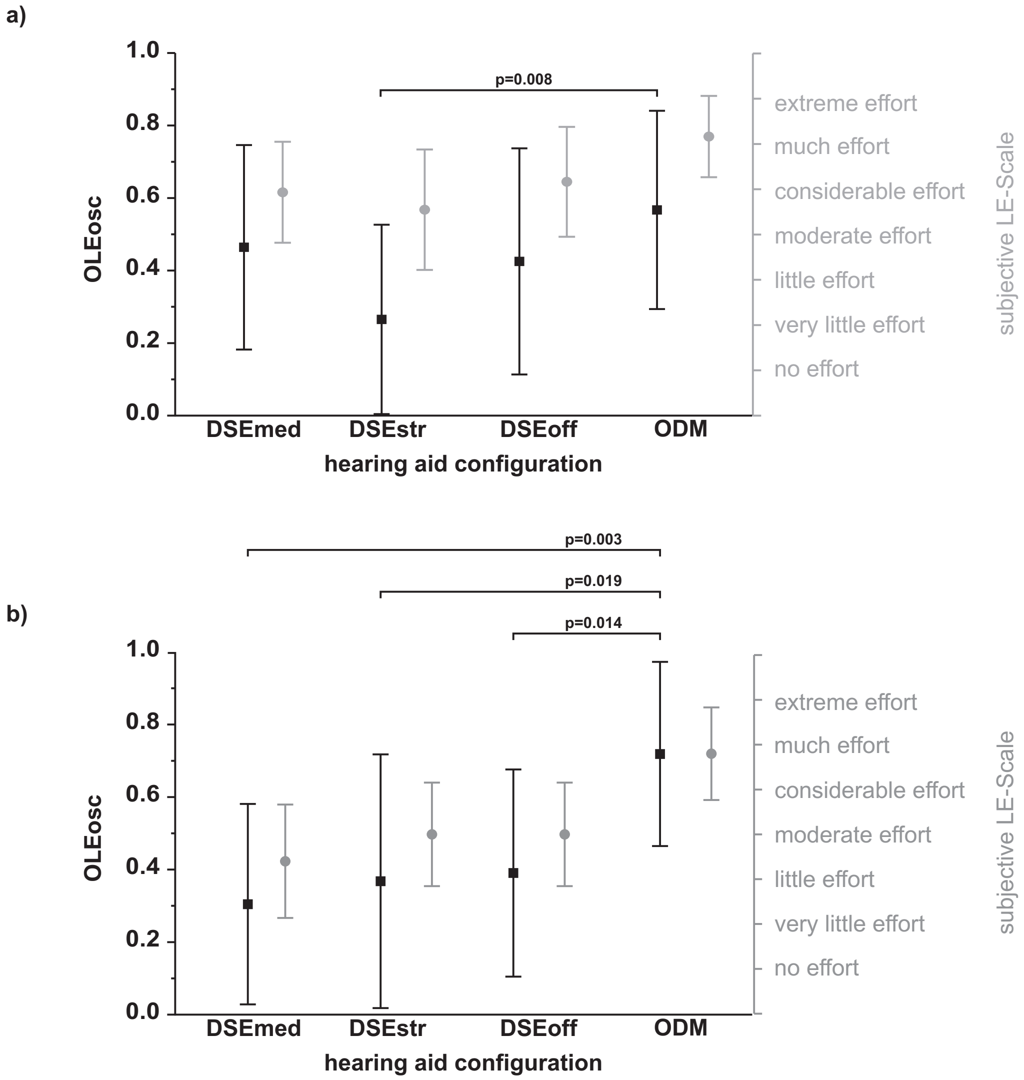

Figure 3.6.: Mean and standard deviation values of the objective listening effort measure ($OLEosc$; black squares; left y-axis) and the subjective listening effort rating (grey circles; right y-axis) from the a) active (mean over 13 subjects) and b) passive condition (mean over 10 subjects). Note that larger values of the $OLEosc$ indicate a larger listening effort.

3.2.1.2. Speech intelligibility

Fig. 3.7 depicts the mean percentage of words correctly repeated over the four hearing aid configurations of the active part of the study. The subjects reached a mean percentage of correctly repeated words around 80 % for all settings, except for the hearing aid with the ODM setting.

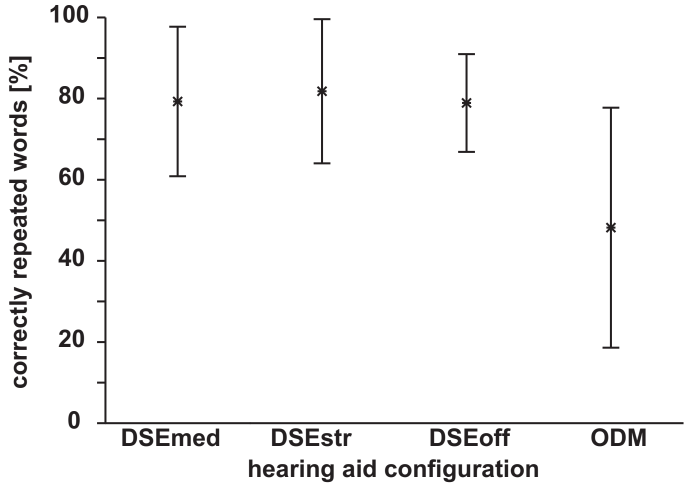

Figure 3.7.: Mean and standard deviation values of the percentage of correctly repeated words for each hearing aid setting for the active condition.

In Fig. 3.8 the mean results of the subjective speech intelligibility scale over the hearing aid configurations for the active condition are shown.

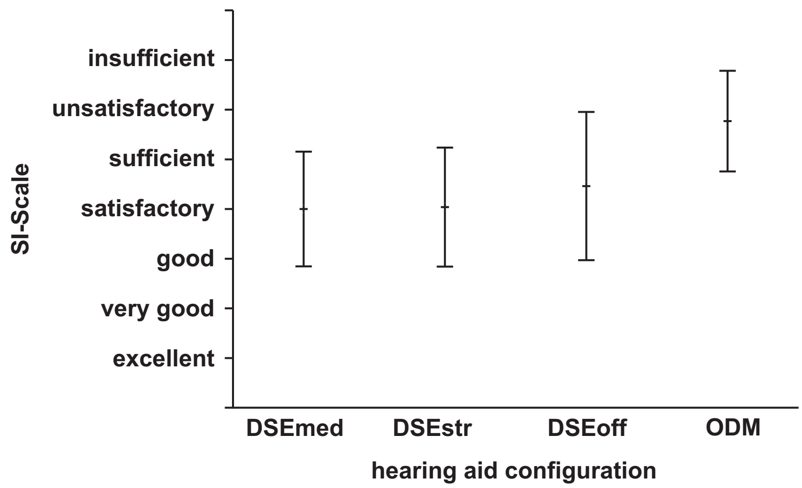

Figure 3.8.: Mean and standard deviation values of the subjective speech intelligibility scale for the active condition.

Again, the negative control configuration ODM achieved the poorest results. The mean subjective speech intelligibility rating is between "sufficient" and "unsatisfactory". The comparison of the other three hearing aid modes with each other shows that the DSEmed and DSEstr configurations are in the range "satisfactory" and are slightly better rated as the DSEoff setting, which is between "satisfactory" and "sufficient".

In Fig. 3.9, a similar behaviour of the rated speech intelligibility can be seen for the passive condition. Compared to the active condition, the speech intelligibility for the DSEmed, DSEstr and DSEoff configurations is slightly better rated, the SI is in a range between "good" and "satisfactory". Here, the mean percentage of correctly answered questions for all four hearing aid settings was 64 ± 14.59 %.

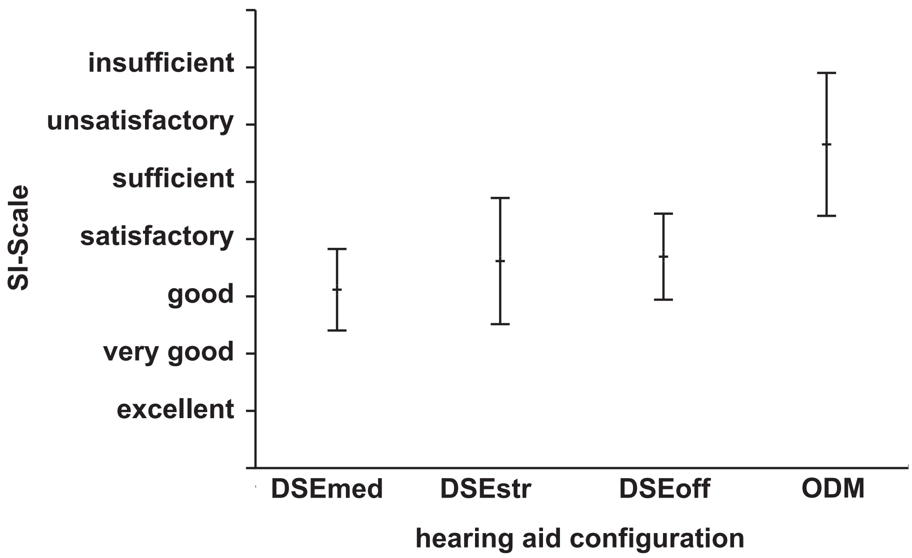

Figure 3.9.: Mean and standard deviation values of the subjective speech intelligibility scale for the passive condition. The mean percentage of correctly answered questions was 64 ± 14.59 %.

