

Performance optimized signal processing in objective audiometry – Digital tools for the efficient measurement and use of AEP and OAE

Abstract

The objective methods of audiometry are based on the registration of noisy signals of small amplitude which are contaminated with interferences of diverse and variable sources. The reconstruction of the target signal is performed by selection and averaging of many signal epochs. Beyond these basic digital tools of signal processing, whose practical value is well-proven, other procedures which have never been established in commercial devices are potentially suitable to further improve the signal quality and the reliability of its detection. They are subject of this paper, as well as several new approaches, some of which have already been proven as useful in clinical practice while others are still awaiting their practical application. By implementation of these procedures, the precision and reliability of the diagnostic conclusions derived from otoacoustic emissions (OAE) and auditory evoked potentials (AEP) can be enhanced substantially.

This review aims to exploit the potential of signal processing to the highest possible extent. Its main topics are a digital filter optimized for the detection of AEPs, the extension of averaging to the polarity of the signal, the extraction of parameters qualified as measures for quality and significance, and the time-differential analysis of correlation. Furthermore, the digital superposition of several independent recordings, the use of the amplitude distribution density and an algorithm developed to reduce the impact of residual noise on the response threshold are described for the first time.

Keywords: objective audiometry, auditory evoked potentials, otoacoustic emissions, response detection, residual noise, signal statistics

Introduction

Dedicated to my mentor *Dr. Michael Berg* who introduced me to audiology.

Objective audiometry is based on the measurement of signals which are generally superimposed by unavoidable interfering influences of different origin. Since the beginnings of electrical response audiometry (ERA), signals of biological origin have been detected by measuring them in multiple copies and processing the registered signal epochs to an average value already during data acquisition. Signal averaging reduces the influence of the interfering signals of biological and non-biological origin, because the amplitude of the physiological response increases linearly with the number of summations, whereas the amplitude of the noise increases only in proportion to the root of the number of summations: 100 summations lead to a 100-fold increase of the response but only to a ten-fold increase of the interfering noise. However, the resulting simple rule for the gain in noise suppression (e.g. increase of the signal-to-noise ratio by 30 dB for 1,000 iterations) is valid only under the condition of

stochastic interference signals whose relevant properties do not change in the course of the measurement. In real life, the condition of stationarity is at best approximately fulfilled, since there is practically no measurement without slow changes in conditions and/or sporadically occurring disturbances. The number of factors influencing these interferences is large and highly variable both inter- and intra-individually.

Needless to say that there is great interest in maximizing the suppression of interferences. In the case of otoacoustic emissions (OAE), the use of which is limited to obtaining the dichotomous statement “signal detection successful or not successful” in the vast majority of practical applications, the efficient reduction of background noise alone constitutes the value of the method, since its impact can go as far as the complete masking of actually present responses. Likewise, the determination of response thresholds by means of the auditory evoked potentials (AEP) is practically impossible or at least inadmissibly inaccurate without a differentiated consideration of the disturbing influences.

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These initial remarks are intended to make clear that, firstly, the quality of recording determined by the residual noise must be promoted with all available tools, secondly, that this must be done on a case-by-case basis, and thirdly, that the success achieved must be described with suitable measures and given to the user in the documentation. The assessment “signal detection not successful” can have many reasons – a pathological event is only one of these reasons.

It is the aim of this review to help the signal processing methods suitable and available for the reconstruction of physiological responses to become more popular and to contribute to the intensification of their practical application. This endeavor seems justified to the author, because in medical technology it is often observed that established diagnostic procedures are somewhat sluggish with respect to technical extensions or innovations. However, the quite understandable inclination to stick to the tried and tested should not go so far that, for example, the limitations of computer performance in the early days of objective audiometry still determine the procedures applied today. The author is convinced that the procedures described below contain dormant reserves, the consistent use of which could further improve the performance of objective audiometry.

Digital filters

The measurement probe (i.e. the ear canal microphone in the case of OAE and the electrodes for AEP) always captures many signal frequencies, including those that do not contribute to the detection of the physiological signal. For AEP, it is shown in Figure 1 that the EEG signal experienced by the electrodes in the frequency range from 1 to 5,000 Hz contains all components of short, medium, and long latency, but exceeds these target signals by up to 60 dB. For this reason, the digitization of the analog signal is always preceded by electronic hardware filters whose task is to limit the influence of signal components of irrelevant frequencies. It is understandable that a variable choice of the passband or cutoff frequencies is advantageous: For example, in FAEPs (early AEPs) the low frequency components are rather undesirable if the goal is to determine the latencies as accurately as possible, but in near-threshold measurements or in the case of high-frequency hearing loss they are useful or even indispensable (as demonstrated in Figure 2 and Figure 3).

However, signal filters can be realized not only as analog electronic circuits, but also as digitally programmed software components. These have the advantage that they can be flexibly designed and applied not only online but also offline, i.e. after data acquisition has been completed. Unlike a hardware filter, the original data remain untouched in a posteriori filtering and are available for optional further processing.

The elimination of (unwanted) high-frequency signal components is most easily achieved with three-point

smoothing (the k -th value M_k of the curve is calculated from the original value and its neighbors according to the formula $0.23 \cdot M_{k-1} + 0.54 \cdot M_k + 0.23 \cdot M_{k+1}$). This corresponds to a single-pole low-pass filter with a slope of 6 dB per octave. Higher order digital filters allow larger slopes but require more complex calculations. The effect of a four-pole bandpass filter (high-pass 300 Hz, low-pass 1,800 Hz) especially optimized for ABRs is shown in Figure 2. This filter is dimensioned so that the phase shifts of high-pass and low-pass compensate each other at the frequency centroid of the FAEP spectrum (at about 700 Hz) [11]. The programming code comprises only a few lines, relatively independent of the programming language.