3.2.1.3. Individual results

In this part, the individual results of two subjects are shown. Fig. 3.10 represents the individual results of the listening effort of the objectively measured (left panel) and the subjectively rated listening effort (right panel).

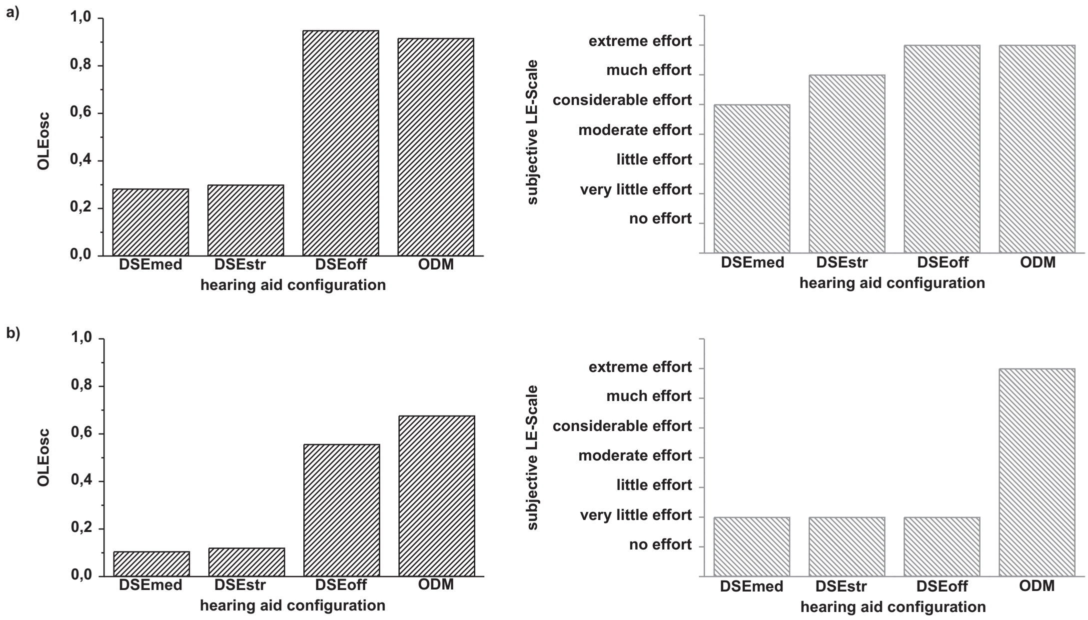

Figure 3.10.: Individual results of the objective listening effort measure (left) and the corresponding subjective listening effort scaling (right) for a) the active and b) the passive condition. Note that the preferred hearing aid setting was for both subjects the DSEmed condition, which showed also the smallest $OLEosc$ values.

The results are shown for a subject performing the active condition (Fig. 3.10a) and a subject performing the passive condition (Fig. 3.10b). Both subjects reported that for listening situations like the presented ones, their preferred hearing aid setting would be the DSEmed configuration. Furthermore, the objective listening effort measure maps the subjectively rated effort in both cases. The two settings which required the smallest listening effort were the DSEmed and the DSEstr conditions. From the 13 included subjects of the active part, 3 subjects did not perceive any differences regarding the four hearing aid configurations; in 7 subjects the $OLEosc$ confirmed the subjectively assessed hearing aid settings which required the smallest listening effort or which were scored as the preferred hearing aid settings. In the passive part, 7 out of 10 included subjects had the smallest $OLEosc$ values either for a hearing aid setting which required the smallest rated effort or for their preferred setting. One subject did not perceive any differences.

3.2.1.4. Effects of the presentation order on the objective listening effort measure

To analyse possible influences of the measurement time on the $OLEosc$, like fatigue effects, the $OLEosc$ values for each subject were sorted according to the presentation order. After this, the mean and the standard deviation values were calculated for the two parts of the study. Note that this was done additionally to the randomised testing of the hearing aid settings during the experiments. The results of this analysis are depicted in Fig. 3.11.

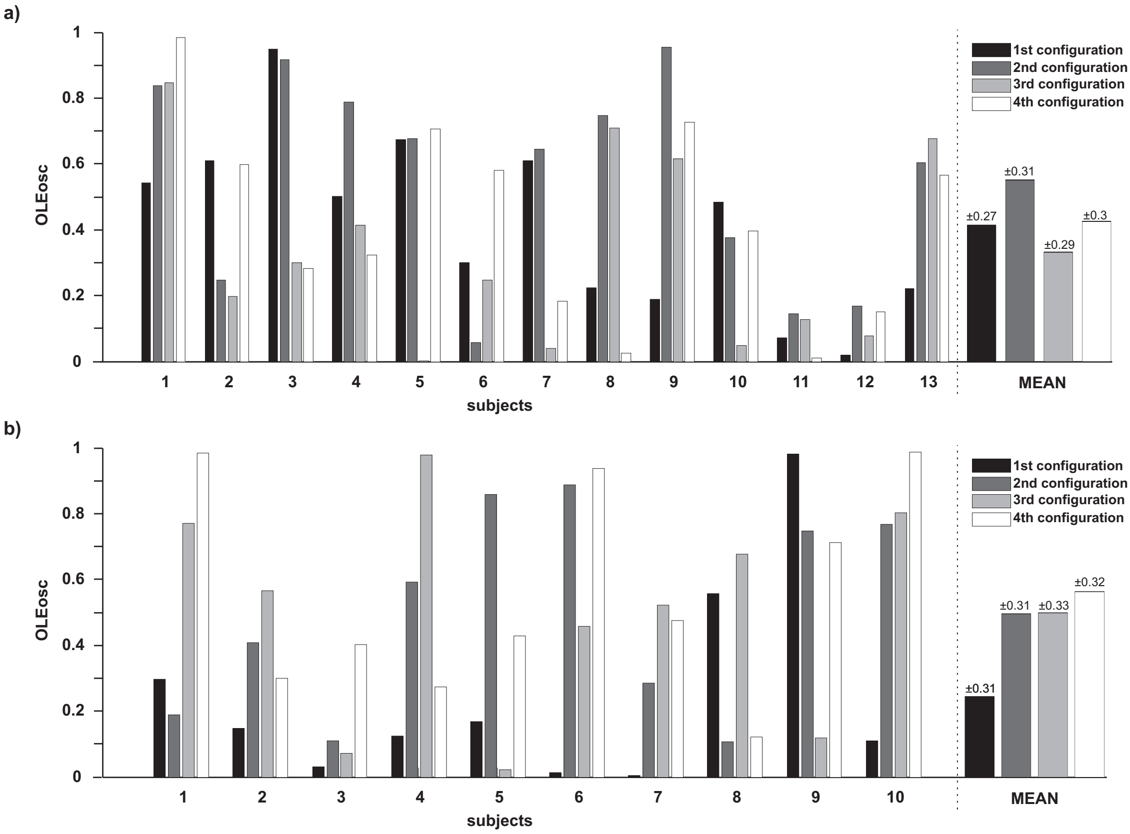

Figure 3.11.: Individual and mean results of the objective listening effort measure sorted by the presentation order of the hearing aid settings for a) the active and b) the passive condition. Note that the ascending tendencies for the subjects 1 (active part) and 10 (passive part) were related to the fact that the ODM condition, which was expected to require the largest listening effort, was presented at the end.

The upper panel (Fig. 3.11a) represents the individual and the mean values of the *OLEosc* sorted by the order of the applied hearing aid configurations (x-axis, 1st to 4th setting, black to white bars) for the active part. The lower panel (Fig. 3.11b) shows the same, but for the passive part. Besides subject 1 (active condition, Fig. 3.11a) and subject 10 (passive condition, Fig. 3.11b), there is no increasing or decreasing tendency of the objective listening effort measure related to the presentation order. In the case of the two aforementioned subjects, the presented hearing aid configurations required also an increased degree of listening effort (see Fig. 3.6, presentation order of subject 1: DSEmed, DSEstr, DSEoff, ODM; presentation order of subject 10: DSEstr, DSEmed, DSEoff, ODM). This means that the omnidirectional negative control condition was presented last and was expected to have the largest effort.

3.2.2. Study 2: Dynamic listening effort correlates

3.2.2.1. Comprehensive objective and subjective listening effort estimation

Fig. 3.12 shows the mean and standard deviation values (n=15 included subjects) of the objective listening effort measure (*OLEosc*; black squares; left y-axis) together with the results of the subjective listening effort rating (grey circles; right y-axis).

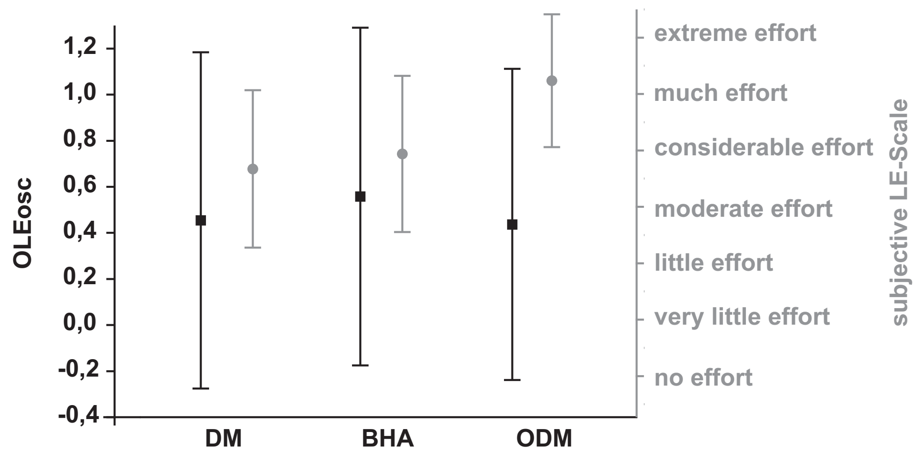

Figure 3.12.: Mean and standard deviation values (n=15) of the objective listening effort measure (*OLEosc*; black squares; left y-axis) and the subjective listening effort rating (grey circles; right y-axis). Note that larger values of the *OLEosc* indicate a larger listening effort.

As shown in Study 1, larger values of the *OLEosc* indicate also a larger experienced listening effort. It is noticeable, despite the fact that the difference between the DM and the BHA setting is small and did not reach statistical significance (one-way ANOVA), the objective listening effort measure resembles the subjectively perceived effort for the hearing aid settings with the directional microphones and the binaurally coupled hearing aid configuration. Furthermore, the negative control condition using a setting with omnidirectional microphones deviates from the expected results. Here, the *OLEosc* shows not the largest listening effort for the ODM condition although the subjects rated this condition as most effortful. This phenomenon will be further discussed in the next subsections and in the discussion chapter.

3.2.2.2. Speech intelligibility and task performance

The speech intelligibility ratings revealed, after adding a number to each level of the SI-scale (1=excellent, 2=good, 3=very good, 4=satisfactory, 5=sufficient, 6=unsatisfactory, 7=insufficient), the following mean results for the respective hearing aid setting:

DM: 4.47±1.46 (range: satisfactory-sufficient),

BHA: 4.70±1.30 (range: satisfactory-sufficient),

ODM: 5.97±1.10 (range: sufficient-unsatisfactory).

The differences between the DM and the BHA condition are minor. The hearing aid setting using omnidirectional microphones revealed the worst speech intelligibility result. Here, the SI is in the range sufficient to unsatisfactory.

Table 3.3 depicts the percentage of correctly repeated words for each subject together with the averaged values and corresponding standard deviations for the active condition. Regarding the mean values, also the difference between the DM and the BHA condition is minor. In case of the setting using omnidirectional microphones, 8 out of 15 subjects could not solve the auditory task (in total less than 50 % of correctly repeated words). Thus, it was assumed that some of these subjects stopped to solve the auditory task. In other words, they ceased paying attentional effort to understand the sentences.

Table 3.3.: Mean and individual results of correctly repeated words (crw; in percent) for each hearing aid condition.

Subject ID	**crw [%] for each hearing aid setting**		
	DM	**BHA**	**ODM**
01	72.40	84.40	25.20
02	89.60	96.40	17.20
03	78.80	92.80	72.27
04	95.60	97.60	55.60
05	29.60	41.60	38.80
06	87.20	68.00	62.40
07	94.40	72.40	70.00
08	62.80	57.60	13.60
09	77.20	68.40	30.40
10	47.20	40.80	12.00
11	71.60	72.00	42.00
12	92.00	82.80	70.00
13	92.00	77.20	85.60
14	90.40	80.80	77.60
15	61.60	49.20	48.80
MEAN	76.16	72.13	48.10
STD	19.19	18.35	24.63

3.2.2.3. Dynamical profiles of listening effort

To test if the assumption that the subjects ceased solving the auditory task in case of the ODM condition is true, the *OLEosc* was calculated for each sentence separately (see Section 2.2.2.5 for details). The mean results of this *OLEosc* analysis for each hearing aid setting (upper panel; DM=dark grey, BHA=light grey, ODM=black) together with the word scores in percent (lower panel; each sentence consisted of 5 words=100 %) over the sentences are illustrated in Fig. 3.13. Note that the solid lines represent the fitted polynominal curves (third order) of the *OLEosc* and the word score data. The figures on the left represent the dynamical listening effort profiles for all subjects (n=15), the figures in the middle show the results of n=7 subjects, which had more than an average of 50 % correctly detected words for the hearing aid configuration using omnidirectional microphones, and the figures on the right represent the results of n=8 subjects, which had less than an average of 50 % correctly detected words for the ODM setting.

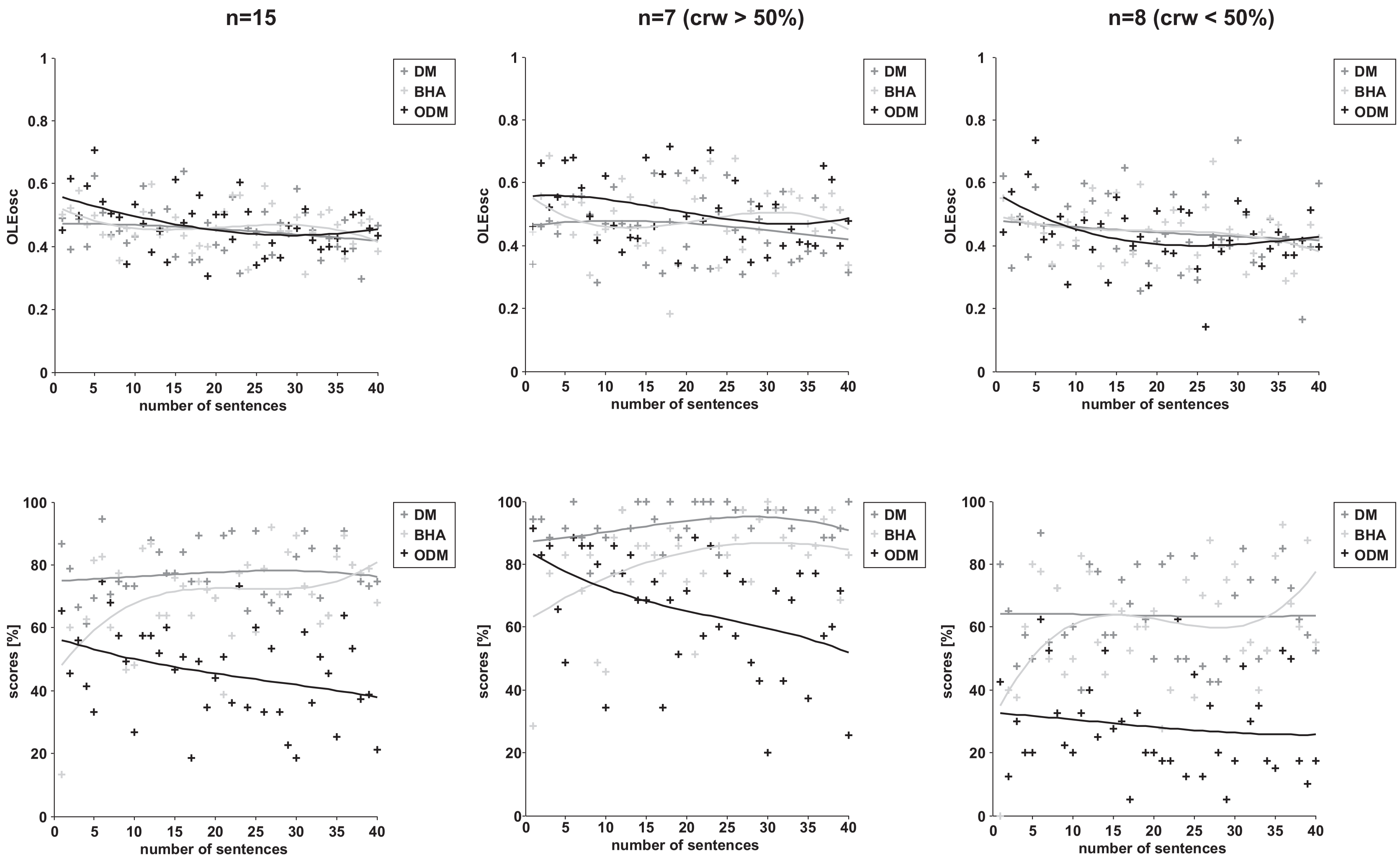

Figure 3.13.: Mean results of the dynamical listening effort profile (upper figures) with the corresponding word scores over the number of sentences for the three different hearing aid settings (DM=dark grey, BHA=light grey, ODM=black). Note that the solid lines represent the fitted polynominal curves (third order) of the *OLEosc* and the word score data. The figures on the left represent the dynamical listening effort profiles for all the subjects (n=15), the figures in the middle show the results of n=7 subjects, which had an average of more than 50 % correctly detected words for the hearing aid configuration using omnidirectional microphones, and the figures on the right represent the results of n=8 subjects, which had an average of less than 50 % correctly detected words for the ODM setting.

In all three figures of the dynamical listening effort, a similar behavior of the *OLEosc* for the ODM condition (black line) can be noticed. The *OLEosc* had larger values at the beginning, followed by a drop around sentence 15, partially under the *OLEosc* values of the DM and BHA settings which should require less listening effort. This outcome is most evident for the subgroup which could not solve the auditory task (less than an average of 50 % of correctly repeated words; right panel). The previously mentioned drop is also visible in the score data (lower figures). It can be noted that compared to the subgroup of subjects which could solve the task (mean starting values around 80 %) also the starting values of the scores were lower (around 40 %) and the decrease is flattened for the subgroup which had less than an average of 50 % of correctly repeated words.