The effect of the same filter realized with different parameters on the SAEP (late AEP) is shown in Figure 4. The cortical responses regularly show a vertex-negative maximum N1 at 100 ms and a minimum P2 at 200 ms, with no relevant dependence on stimulus level. This time structure corresponds fairly closely to a frequency of 5 Hz with little energy at other frequencies. Therefore, narrow-band filtering with a passband of 3 Hz to 7 Hz is possible and beneficial for isolating the physiological signal from the background noise.

Any filter causes amplitude loss in both the wanted and the unwanted signal components; it is the task of filter dimensioning to balance the losses in favor of the useful signal. In Figure 3 it is shown that the high-pass filter improves the detectability of the FAEP responses J1 and J3 in the range of medium stimulus levels. On the other hand, for the representation of J5 near threshold, the broadband version is better, since the long-wave components of the curves are lost in the high-pass filtering.

Beyond the filter described here, several realizations of phase-error-free digital filters are available in almost all of today's measuring systems. It is up to the user to select an option for the reversible post-processing of the measured curves whose properties and dimensioning are appropriate for the respective problem.

When measuring OAE, (digital) filtering reduces the effect of the low frequencies dominating in the acoustical background and thus potentially improves the signal-to-noise ratio. High-pass filtering is implemented in special measurement paradigms (e.g. QuickScreen mode in ILO92, Otodynamics Limited, UK), which lead to better results in case of unfavorable acoustic conditions. In addition to attenuating the low frequency components, the time range of the signal recording is shortened (e.g., 12 instead of 20 ms) because the amplitude of the low frequency responses occurring in the late latency range (from 12 to 20 ms) and generated in the apical regions of the cochlea [4] is reduced by the filter. The combination of the omission of low-frequency and long-latency components improves the signal detection for the high frequency physiological responses of baso-cochlear origin and shortens the examination time.

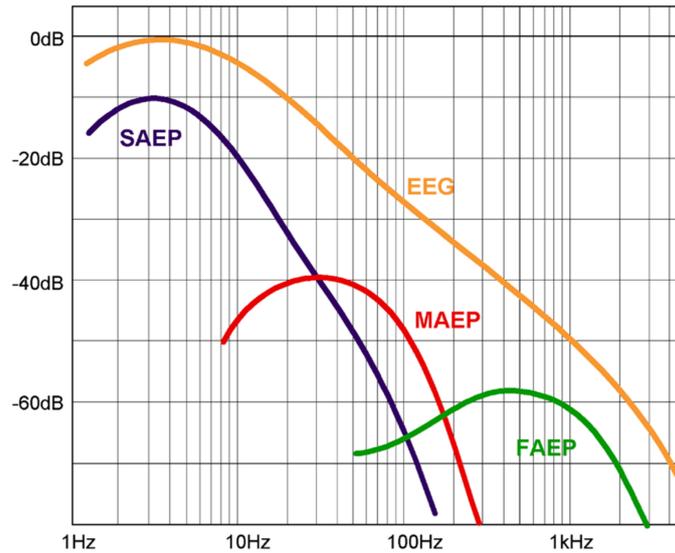
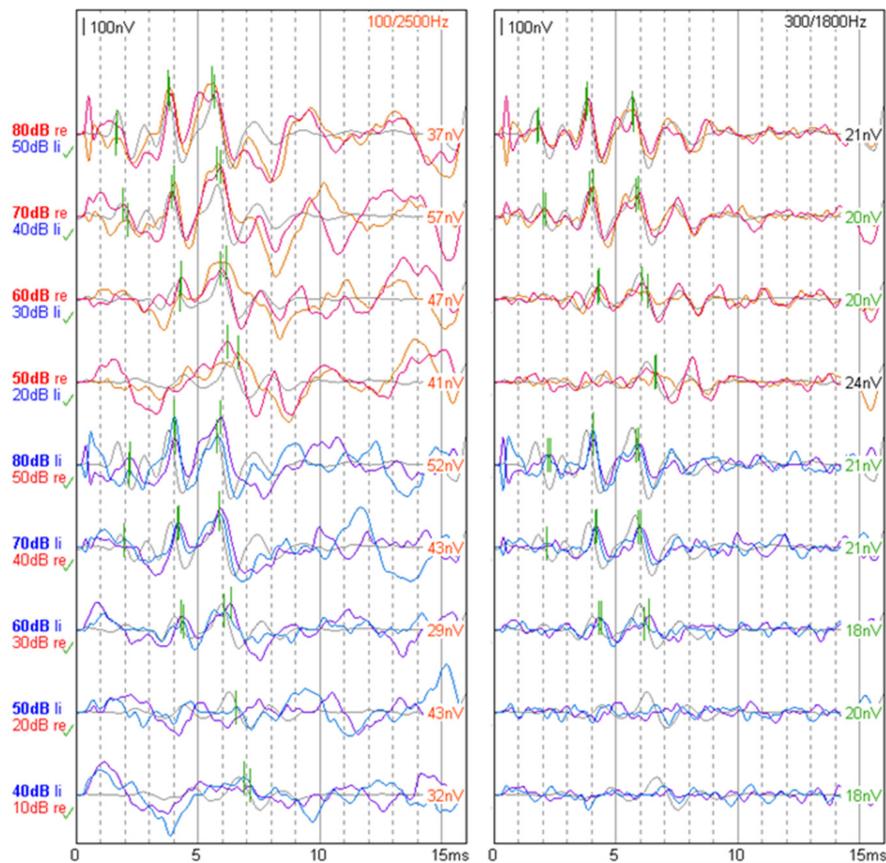


Figure 1: Frequency ranges for the total EEG and the individual groups of early, middle and late auditory evoked potentials (AEP); schematic drawing made according to real measurements.