3.2.2.4. Individual results

In this part, the individual results of two subjects are shown. Fig. 3.14 depicts the individual results of the dynamical listening effort profile (upper figures) and the corresponding word score data (lower figure) for the three hearing aid settings (DM=dark grey, BHA=light grey, ODM=black). The figures on the left show the results of a subject which could solve the auditory task (mean crw > 50 %) wearing hearing aids fitted with omnidirectional microphones. Here, it can be seen in the fitted polynominal curves of the dynamical *OLEosc* that the listening effort as well as the word score data decrease slightly for the ODM condition with the task duration. For the BHA condition, a high listening effort and a low word score is shown at the beginning (until sentence 15). After this period, the *OLEosc* as well as the word score data remain in the same ranges. Only small variations are visible by inspection of the *OLEosc* and the word score fitting curves. Solely a slight decrease of the dynamical listening effort curve at the beginning of the auditory task is visible. On the right side, the results are depicted for a subject which had in the ODM condition less than 50 % of correctly detected words. This means that the subject could not solve the given task. Here, all three fitted curves of the *OLEosc* data start at high values followed by a strong decrease over the sentences (task duration). This could be related to a motivational loss to remain task performance or to fatigue effects, see Section 4.2.2.2. Only the curve representing the results of the BHA condition increases at the end. Regarding the word score data, a clear separation between the curves of the DM and BHA settings and the negative control condition ODM can be noticed. Again, as for the previously depicted subject, the word score data for the BHA condition increases at the beginning (until sentence 15), which could be related to acclimatisation effects, see Section 4.2.2.2.

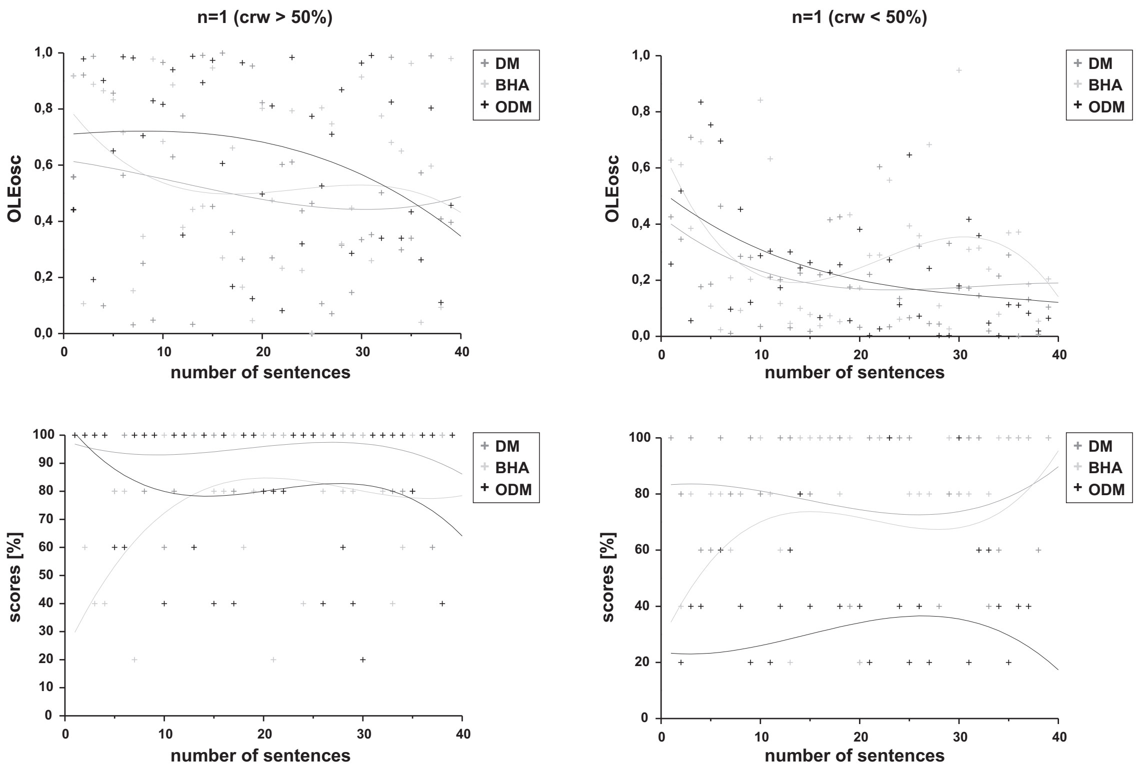

Figure 3.14.: Individual results of the dynamical listening effort profile (upper row) together with the corresponding word scores over the number of sentences for the three hearing aid settings (DM=dark grey, BHA=light grey, ODM=black). Note that the solid lines represent the fitted polynominal curves (third order) of the *OLEosc* and the word score data. The figures on the left illustrate the results of a subject which could solve the auditory task for the ODM condition (mean crw > 50 %), the figures on the right depict the results of a subject which could not solve the auditory task for the ODM condition (mean crw < 50 %).

4. Discussion

4.1. Estimation of listening effort by means of auditory evoked responses

This part of the thesis was designed to extract neural correlates of listening effort from the evoked EEG activity in young, middle-aged and subjects with hearing loss. Furthermore, behavioral data was taken into account to complement the electrophysiological results.

4.1.1. Behavioral measures and their relation to listening effort

The "accuracy data" suggests that for all middle-aged subjects the perception of information, namely the detection of the target syllable, was significantly more challenging in the case of the difficult paradigm. A comparison of the d'prime values of the participants with mild hearing loss and of the middle-aged subjects with normal hearing sensitivity revealed similar results, see Fig. 3.1. However, the accuracy measure does not reflect the effort needed to solve a task. Edwards (2007) and Downs (1982) remarked that individuals with hearing loss could have similar speech discrimination scores as normal hearing people, but they spend a different amount of cognitive effort.
The increased reaction times for the people with hearing loss could be interpreted in a way that these individuals needed a longer processing time compared to the normal hearing control groups - regardless of their age. Also Oates et al. (2002) noticed this enhancement of the reaction time in an ERP-study in people with hearing

loss. They mentioned that this observation could be related to a possible slowing of the cognitive processing. A poorer reaction time of the children with hearing loss compared to normal hearing children was likewise reported by Hicks and Tharpe (2002a). There, the listening effort was investigated in normal hearing and children with hearing loss wearing personal hearing aids using a classical dual-task paradigm. They interpreted this result as an increase of listening effort of the children with hearing loss due to poorer language skills, worse audibility of the speech material and a greater effect of the background noise.
Additionally, the reaction time is slightly enhanced for the difficult condition (DSP) for all subject groups, except for the subjects with mild hearing loss. Nevertheless, neither the inter- nor the intragroup comparison of the reaction time showed significant differences. Thus, it can be interpreted that for an experimental design like the presented one, the reaction time cannot be used as reliable measure of listening effort.

4.1.2. Electrophysiological correlates of listening effort gained from the evoked activity

Another part of this study was to analyse a possible linkage between the amplitudes and latencies of the N1 wave and listening effort for all subjects and conditions, see Tab. 3.1. The enhanced amplitudes and decreased latencies of the N1 component for the DSP suggest that there was an increased attention to detect the target syllable (Picton and Hillyard, 1974). However, this difference did not reach any statistical significance. Also Rao et al. (2010) reported increased amplitudes for a more difficult task.
The main focus of this study was to examine if the WPSS in the time interval of the N1 wave could serve as an indicator for listening effort in different age groups and in people with hearing loss.
An objective discrimination between a more effortful (DSP) and an easier listening situation (ESP) could be achieved in all middle-aged subject groups, regardless of their hearing sensitivity. The WPSS in the expected N1 time interval was significantly enhanced for the difficult task, see Fig. 3.3. Thus, it can be assumed that the subjects required more effort to solve the difficult paradigm. This result also supports the findings of previous studies (Strauss et al., 2009, 2010; Bernarding et al., 2010a). Only the young normal hearing group constitutes an exception.

Here, the WPSS showed almost the same morphology for both paradigms. In one of the former studies (Strauss et al., 2010; Bernarding et al., 2010a), the same conditions were tested in young normal hearing persons. There, the paradigms were embedded in a competing multitalker babble noise. It can be assumed that in this noisy surrounding, the detection of the target syllable exerted more effort from the younger subjects, especially for the DSP. Here, the listening effort was increased and could be objectively determined with the help of the WPSS. In the current study the noise is removed, so that both paradigms are easy to solve for the younger normal hearing subjects. Also Murphy et al. (2000) noticed in a memory-related recall task that young people tested in noisy surroundings have a similar performance as their older counterparts in quiet environments. They argue that the additional noise could be "associated with a reduction in processing resources" (Murphy et al., 2000), meaning that attention was divided.
Due to the study design, it was expected that the DSP was more difficult to solve and required more listening effort from the subjects. This tendency is visible in the middle-aged normal hearing as well as in the middle-aged subject groups with hearing loss. There, the d'prime, which reflects the difficulty level, is decreased and the proposed objective listening effort measure WPSS is enhanced for the DSP. Objectively, the difficulty level is reflected in the d'prime values and the listening effort should be objectively determined by the WPSS.
In order to reveal an intergroup comparison, the difference of the LE-level (DSP-ESP) was calculated. By applying this simple linear approach, it was assumed to reduce the exogenous components. These exogenous influences could arise due to different stimulation intensities because of elevated hearing thresholds. In this way, the focus of the analysis was more on endogenous parts which are rather linked to listening effort (see Section 2.1.5.2). Regarding the small mean difference of the LE-level for the young subjects, it can be interpreted that both paradigms were equally easy to solve by this subject group. It can be noticed that the averaged difference of the LE-level is slightly increased for the subject group with mild hearing loss compared to the young and the middle-aged normal hearing subjects. On the other hand, a small decrease of this averaged difference is visible in the results of the group with moderate hearing loss. Note that a loudness adjustment to a comfortable loudness level was done for each participant with moderate hearing loss. This last finding is not unexpected as the WPSS is analysed in the time interval of the N1 wave. Particularly, the WPSS is not only influenced by

endogenous factors like attention effects (Woldorff and Hillyard, 1991), but it is also affected by exogenous factors like intensity effects (Adler and Adler, 1989) or different interstimulus intervals (Hall, 1992, 2007). It is not transparent to which proportions the N1 component is influenced by increased attention (Picton et al., 1974; Muller-Gass and Schröger, 2007) or by the different stimulation intensities (Hall, 1992; Schadow et al., 2007) due to the loudness adjustment of the syllables. In other words, the stimulation intensity was higher for the subjects with moderate hearing loss. Due to the higher presentation level it can be argued that the speech intelligibility was additionally improved resulting in a decreased task difficulty. In this case a less endogenous effortful modulation is present. Therefore, the results of the subjects with moderate hearing loss cannot be compared with the other groups.

Moreover, it can be argued that the difficult paradigm was simplified regarding the loudness adjustments, so that the listening effort was reduced. Nevertheless, the objective discrimination via the WPSS between the two testing conditions could also be achieved in this subject group with hearing loss. Due to the variation of the intensity level of the paradigms, the subject group with moderate hearing loss was not considered for the further analysis.

4.1.3. Effects of age and hearing loss on listening effort

To examine a possible age and/or hearing loss related effect on the listening effort, the LE-level difference was calculated separately for each subject. The data were separated according to the hearing threshold (PTA^{7f} of all 7 tested frequencies), aligned to three PTA^{7f} groups and finally averaged. A clear increase of the LE-difference with larger PTA^{7f} is evident, see Fig. 3.4. This leads to the following hypothesis: Listening effort is enhanced with decreasing hearing sensitivity. The same can be noticed, if the data is arranged by age, see Fig. 3.5. As the age increases, the difference of the LE-levels shows the same tendency. Furthermore, also the hearing threshold raises with increasing age, which is in accordance to the standards (International Organization for Standardization, 2000).

In other studies to determine listening effort using dual-task paradigms, the authors noticed a similar age (Tun et al., 2009; Gosselin and Gagné, 2011) and hearing loss (Tun et al., 2009) related effect on the listening effort. The listening effort was enhanced for the subjects with hearing loss and the elder subjects. The age related increase of perceptual effort was interpreted as an outcome of a reduced cognitive performance (see Pichora-Fuller and Singh (2006) for an overview). In these studies however, the age difference between the subject groups was more pronounced. In the present case, it can be suggested that due to the small age difference between the young and the middle-aged subject groups, there is a tendency of a predominantly signal degradation related effect on listening effort.

4.2. Estimation of listening effort by means of the ongoing oscillatory EEG activity

The main objectives of this part were: (i) to estimate listening effort by means of electroencephalographic data, (ii) to investigate the effects of different hearing aid configurations on the listening effort and (iii) to uncover dynamics of listening effort which are not represented by an overall measure of listening effort.

4.2.1. Comprehensive extraction of listening effort correlates

4.2.1.1. Electrophysiological correlates of listening effort gained from the ongoing oscillatory EEG activity

The most important finding of this study is that the new objective estimate of listening effort reflects the subjectively perceived effort of the subjects with hearing loss in the active as well as the passive part of the study, see Fig. 3.6.

The results indicate that a higher value of the proposed objective listening effort measure $OLEosc$, which is defined as $1 - Pr$, mirrors a higher subjectively rated effort. This suggests that the distribution of the instantaneous phase of the EEG in the range of the theta band is correlated with cognitive effort, which means that the phase is more clustered for a demanding condition. Regarding neuronal entrainment, the cortical oscillations can be modulated by an exogenous stimuli or an endogenous source (Weisz and Obleser, 2014). Peelle et al. (2013) showed in an MEG study using noise vocoded speech that slow cortical oscillations become entrained when linguistic information is available. They argued that this phase-locking does not only rely on sensory characteristics, but also on the integration of multiple sources of knowledge, like top-down processes. Similar to these findings, Kerlin et al. (2010) found in their EEG study an attentional enhancement of the 4-8 Hz signal in the auditory cortex. They discussed that for a successful encoding of the speech, the phase-locked cortical representation of the interesting speech stream is enhanced via an attentional gain mechanism. Regarding these aspects, it can be interpreted that the EEG phase clustering in the frequency range of the theta band reflected in a high $OLEosc$ value is due to an increased effortful endogenous modulation.

4.2.1.2. Theoretical coupling of the two proposed electrophysiological measures of listening effort

It can be hypothesised that the here defined measure of the ongoing EEG activity, which depends on the phase distribution of the ongoing EEG activity, can be linked via the phase reset theory to the results gained from the analysis of the phase synchronisation stability of auditory evoked responses.

The phase reset theory states that the evoked responses can be generated by a phase resetting of ongoing frequencies. In the visual domain it was shown that the N1 wave is mainly generated by a phase alignment in the alpha and theta frequency range (Sauseng et al., 2007). Low and Strauss (2009) investigated the connection between the AERs and the EEG as well as the influence of attention on the instantaneous phase. Tone-evoked AERs were recorded from subjects in an unfocused and in a focused condition. In the focused condition, the subjects had to pay attention to a specific target. It was shown that an artificial phase reset at a specific frequency in the range of the alpha-theta band of the unfocused data resulted in an increased N1 amplitude. This modified N1 amplitude was similar to the one gained from the attentional condition. Additionally, it was demonstrated that smaller variations in the instantaneous phase of the EEG lead to an enhancement of the attention dependent N1 amplitude (see Section 1.4.2). Regarding the AER phase clustering due to auditory attention, it can be hypothesised that the presented result of a more non-uniform phase distribution of the ongoing EEG activity is due to a similar endogenous, effortful attentional modulation of the ongoing EEG.

4.2.1.3. Speech comprehension and listening effort

Comparing the *OLEosc* values with the percentage of correctly repeated words in the active condition, it can be noticed that the mean of the correctly repeated words for the three directional processing modes is almost in the same range (approx. 80 %), but the expended cognitive effort differs, see Fig. 3.6a and Fig. 3.7. Thus, it can be interpreted that the subjects had to spend a different amount of listening effort to achieve a comparable speech comprehension (Edwards, 2007; Downs, 1982). A similar effect was also found by Brons et al. (2013). There, the listening effort ratings of normal hearing subjects were obtained for different noise reduction algorithms and SNRs. They noticed that the cognitive effort needed to achieve similar speech intelligibility scores differs between the listening conditions. Also Hicks and Tharpe (2002a) found in a dual task paradigm similar speech recognition scores in the primary task for children with hearing loss and normal hearing children, but prolonged reaction times in the secondary task which served as index of listening effort. This shows that speech comprehension/intelligibility and listening effort are linked to each other, but they reflect different dimensions of auditory processing.

4.2.1.4. Effects of the presentation order on the objective estimate of listening effort

A dominant effect of the test order on the proposed objective listening effort measure can be excluded. This was ensured by a randomised presentation order of the hearing aid settings. Additionally, the *OLEosc* values were sorted and analysed in chronological order. The individual results show no systematic change over the measurement time, like an increasing or a decreasing tendency of the *OLEosc* measure. Such tendencies could be expected due to fatigue effects (Boksem et al., 2005), stress or a lack of concentration according to the duration of the measurement. As a result, the subjects could either spend an additional effort to solve the auditory task or they could lose the motivation to perform the task (Sarter et al., 2006).

4.2.1.5. Effects of the different hearing aid configurations on listening effort

The mean results of the *OLEosc* measure show that for the active and the passive listening condition the listening effort varies for the different hearing aid configurations. In all cases, the *OLEosc* mirrors the subjectively rated effort. As expected, the hearing aid configuration using the omnidirectional microphone technique, required the largest amount of listening effort in the active as well as in the passive listening condition. Ricketts (2005) discussed in a review that the use of the directional microphone technique can be an advantage for particular listening environments, for example, environments where an increase of the SNR between 4 and 6 dB leads to an adequate level of speech intelligibility. Regarding this aspect, it can be interpreted that the speech to noise ratio is improved followed by a decrease of the cognitive effort. On the other hand, Hornsby (2013) found no additional benefit of the usage of a directional processing mode. There, the listening effort was assessed by subjective listening effort ratings, word recall and the (visual) reaction time gained from a dual-task paradigm. They concluded that these outcomes could be explained by limitations in the experimental design and that a subjective rating scale may not be sensitive enough to detect small differences in listening effort between different hearing aid settings. These results were gained in a laboratory setting. Tremblay and Miller (2014) reported that findings in a laboratory setting cannot be translated into a realistic environment. As an example they alleged that participants who tested hearing aids for four weeks in daily life preferred an omnidirectional over a directional microphone setting and not as expected the directional processing mode. However, in these cited studies of Wu and Bentler (2010a,b), the main objectives were to investigate the effects of visual cues in laboratory and field tests on the overall preference. Here, speech clarity and intelligibility, self reported SNR and personal preference for the two microphone modes were investigated; listening effort was not considered.