Auditory brainstem responses (ABR) recorded from a 60-year-old female patient with (left pane) only analogue hardware filtering (100 to 2,500 Hz) and (right pane) after running through a combined recursive and non-recursive digital bandpass filter (300 to 1,800 Hz) designed for optimal compensation of phase effects. It can be seen that the peaks of the single components are more pronounced in the filtered version, whereas the responses elicited close to threshold are less clearly discernible after filtering. For every stimulus level, two partial averages are shown (in color), along with the traces representing the normal curves (gray). The numbers at the right border denote the residual noise, coded in colored letters as described at Figure 7.

Figure 2: Effect of digital filters on auditory brainstem responses (ABR).

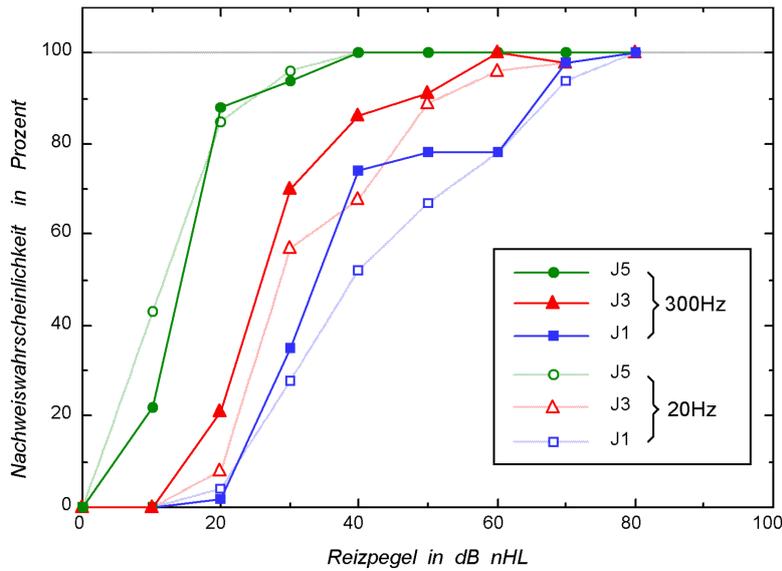
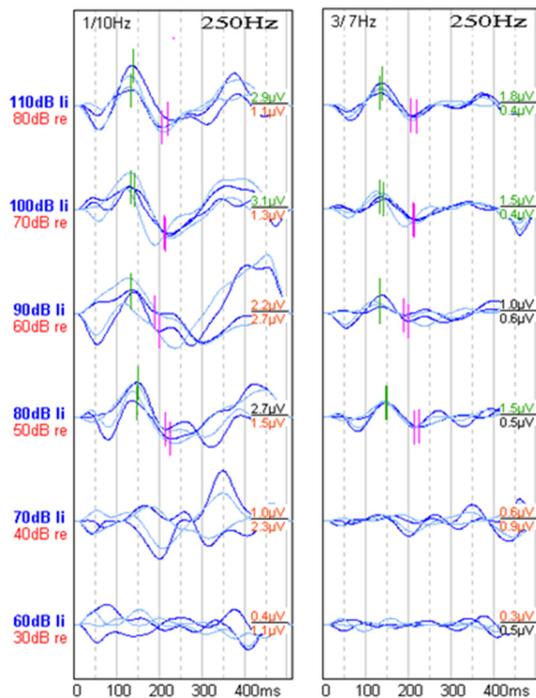


Figure 3: Effect of the application of a high pass filter with a frequency limit of 300 Hz on the detectability of the FAEP responses in comparison with a cut-off frequency of 20 Hz. The probability of response detection is higher for J1 and J3 at moderate supra threshold stimulus levels but lower for J5 close to threshold. The data are the results of the observation in 29 normal hearing subjects. [22]



At each stimulus level, two ipsilateral (intense color) and two contralateral (pale color) partial averages are recorded. The voltage complex N1–P2 (highlighted with green and violet marks respectively) is well discernible at suprathreshold stimulus levels. Bandpass filter is from 1 to 10 Hz (left pane) and from 3 to 7 Hz (right pane). The numbers at the left margin specify the level of stimulus (burst with 250 Hz of 500 ms duration) and contralateral masking. At the right border of each curve, the effective amplitudes of sum (upper number) and difference (denominator) of the partial averages are given in order to describe residual noise (green letters if lower than 500 μ V) and signal-to-noise ratio (green letters if above 6 dB).

Figure 4: Effect of digital filters on late or slow cortical auditory responses (SAEP).

Averaging of signal polarity

The conventional procedure for reconstructing the physiological response out of the background is based on (linear) averaging of the amplitude of stimulus-synchronously recorded signal segments. An alternative procedure, which has both limitations and merits, consists of averaging the polarity of the signal in place of its amplitude and interpreting the result using binomial statistics [27], [10]. The averaging of polarity is only at

first glance a “1-bit averaging”; in fact, each sample contributes to the summation result in the manifestation of three possible categories (“negative”, “zero”, and “positive”). The procedure was implemented in many laboratory instruments in the early days of ERA. Only in OAE screening devices it has become a widely used and essential part of signal evaluation in practice even today. The (indirect) contribution of polarity averaging to signal detection is based on its sensitivity to systematic deviations of a signal from the random process. For a randomly determined time signal (free of DC offset), each

individual sample has a positive or negative sign with equal probability. The addition of the sign or polarity over many sweeps will therefore statistically result in the expected value 0 at every point of the resulting curve. A deviation from zero which is significant according to the binomial statistics – which for a sufficiently large number of trials (averaging iterations) can be approximated by the Gaussian distribution – indicates that the underlying signal cannot be described by a random process.