In the active part, the noise reduction setting DSEstr demanded, for a listening situation as the presented one, the smallest effort from the subjects with hearing loss. For the passive listening condition, the DSE feature in a medium setting had the smallest *OLEosc*. It can be interpreted that compared to the directional processing mode (DSEoff) without any noise reduction algorithm, both directional processing modes combined with the noise reduction setting reduced the cognitive effort in specific listening situations. Note that these results were obtained in a laboratory setting, where the variations of the acoustic variables like SNR were kept to a minimum in both conditions. Furthermore, the noise annoyance is dependent on individual preferences (Brons et al., 2013). Thus, a general recommendation on which noise reduction setting (medium or strong level) reduces the listening ef-

fort maximally cannot be made. Nevertheless, the results indicate that the applied noise reduction algorithm has a positive effect on listening effort. Also Bentler et al. (2008) noticed that a digital noise reduction setting was preferred over a setting where the noise reduction was turned off. Here, the subjects rated their "ease of listening" in a ten-point scale.
The here presented individual results of the *OLEosc* are in line with the mean *OLEosc* results. Again, the objectively measured listening effort for both noise reduction settings revealed the smallest values. If these results are compared with the data of the subjective listening effort scale, discrepancies are visible for the subject performing the passive condition. Here, the subjectively rated effort is the same for all three directional processing modes. This could also be related to the step size of the ordinal rating scale. However, the subjectively preferred feature is the DSEmed condition. This is in line with the objective listening effort measure. It can be hypothesised that the objective measure is more sensitive as the subjectively obtained data. Furthermore, an advantage of the new measure is that the listening effort is obtained directly during the auditory task, which means that the complete time of the task is analysed. Thus, it can be assumed that the *OLEosc* is a better measure of listening effort than a subjective rating with a single value at the end of the task (Brons et al., 2013).

4.2.2. Dynamic listening effort correlates

4.2.2.1. Interpretation difficulties using an overall measure for listening effort

Regarding the mean results of the *OLEosc* of Study 2, the objective listening effort mirrored again the subjectively rated listening effort, but only for the hearing aid settings using directional microphones and a binaurally coupled hearing aid configuration. The differences between these two settings are minor. This could be related to the fact that the participants were not familiarised or acclimatised with the new BHA setting. Here, auditory acclimatisation means a performance improvement over a time period so that the hearing aid wearer can make an optimal use of the novel acoustic cues (Dawes et al., 2013). It can be hypothesised that the output of the hearing aids adapted by the BHA setting resulted in such a new sound perception. Thus, more attention was attracted by the auditory stimulation

presented via the hearing aids. This attention gathering could be an explanation for the slightly enhanced mean *OLEosc* for this condition in comparison to the DM setting. However, also Picou et al. (2014) found slight, but non-significant improvements between a moderate to a strong directional processing mode and a mild directional processing setting by using a dual-task paradigm.
Surprisingly, the mean *OLEosc* values do not reflect, as proposed in the subjective rating scales, the largest listening effort for the ODM condition. However, with respect to the results of the word score data, the mean score for the ODM condition was less than 50 %, which was also true for eight out of 15 participants. All in all, these outcomes indicate that the auditory task could not be solved by half of the subjects. This could be related to the intense background noises. In this condition cafeteria and multi-talker babble noise were presented from 8 loudspeakers placed in equal distances around the participant. Therefore, it can be considered that these subjects ceased listening to the sentences followed by a drop of attentional effort. Sarter et al. (2006) described that the attentional effort is not only dependent on the task difficulty, but also on the subject's motivation to maintain performance. Furthermore, Sarter et al. (2006) mentioned that the attentional effort is also triggered by the participants' realisation of performance failure.
Considering the results of the above-mentioned studies, the following interpretation can be made for the present one: As soon as the subjects noticed that the auditory task was at least partially unsolvable for them, they lost their motivation to maintain performance. As a consequence, the participants ceased to spend listening effort for the comprehension of the presented sentences.
Thus, a more detailed analysis was necessary to unveil if the subjects' exerted listening effort changed over the duration of the auditory task. For this, the dynamical listening effort profile was determined. Particularly, the *OLEosc* was calculated for each sentence and analysed together with the results of the word scores.

4.2.2.2. Insights given by a dynamic measure of listening effort

The grand average (n=15) of the dynamic listening effort measure, more precisely the fitted polynominal curve of the *OLEosc* for each sentence, shows that the objective listening effort decreases for the ODM condition with the duration of the task. This means, that the phase is more synchronised at the beginning of the measurement, with respect to the synchronisation of the instantaneous phase due

to attention in the frequency range of the theta band (see Kerlin et al. (2010) and Section 4.2.1.1). In other words, more attentional effort was paid at the beginning of the measurement and it ceased over the measurement time. This finding can be interpreted in the context of habituation ("degradation of the degree of attention over time" (Trenado et al. (2009a), p. 240)) or fatigue effects defined as "performance decrement over time" (Hornsby (2013), p. 529) or "failure to sustain effort over the course of a continuous [...] task" (Bryant et al. (2004), p. 114) and on a motivational decline to maintain performance (Sarter et al., 2006). To analyse this outcome more detailed, the subjects were divided into two groups based on their performance of correctly repeated words (group 1: n=7, mean crw $<$ 50 %; group 2: n=8, mean crw $<$ 50 %, see Fig. 3.13). The dynamic listening effort curve for the ODM condition shows for the subgroup, which could solve the paradigm (group 1), a slight decrease over the complete measurement time. Also the word score data shows the same tendency (decrease from approx. 80 % to 60 %). Due to the high word score data at the beginning, it can be hypothesised that the subjects' motivation to perform the task was also high. Here, the slight decrease of the $OLEosc$ could be more related to habituation and fatigue effects with respect to the measurement time than to a strong motivational loss resulting from the participants' detection of performance failure. Trenado et al. (2009a) observed the same decreasing tendency which was caused by habituation effects. There, the subjects had to solve a tone-paradigm. The mean phase difference between consecutive pairs of sweeps of auditory late responses was analysed over the sweep number (1000 sweeps). These results showed that the phase difference decreased over the time.
In the corresponding individual results (subject with crw $>$ 50 %), a clear differentiation between the ODM and the other two settings in the dynamic listening effort profiles is depicted. Nevertheless, also fatigue effects are visible which are represented in a decrease of the dynamic $OLEosc$ over the measurement time.
Note that despite fatigue effects, a small difference of the objective listening effort measure between the ODM and the DM setting is noticeable in group 1. This means that the ODM requires slightly more attentional effort.
A similar fatigue effect was shown in the dual-task study of Hornsby (2013). There, the listening effort was investigated in unaided and aided (basic and experienced fitting) subjects using as primary task a word recall task and as secondary task the reaction time to a visual marker. The auditory task was divided in blocks of 200 words. Thus, the reaction time and the word recognition could be presented

over the blocks (time). The results of Hornsby's study showed firstly an increased reaction time over the duration of the measurement, which was interpreted as an increase of fatigue, and secondly stable (only slight increase of approx. 1 %) word recognition scores over the measurement time, which were seen as being resilient to fatigue.

Regarding the outcomes of group 2, which could not solve the auditory task in the ODM setting, the polynominal curve of the word score data depicts a low starting score (approx. 35 %) followed by a slight decrease. The fitted curve of the *OLEosc* illustrates a high starting value (also higher compared to the DM and the BHA setting) followed by a decrease until approx. sentence number 15. Then the dynamic listening effort remains stable on a low value. These results may be explained in the context of a high exerted attentional effort to extract the interesting speech stream at the beginning of the auditory task, which is related to a high motivation of the subjects to perform the task. During the course of the measurement, it can be assumed that the participants recognised their performance failure followed by a self-motivated loss to sustain effort to optimise their performance. This is also mirrored in the individual results of the subject with a crw $<$ 50 % in the ODM setting. Furthermore, a difference regarding the objective listening effort estimate between the ODM and the DM setting is only possible at the beginning of the task; more effort was paid to solve the auditory task using hearing aids fitted with ODM. As mentioned before, the participant ceased to spend attentional effort to solve the auditory task over the time span of the measurement.

Boksem et al. (2006) emphasised a strong relationship between mental fatigue and the motivation of the participant to continue a specific task. The dependence between fatigue and motivation was investigated in their study by analysing the error-related negativity and the N2 potential of visual evoked potentials. The subjects had to solve a task over 2 h. Then, the potentials were averaged at the beginning (non-fatigued), at the end of the task (fatigued) and after a motivational instruction (monetary reward) at the end of the task (fatigued and motivated). The amplitudes showed a decrease between the non-fatigued and the fatigued condition as well as a regrowing of the potentials of the fatigued condition due to the motivational instruction. Thus, it can be stated that a high cognitive effort leads to fatigue. If the reward is additionally unsufficient, the motivation of the participant will also decline followed by a drop in focusing attention (Wascher et al., 2014) and a ceasing to solve the task. However, these fatigue effects on the *OLEosc* have to be analysed

in detailed.
With respect to the BHA condition, the word scores increased over the measurement time. This is noticeable in the mean results of all subjects, in the data of the two subgroups and in the individual results. Furthermore, the listening effort decreased slightly over the task duration. It can be argued that at the beginning of the measurement a higher effort was necessary to detect the interesting auditory stream followed by a slight increase because less effort was needed to maintain task performance. This increased attentional effort could also, as mentioned in the previous section, be related to acclimatisation effects. Thus, the participants had a new sound impression by the new hearing aid feature and had to acclimatise themselves to the new hearing aid fitting. However, if such an acclimatisation effect exists is controversy discussed in literature (Dawes et al., 2013) and has to be investigated in detail.

4.3. Future work and limitations

The objective listening effort estimate using phase correlates of auditory evoked potentials differentiates well between conditions requiring different degrees of listening effort in laboratory settings. This means that the presented measure can be applied to estimate listening effort using auditory stimulations with well defined parameters to exclude exogenous effects resulting from the physical characteristics of the auditory signals. Thus, in relation to the limitations in the design of the auditory stimulation, this first measure based on AERs cannot be applied directly in real world settings yet.
Regarding the extraction of listening effort by analysing the instantaneous phase of the oscillatory EEG, the measure has to be validated using a larger population of subjects. These measurements should be done in a laboratory setting using controlled conditions (e.g. a fixed auditory task). Furthermore, an acclimatisation period to the hearing aid before the test session could be of advantage in order to investigate different hearing aid settings, like a binaural hearing aid setting or a directional microphone setting. Thus, additional attention effects due to a new sound perception could be minimised.
With respect to the dynamic profiles of listening effort, also the number of subjects has to be increased to evaluate this measures. Additionally, it has to be examined

how the length of the EEG segments has to be chosen to reach an optimum balance between the required EEG data to obtain a stable measure and to have a good temporal resolution at once.
After a validation of this measure in a laboratory environment, future research should investigate if the findings can also be translated in real world settings. For this, the EEG data could be acquired using a mobile EEG system. For the extraction of listening effort correlates, the subjects could be instructed to wear the hearing aids and to attend to daily (noisy) situations. For instance, having a conversation in a cafeteria, whilst the EEG is recorded. By changing the hearing aid programme different settings could be tested. In this case, a trigger signal recorded with the EEG by the mobile biosignal amplifier could indicate the change of the hearing aid programme. To extract the listening effort correlates, the EEG could be segmented, as it was done for the EEG artifact removal of the passive part of Study 1, by using a moving time window. For this, the ideal time frame has to be determined.
Furthermore, it would be of interest to see long term effects regarding listening effort, especially with respect to a possible acclimatisation effect to the "novel acoustic cues" generated by the different hearing aid settings.

5. Conclusions

In this thesis, two new approaches were presented to extract listening effort correlates using electroencephalographic data. The two methods have in common that they are based on the distribution of the instantaneous phase of the EEG data. The working hypothesis stated that the attentional effort causes a clustering of the instantaneous phase on the unit circle due to an endogenous effortful modulation. This means that the phase values extracted from the EEG for a specific frequency have a more non-uniform distribution.
First, the concept was examined by using auditory late responses. Here, the results indicated that the inter-sweep phase synchronisation stability in the time-interval of the N1 wave reflects the attentional effort needed to solve an auditory task. Additionally, as the exogenous variables regarding the auditory stimulation were kept to a minimum, it was also shown that this synchronisation of the phase in the EEG theta band is a result of an endogenous modulation due to an increased listening effort. This method was tested in young as well as in middle-aged subjects and in people with a different degree of hearing loss.
A further finding was that reaction time data or N1 wave amplitude information hardly reflect the invested effort. Moreover, an age as well as a hearing sensitivity related effect on the listening effort by intergroup comparison of the WPSS difference was noticeable.
These findings served as a basis for the investigation of the ongoing oscillatory EEG activity, which was conducted in the second part of this work. The results indicated that the estimate of listening effort, which is also based on the distribution of the instantaneous phase of the EEG in the theta range, reflected the exerted listening effort of subjects with hearing loss. Additionally, different directional processing modes of the hearing aids with respect to a reduction of the listening effort were tested. The new estimate of listening effort indicated that the combination of a

directional processing mode together with a noise reduction algorithm in a medium and a strong setting could reduce the listening effort in specific listening situations. Furthermore, the dynamic listening effort profiles have indicated that with this new measure it is possible to detect time-varying effects like fatigue. Moreover, this data showed that overall measures have to be analysed carefully. In other words, as the listening effort estimate is based on an effortful endogenous modulation, it also reflects a ceasing to solve the auditory task. Therefore, a loss to exert listening effort can likewise be detected.

Thus, for the future objective estimation of listening effort, the auditory paradigms have to be designed in a way that they are not too hard to solve in order to maintain the listeners motivation on a similar level.

It is finally concluded that the listening effort can objectively be estimated by analysing the instantaneous phase of the EEG. A promising approach is proposed which could support the hearing aid fitting procedures in future. However, the objective estimate of listening effort extracted from the ongoing EEG has to be evaluated by a larger population of subjects. Furthermore, possible acclimatisation effects have to be taken into account to investigate the benefit of new hearing aid features with respect to a reduction of listening effort. Lastly, the findings have to be translated from specific laboratory to real listening environments.

A. Appendix

Linking the probabilitistic model via corticofugal gains to structures within the auditory pathway

As mentioned in Section 1.4.2.1, it is assumed that the abstract weights $\mathbf{w}$ of the auditory streams are modulated by three corticofugal gains G1, G2 and G3 which can be mapped to different structures of the auditory pathway and represent the corticothalamic feedback dynamics, see also Fig. A.1.

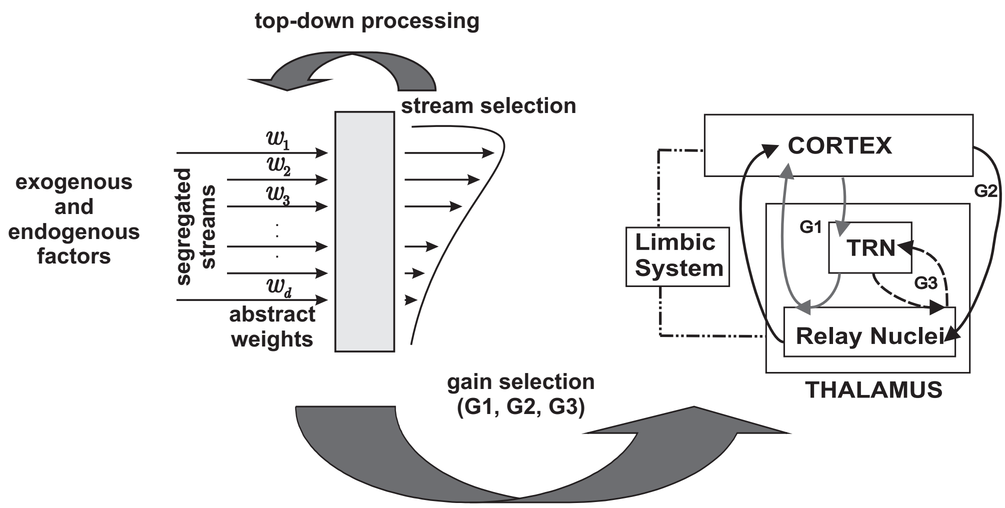

Figure A.1.: Simplified probabilistic model of the auditory stream selection together with the three corticofugal gains G1, G2, G3 and the structures of the auditory pathway (adapted from Strauss et al. (2010)).

In Strauss et al. (2010) the mapping of the gains and the gains themselves are described as follows:
Gain G1: The thalamic reticular nucleus (TRN) receives direct inhibitory input from the dorsal thalamic nuclei (relay nuclei) and indirect excitatory input from the auditory cortex via axon collaterals of projections from the cortex to the thalamus. This inhibitory effect of the TRN to the dorsal thalamic nuclei, specifically to the medial geniculate body (MGB) in terms of the auditory system, enables a regulatory effect on the corticofugal information stream. Because of this regulatory effect, gain G1 has positive as well as negative values. Furthermore, the TRN projects to the ventral subnucleus of the MGB (VMGB), which plays a role for the processing of auditory information, as well as to the medial subnucleus of the MGB (MMGB), which gets information from non-auditory pathways. Then, the VMGB has projections to the anterior and posterior auditory field and the primary auditory cortex. The MMGB projects to the primary and secondary (ipsilateral area) auditory cortex and to the ipsilateral part of the anterior and posterior auditory fields. Regarding ALRs and the numerical results obtained by Trenado et al. (2009b), it was shown by this ALR model that for an increased endogenous modulation of the incoming information, the phase synchronisation stability of single sweeps is enhanced in the time interval of the N1 component, which results in stable gains (mainly G1).
Gain G2: The medial geniculate body receives input from earlier stages of the auditory pathway and directly from the auditory cortex. Furthermore, its subnuclei (VMGB and MMGB) project back to the auditory cortex as described above. Additionally, the dorsal subnuclei of the MGB has also efferents to the auditory cortex. Gain G2 takes positives values due to the excitatory effects on the thalamus as well as on the cortex.
Gain G3: The TRN has no direct connection to the auditory cortex, i.e. the auditory cortex projects via axon collaterals (as described for Gain G1) of corticothalamic projections to the TRN. Additionally, the TRN gets input via axon collaterals of thalamocortial projections. Furthermore it projects to specific thalamic nuclei, like the MGB. These inhibitory projections, especially to the dorsal part of the thalamus, are reflected by Gain G3.

Mathematical model of corticothalamic feedback

The following is based on the large scale evoked potential theory proposed in Robinson et al. (1997); Rennie et al. (2002); Robinson et al. (2005) and is adapted to the scheme as in Strauss et al. (2010).