The result “zero” in the polarity average means that the total signal composed of signal and noise was positive as often as negative in the totality of the sweeps (or exactly zero in *all* epochs). Due to DC voltages or signal components of low frequencies, which may still be present despite the high-pass filtering, the reference line can be different from zero. The (vertical) deviation from this line indicates the predominance of one of the two polarities. The special value of polarity averaging is that a perturbation of large amplitude affects amplitude averaging with the weight of its numerical magnitude, whereas in polarity averaging it contributes only a single binary unit. This makes polarity averaging very robust against perturbations compared to conventional amplitude averaging.

If the polarity average after n averages exceeds the limit [Formula 1]

$$S = c_p \cdot \sqrt{n}$$

– with $c_p=1.645$ for $p=0.05$, $c_p=2.326$ for $p=0.01$ and $c_p=3.090$ for $p=0.001$ – a signal significantly deviating from the random process is present at the time of the exceedance with probability $(1-p)$ [22]. If the original signal is strongly noisy, a given significance level is reached only after more averaging steps or at a later time. In contrast to the curve which shows the amplitude average of the EEG sections, the polarity average does not yield information about the size of the response, but about its significance.

If the polarity average is calculated only for one sample within the whole time-dependent signal, it cannot be excluded that its “deviation from chance” is just its property to be a zero point. Therefore, the method unfolds its full potential only if the *entire* acquired time range is included in the algorithm (see Figure 5). The construction of the reference curve is done during the data acquisition, its visual observation allows to judge the progress of the signal detection and to make a qualified decision about abort, continuation or termination of measurement – resembling the method based on the single point variance [7], [9].

Reproducibility and signal-to-noise ratio

In many statistical procedures, the rule applies that two parts contain more information than the ensemble composed of these parts. With respect to the averaging of stimulus-correlated signals of biologic origin, this means that the generation of one single curve emerging from all

individual responses does not allow an assessment of the reproducibility of the result. However, it requires neither instrumental nor programming effort to consistently organize the data acquisition into two partial averages $A(t)$ and $B(t)$ and to offer the option to calculate and display the overall mean $(A+B)/2$. The construction of $A(t)$ and $B(t)$ must be done during data acquisition by alternately summing the signal segments registered after the stimuli into one of two buffers. This “quasi-simultaneous” construction of partial averages is very helpful in deciding whether the response is to be judged as “real”, i.e. significantly protruding from the background (Figure 6).

According to [24], the reproducibility r (correlation coefficient calculated from the curves $A(t)$ and $B(t)$) is nearly equivalent to the signal-to-noise ratio q according to the almost exactly valid equation

[Formula 2]

$$r = \frac{q - 1}{q + 1}$$

with

[Formula 3]

$$q = \frac{\text{var}(s+n)}{\text{var}(n)}$$

where $(s+n)$ stands for the total signal composed of signal and residual noise and n for the residual noise determined from the (half) difference of the partial averages. The user is free to decide whether he prefers to consider the reproducibility or the signal-to-noise ratio. In both cases, a suitable time window must be selected for the calculation, which can optionally be shifted along the time axis in order to display the correlation differentially in time (see chapter below and [19], [16]).

The availability of two partial average time functions $A(t)$ and $B(t)$ further opens the useful option to calculate the cross power spectrum from real part Re and imaginary part Im of the spectra $A(f)$ and $B(f)$:

[Formula 4]

$$E(f) = Re[A(f)] \cdot B^*(f)$$

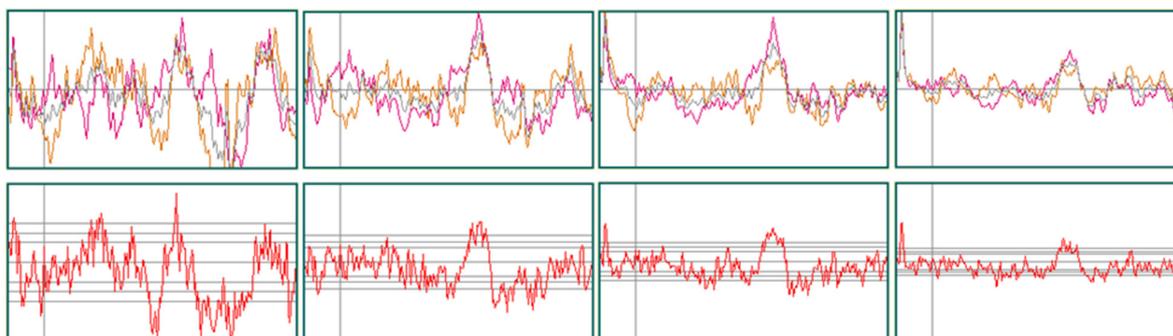
$$= Re[A(f)] \cdot Re[B(f)] + Im[A(f)] \cdot Im[B(f)]$$

which corresponds to the coherent spectral components common to the partial averages A and B , and the incoherent component

[Formula 5]

$$N(f) = FFT\left(\frac{A(t) - B(t)}{2}\right)$$

which reproduces the spectrum of the residual noise N (FFT = Fast Fourier Transformation). The separation of the coherent from the incoherent components is particularly useful for TEOAE and is also common practice there [26]. It seems appropriate to note that ipsi- and contralateral derivations of the AEP are not independent of each other to the same extent as two partial mean curves registered quasi-simultaneously under the same conditions. Their comparison is therefore only of limited use as a control of reproducibility.



For the amplitude average, the partial averages for condensation (red) and rarefaction (orange) as well as the total average (grey) are shown. For the polarity average, the limits for a significance of 95%, 99% and 99.9% are depicted as horizontal grey lines. The (horizontal) time axis runs from -2 ms to $+14$ ms (related to the start of the click stimulus). The vertical axes are equal for all panels in one row, i.e. not dependent on the number of averages.

Figure 5: Average of amplitude (upper row) and polarity (lower row) of click-evoked FAEPs (recorded from a normal hearing subject at $L = 20$ dB nHL), running from 500 (left) to 1,000, 2,000 and 4,000 (right) averages.

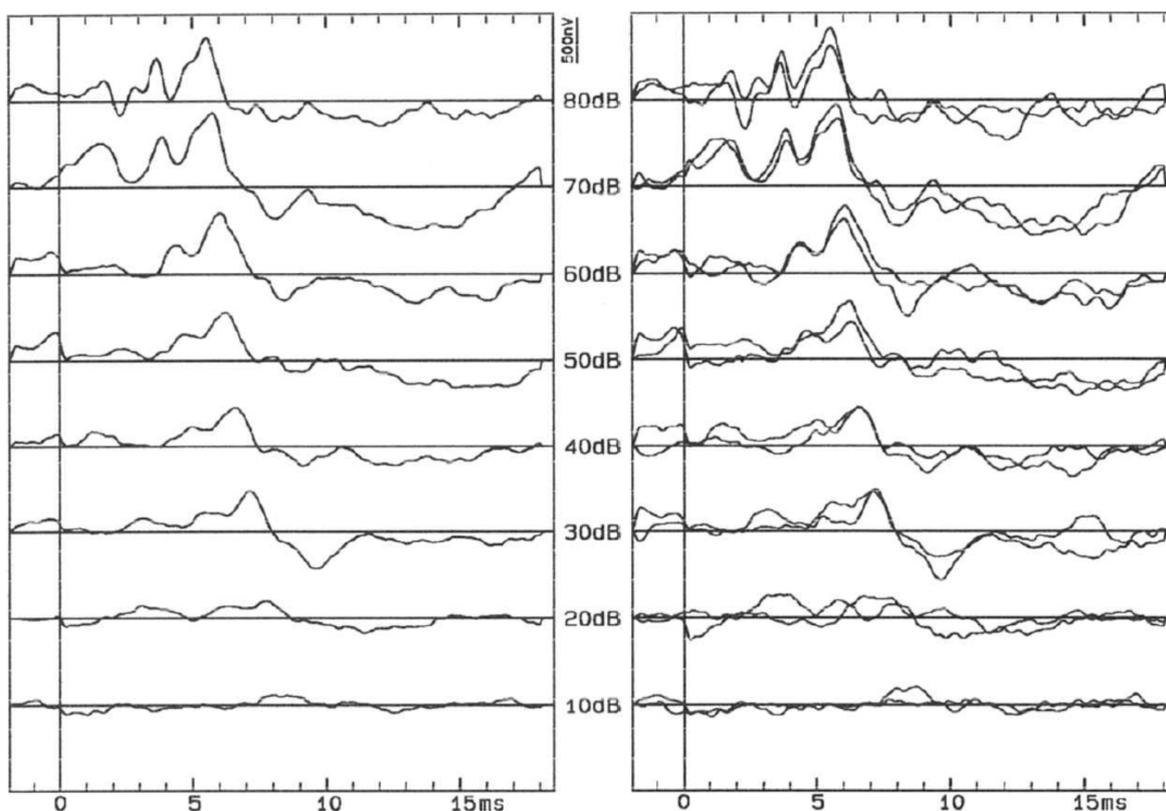


Figure 6: Total average $(A+B)/2$ (left panel) after 2,000 iterations and partial averages $A(t)$ and $B(t)$ after 1,000 iterations each (right panel). Click-evoked FAEPs of a normal hearing volunteer are shown (analog filter from 20 to 2,000 Hz, digital low pass filter with cut-off at 1,800 Hz). The two data sets represent the same recording. In contrast to the partial averages, the grand mean is not suitable for a well-founded assessment of the responses obtained at 20 and 10 dB HL. [22]

Quality measures

The detection of the physiological response hidden in the noise depends decisively on the extent to which it is possible to keep the residual noise as small as possible. When the stimulus intensity approaches the response threshold, the response disappears in the noise – the value zero is assumed only at even lower stimulus levels. Therefore, the detection threshold is always higher than

the hearing threshold; the greater the residual interference, the greater the distance between the two thresholds. In the limiting case of very large residual noise, no suprathreshold responses will be detectable. These considerations make clear that the residual noise σ is a suitable quantity for assessing the quality of AEP or OAE recordings. It is defined as the rms voltage of half the difference of the partial averages A and B : [Formula 6]

$$\sigma = u_{eff} = \sqrt{\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} \left[\frac{A(t) - B(t)}{2} \right]^2 dt}$$

$$= \sqrt{\frac{1}{i_2 - i_1} \sum_{i=i_1}^{i_2} \left(\frac{A_i - B_i}{2} \right)^2}$$

Here, the temporal integration extends over a suitable time window (or the summation over the time series of the assigned samples), e.g. from $t_1=2$ ms to $t_2=12$ ms for the FAEP. The rms voltage is equivalent to the standard deviation of the residual noise.

The standards DIN EN 60645-6 and 60645-7 [5], [6] require the display of an (estimated) quality measure. This requirement is met by many but not all commercial devices. Especially for FAEPs, the graphical combination of the residual noise with the amplitude growth function is useful to facilitate the identification of significant responses. A user-friendly, practical and clear implementation of this concept is shown in Figure 7.