"According to Freeman (1991), within a neuron the relationship between the rate of incoming pulses from excitatory or inhibitory neurons Q_{ae} or Q_{ai} and their corresponding soma potentials V_e or V_i, can be obtained by an impulse response equation of the form,

$$V_{e,i}(r,t) = g\int_{-\infty}^{t} w(t-t')Q_{ae,ai}(r,t')dt', \tag{1.0.1}$$

where $w(u)$ is a causal weight function with a characteristic width that satisfies

$$\int_{0}^{\infty} w(u)du = 1.$$

As stated in Robinson et al. (1997), a suitable choice for $w(u)$ is given by

$$w(u) = \begin{cases} \frac{\alpha\beta}{\beta-\alpha}(e^{-\alpha u} - e^{-\beta u}), & \beta \neq \alpha \\ \alpha^2 u e^{-\alpha u}, & \alpha = \beta \end{cases} \tag{1.0.2}$$

for $u > 0$, where α and β represent the rise and decay times of the cell–body potential produced by an impulse at a dendritic synapse. By combining equations (1.0.1) and (1.0.2) for the case $\beta \neq \alpha$, the mean field soma potential V_a ($a = e, i$ excitatory and inhibitory), representing the synaptic inputs from various types of afferent neurons that are summed after being filtered and smeared out in time as a result of receptor dynamics and passage trough the dendritic tree, is governed by the following equation

$$D_\alpha V_a(r,t) = \sum_b N_{ab}S_{ab}\phi_b(r, t-\tau_{ab}), \tag{1.0.3}$$

where

$$D_\alpha = \frac{1}{\alpha\beta}\frac{d^2}{dt^2} + (\frac{1}{\alpha} + \frac{1}{\beta})\frac{d}{dt} + 1, \tag{1.0.4}$$

N_{ab} is the average number of synapses from neurons of type $b = e, i, s$ on neurons of type $a = e, i$, where s stands for subcortical, S_{ab} represents the magnitude of postsynaptic potentials, ϕ_b represents fields of incoming pulses and τ_{ab} are synaptic time delays.

In accordance to Rennie et al. (2002), the mean firing rate Q_a related to cell–body potential V_a is given by a sigmoidal–type function

$$Q_a = \frac{Q_{max}}{1 + exp\{-[V_a(r,t) - \theta_a]/\sigma_a\}}, \tag{1.0.5}$$

where θ_a is the mean firing threshold of neurons of type a, σ_a is the standard deviation of this threshold in the neural population, and Q_{max} is the maximum attainable firing rate.

An assumption that reflects the large–scale effect of neural populations is that each part of the corticothalamic system produces a field ϕ_a of pulses, which travels at a velocity v_a through axons with a characteristic range r_a. Approximately such pulses propagate according to the damped-wave equation

$$\left(\frac{1}{\gamma_a^2}\frac{\partial^2}{\partial t^2} + \frac{2}{\gamma_a}\frac{\partial}{\partial t} + 1 - r_a^2\nabla^2\right) C_a(r,t) = Q_a, \tag{1.0.6}$$

where $\gamma_a = v_a/r_a$. The simulation of auditory evoked cortical streams S is achieved by using a thalamic stimulus $A_n(k,w)$ of angular frequency w and wave vector k, which for convenience is chosen to be a unit Gaussian in time and space (centered at t_{0s} and r_{0s} with standard deviations t_s and r_s), together with a function $C_l(k,w)$ which refers to an excitatory cortical activity of a short-range neural population with local connectivity, to define a corticothalamic transfer function

$$\frac{C_l(k,w)}{A_n(k,w)} = A\frac{L_l L_s}{(1 - L_l I_{ll} - L_i I_{ii})}\frac{e^{iwt_d}/2}{(1 - L_s L_r(G3))}\times$$

$$[1 + \frac{1}{(k^2r_e^2 + q^2r_e^2)}(\frac{L_e I_{ee}}{(1 - L_l I_{ll} - L_i I_{ii})} +$$

$$\frac{L_e L_s((G2) + (G1)L_r)e^{iwt_d}}{(1 - L_l I_{ll} - L_i I_{ii})(1 - L_s L_r(G3))}],$$

in which t_d represents the time delay between thalamus and cortex, A represents an amplitude scaling factor, $G1, G2$, and $G3$ are the relevant corticofugal and intrathalamic gains, L_z represents a dendritic transfer function that exerts a low pass filtering effect as given by

$$L(w) = (1 - \frac{i\omega}{\mathcal{N}_1})^{-1}(1 - \frac{i\omega}{\mathcal{N}_2})^{-1}, \tag{1.0.7}$$

where $\mathcal{N}_1 = \alpha, \eta_1$ and $\mathcal{N}_2 = \beta, \eta_2$, and I_{xy} stands for secundary gain factors. The indices for the parameters $x, y, z = e, i, l, s, n, r$ denote e excitatory, i inhibitory, l

excitatory cortical neurons with local axons, s excitatory thalamic neurons in specific and secondary relay nuclei, n excitatory afferents to thalamus, and r inhibitory thalamic reticular neurons. The response to the thalamic stimulus is given by

$$\Delta(r,w) = \int_0^\infty \frac{kdk}{2\pi}\frac{C_l}{A_n}e^{-1/4k^2r_s^2}J_0(k|r-r_{0s}|), \tag{1.0.8}$$

and the desired evoked potential streams can be obtained by using

$$R(r,t) = \mathcal{F}^{-1}\{e^{-1/2\omega^2 t_s^2}e^{i\omega t_{0s}}\Delta(r,w)\}(r,t), \tag{1.0.9}$$

here J_0 denotes the bessel function of first kind, and $\mathcal{F}^{-1}$ denotes the inverse Fourier transform, see Rennie et al. (2002) for more details." (Strauss et al. (2010), p. 129f).

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List of Publications

Journal Papers (MEDLINE listed)

C. Bernarding, D. J. Strauss, R. Hannemann, H. Seidler, and F. I. Corona-Strauss. A New Technique for the Objective EEG-aided Assessment of Listening Effort - Influence of Different Hearing Aid Settings, *in preparation*.

C. Bernarding, D. J. Strauss, R. Hannemann, H. Seidler, and F. I. Corona-Strauss. Neural correlates of listening effort related factors: Influence of age and hearing impairment, *Brain Research Bulletin*, vol.91, pp. 21-30, 2012.

D. J. Strauss, F. I. Corona-Strauss, C. Trenado, **C. Bernarding**, W. Reith, M. Latzel, and M. Froehlich. Electrophysiological Correlates of Listening Effort: Neurodynamical Modeling and Measurement. *Cognitive Neurodynamics*, Vol. 4 (2), pp. 119-131, 2010.

Papers in Proceedings (MEDLINE listed)

C. Bernarding, D. J. Strauss, R. Hannemann, H. Seidler, and F. I. Corona-Strauss. Objective Assessment of Listening Effort in the Oscillatory EEG: Comparison of Different Hearing Aid Configurations,*In Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, pp. 2653-2656, 2014.

C. Bernarding, D. J. Strauss, R. Hannemann, and F. I. Corona-Strauss. Quantification of Listening Effort Correlates in the Oscillatory EEG Activity: A Feasibility Study, *In Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, pp. 4615-4618, 2012.

C. Bernarding, D. J. Strauss, M. Latzel, R. Hannemann, J. Chalupper, and F. I. Corona-Strauss. Simulations of Hearing Loss and Hearing Aid: Effects on Electrophysiological Correlates of Listening Effort, *In Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, pp. 2319-2322, 2011.

C. Bernarding, D. J. Strauss, R. Hannemann, M. Latzel, H. Seidler, U. Jobst, A. Bellagnech, M. Landwehr, and F. I. Corona-Strauss. The Effects of Age and Hearing Impairment on the Extraction of Listening Effort Correlates, *In Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, pp. 2323-2326, 2011.

A. Bohr, **C. Bernarding**, D. J. Strauss, and F. I. Corona-Strauss. Effects of Auditory Selective Attention on Chirp Evoked Auditory Steady State Responses, *In Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, pp. 2013-2016, 2011.

C. Bernarding, F. I. Corona-Strauss, M. Latzel, and D.J. Strauss. Auditory Streaming and Listening Effort: An Event Related Potential Study. *In Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, pp. 6682-6685, 2010.

C. Bernarding, D.J. Strauss , M. Latzel, and F. I. Corona-Strauss. Non-Listening Effort Related Parameters in Auditory Discrimination Paradigms. *In Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, pp. 6817-6685, 2010.

D.J. Strauss, F. I. Corona-Strauss, **C. Bernarding**, W. Reith, M. Latzel, and M. Froehlich. On the Cognitive Neurodynamics of Listening Effort: A Phase Clustering Analysis of Large-Scale Neural Correlates. *In Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, pp. 2078-2081, 2009.

Conference Papers (Peer-Review, not MEDLINE listed)

C. Bernarding, D. J. Strauss, R. Hannemann, H. Seidler, and F. I. Corona-Strauss. Objective Extraction of Dynamical Listening Effort Profiles, *accepted*.

C. Bernarding, R. Hannemann, D. Herrmann, D. J. Strauss, and F. I. Corona-Strauss. Extraction of Listening Effort Correlates in the Oscillatory EEG Activity: Investigation of Different Hearing Aid Configurations, *In Proceedings of the 6th International IEEE EMBS Conference on Neural Engineering*, pp. 1258-1261, 2013.

F. I. Corona-Strauss, **C. Bernarding**, M. Latzel, and D. J. Strauss. Syllable Evoked Auditory Late Responses: Effects of Noise Onsets and Noise Types, *In Proceedings of the 5th International IEEE EMBS Conference on Neural Engineering*, pp. 140-143, 2011.

C. Bernarding, M. Latzel, D. J. Strauss, and F. I. Corona-Strauss. On the Objective Electrophysiological Estimation of Listening Effort in Different Age Groups, *In Proceedings of the 5th International IEEE EMBS Conference on Neural Engineering*, pp. 671-674, 2011.

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