The residual noise is a suitable measure for the formation of weighted averages (labeled “GMW” in the table of Figure 7) from parameters that have been determined several times from different recordings. For example, the differences of latencies, especially the cochleo-mesencephalic latency difference t_5-t_1 (“central conduction time”), are independent of the stimulus level to a good approximation, and therefore a mean value can be calculated from the individual values measured at different stimulus levels, where the individual values are usefully weighted by a factor proportional to the reciprocal of the respective residual noise. This ensures that the parameters derived from recordings with good quality have a greater impact on the final result. The weighting and the differentiation between reliable and less reliable values can be omitted if care is taken to ensure that the residual noise is constant across all recordings.

In combination with the amplitude of the response, the residual noise is the relevant measure for the significance of the signal detection. Therefore, it is useful to calculate this quantity continuously already during data acquisition, to display it to the examiner and to use it optionally as a stop criterion for the completion of the measurement. For transitory evoked OAEs (TEOAE), it has been demonstrated that expert identification of the response is successful in 87% of cases when the level of residual noise is below -1.5 dB SPL, but in only 71% of cases when this value is above $+1.5$ dB SPL [16], [15]. For the “numerical” signal detection based on reproducibility, the corresponding detection rates are 94% and 84%, respectively, each based on expert judgment. For the otoacoustic distortion products (DPOAE), qualitatively quite similar but quantitatively less critical rules are valid since the residual noise is much lower for metrological reasons [18].

Digital superposition

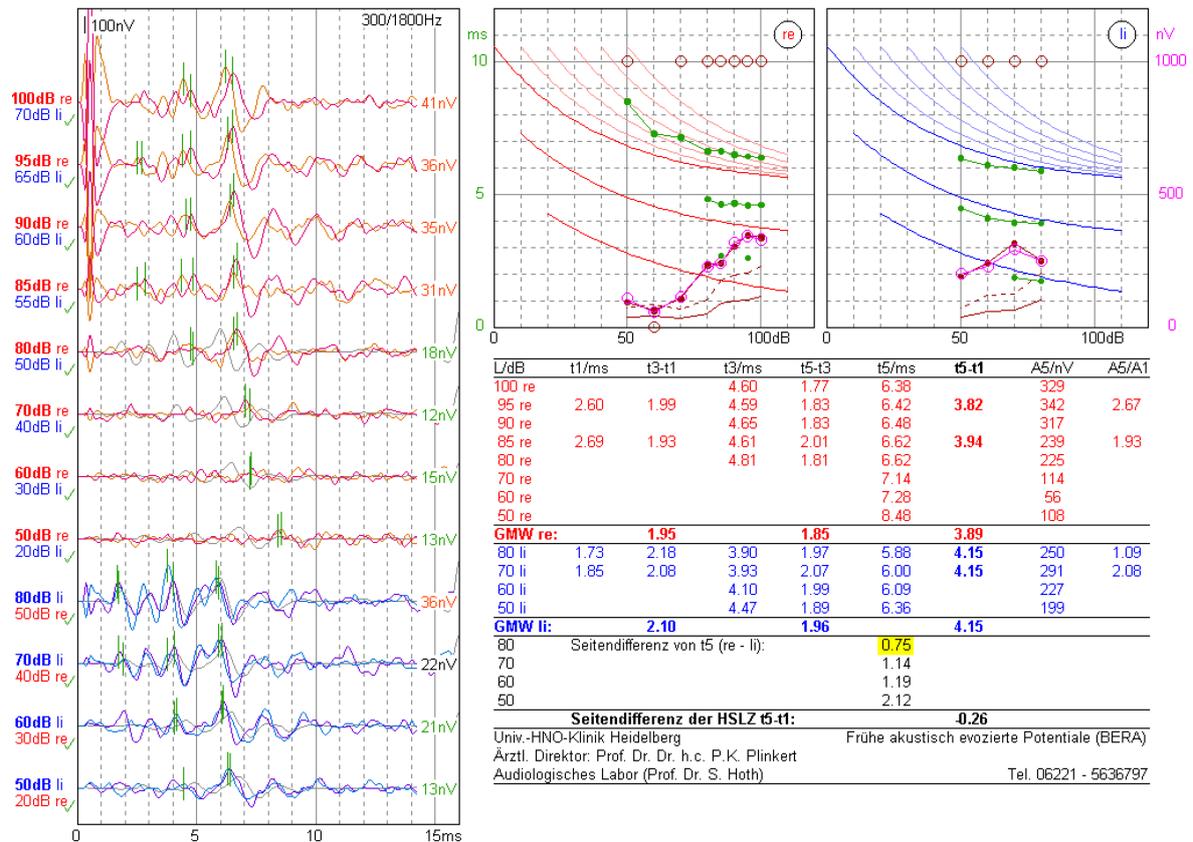
In electrical response audiometry (ERA), a large part of the effort to increase quality and reliability is aimed at objectifying signal detection and determining the response threshold. In addition, however, ERA is also used in clinic and practice for the purpose of obtaining evidence of maturation delays or of space-occupying lesions (such as a vestibular schwannoma or a vascular loop in the internal auditory canal or cerebellopontine angle). Obtaining answers to these questions is also based on the dichotomous and primarily qualitative utilization of the final recording with respect to the identification of the peak J5, but additionally for the detection of J1 and the most accurate quantitative measurement of the latencies of both components.

With rare exceptions, the amplitude of J1 is smaller than that of J5. In addition, the amplitude of J1 is affected to a much greater extent than that of J5 by high-frequency hearing loss, which is present in many of the suspected cases. Therefore, this component, indispensable for the determination of the cochleo-mesencephalic latency difference t_5-t_1 (“central conduction time”), is not clearly discernible in many cases. The scheme of later offline processing of the data described below serves to highlight more clearly the peak J1 without additional measurements.

Typically, two or more click-evoked FAEP recordings gained at different stimulus levels with a component J5 identified as significant response are available as starting material for the application of the method. Among these, the investigator makes a selection for further processing using a special software module, if available. The software corrects for the level dependence of the latencies by shifting each of the selected traces along the time axis according to the respective value of t_5 . Since the latency difference t_5-t_1 does not depend on the stimulus level to a good approximation [22], this measure also synchronizes J1. An average curve is calculated from the shifted curves and displayed (“aligned overlay”). The intended and achieved compensation of the level-dependent latency t_5 in a systematic way follows the same principle as the elimination of the frequency dependence in the formation of the “stacked derived-band response” [8]. From the resulting “latency-corrected digital superposition”, the desired latency difference t_5-t_1 can be determined (Figure 8).

In many cases, the outcome of the latency-compensated superposition consists in a clear reconstruction of all peaks from J1 to J5, which allows an unambiguous determination of the diagnostically important latency difference t_5-t_1 . If this is not successful, the examiner formulates the statement “central conduction time cannot be determined” on a much better basis than without this digital evaluation aid.

According to general rules, processing several curves improves the signal-to-noise ratio, e.g. by a factor of two with four curves of comparable amplitude. Alternatively, the number of averages could be increased by the same



The residual noise calculated in the latency range from 2 to 12 ms is given numerically in nanovolts at the right border of each trace in green letters if less than 21 nV, in red letters if greater than 29 nV and in black letters if between these limits (33-percentiles deduced from a large clinical population). The characteristic diagram (right top) shows the residual noise (multiplied with $\sqrt{2}$), which according to [14] is identical with the error of the amplitude differences (single standard deviation as continuous brown line, twofold standard deviation as dashed brown line). Accordingly, the amplitudes of significant responses (hollow and filled violet circles) lie above the dashed line. According to this assessment, the questionable response which could be guessed in the 60-dB recording from the right ear is not classified as significant and the corresponding hollow brown circle is positioned at the lower edge in the characteristic diagram, in contrast to all other circles positioned at high level ($y=10$ ms resp. 1,000 nV) indicating a positive response detection.

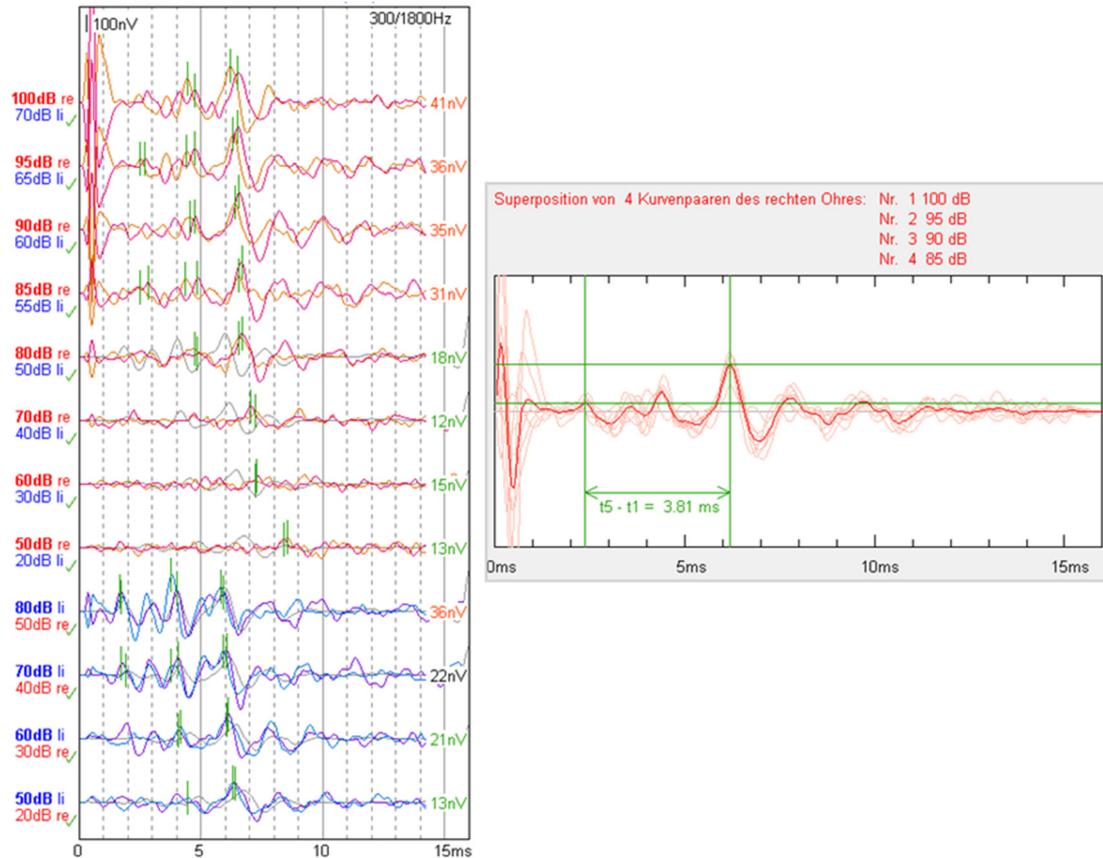
Figure 7: Complete documentation of an ABR-recording containing the primary recordings (left), the level dependence of latency and amplitude (right top) and the numeric results (right bottom).

factor, but this results in a correspondingly higher burden on the patient and prolongation of the examination time. The practical benefit of digital superposition cannot be described quantitatively in empirical figures. However, in the author's clinical work it has proven to be a tool used almost daily and highly appreciated for its usefulness.

RMS amplitude as an alternative to amplitude difference

There are good reasons for taking a closer look at a parameter of the FAEPs that is generally rather neglected, namely the amplitude A5 of component J5. In general, it is reasonable to assume that the magnitude of a physiological response is closely related to the strength of the stimulus and to the functionality and vitality of the stimulated biological structure and at best even diagnos-

tically useful. In this context, in view of the inevitable presence of noise, the question of an unambiguous and robust definition of the target quantity deserves attention. Conventionally, A5 is obtained from the difference of two amplitude values, mostly maximum and minimum (Figure 9). This quantity corresponds to the vertical extent of the peak. The use of this relatively simple linear amplitude difference $A5_{lin}$ is not the result of an optimization in terms of accuracy or robustness, but rather due to the fact that its calculation requires little effort. In technical signal processing, however, it is common practice to consider the rms amplitude rather than the difference of two individual values for the quantitative description of time-dependent signals, especially in the case of irregular signal characteristics. One of the reasons for this is that the effective amplitude is more likely to reflect the relevant dimension, namely the physical power contained in the signal. This is the case with the linear ampli-



The curve resulting from this superposition is shown in intense red color, while the pale lines represent the single constituents (right frame). In the curve resulting from the superposition, the peak J1 is clearly pronounced and unambiguously discernible, whereas it can only be guessed in the original recordings. From the latency-compensated superposition, a value of 3.81 ms is deduced for the central conduction time $t_5 - t_1$. The data are the same as those shown in Figure 7.

Figure 8: Digital superposition of four curves (recorded from the right ear at 85 to 100 dB HL, left frame) after correction of the effect of stimulus level on latency t_5 ("aligned overlay").

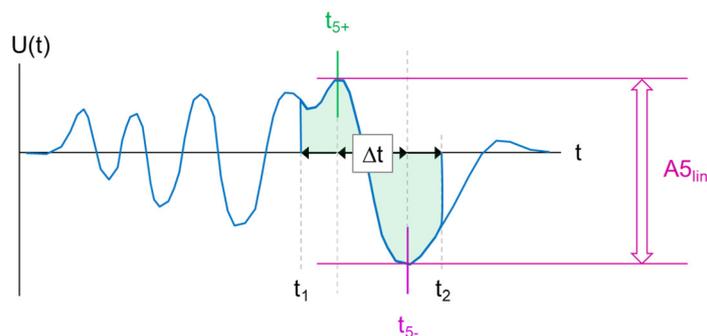


Figure 9: Principle of computation of the effective amplitude $A5_{eff}$ based on the coordinates of peak and trough supplied by the examiner. The linear amplitude difference $A5_{lin}$ is shown for comparison.

tude difference if and only if all changes of the component J5 caused by e.g. stimulus parameters or pathological processes are purely linear scale transformations (without deformations or distortions such as flattening).

The rms amplitude $A5_{eff}$ is defined as the root of the continuously or discretely calculated mean square deviation of the measured voltage $U(t)$ from its mean value \bar{U} : [Formula 7]

$$A5_{eff} = \sqrt{\frac{1}{t_2 - t_1} \int_{t_1}^{t_2} [U(t) - \bar{U}]^2 dt}$$

$$= \sqrt{\frac{1}{i_2 - i_1} \sum_{i=i_1}^{i_2} (U_i - \bar{U})^2}$$

The quantity $U(t)$ is the mean value the EEG voltage calculated from the partial mean values $A(t)$ and $B(t)$, \bar{U} is the mean amplitude of this curve in the time interval of integration or summation, which extends from $t_1 = t_{s+} - \Delta t/2$ to $t_2 = t_{s-} + \Delta t/2$; position and duration of this time interval are derived from the latencies of maximum and minimum (see Figure 9).

The calculation of the effective amplitude $A5_{\text{eff}}$ does not involve any additional effort for the examiner. The only intervention consists in the anyway usual identification of maximum and minimum of the component J5. From the latencies t_{s+} and t_{s-} thus determined, the time range of integration or summation given by the double horizontal distance Δt between peak and trough is obtained. With this choice of the time window the coverage of the entire relevant signal section is ensured.

The rms amplitude is not the area enclosed by the potential curve and the zero line, but the root mean square (RMS) of the time-dependent amplitude. It is therefore a statistically defined quantity which, except for the time window, is defined exactly similar to the residual noise of the recording as defined above (and can therefore be directly compared with it). In contrast to the linear amplitude difference, this parameter arising from much more than just two samples is less susceptible to variations in the background than either of its components and is thus a more accurate measure of signal strength than the linear amplitude difference.

The rms amplitude can be used as a signal descriptor even if no response is present. For this purpose, in registrations derived with stimulus levels below the response threshold, the time window for the calculation of $A5_{\text{eff}}$ is determined manually or automatically from the extrapolation of the exponential latency characteristic [12]. At low stimulus levels, the rms amplitude is distributed randomly around the background level; at the response threshold, its value increases systematically with increasing stimulus level L (Figure 10). A kink in the amplitude function $A5_{\text{eff}}(L)$ marks the response threshold. It is close to triviality to state that for a reliable determination of the threshold, measured values below as well as above the threshold must be available [13]. In addition to the kink in the graph identified visually by the investigator, a numerical criterion is determinative of responses significantly exceeding the noise level. The condition is that $A5_{\text{eff}}$ is at least one standard deviation larger than the residual noise σ identical to this standard deviation [14]. The concept of effective amplitude opens up the possibility of adding a useful option to electrical response audiometry. Through its application, the classification as "clear response" is not solely "in the eye of the beholder", but is based on a precisely defined numerical criterion. This leads to a considerable increase in the reliability of the objective response threshold.

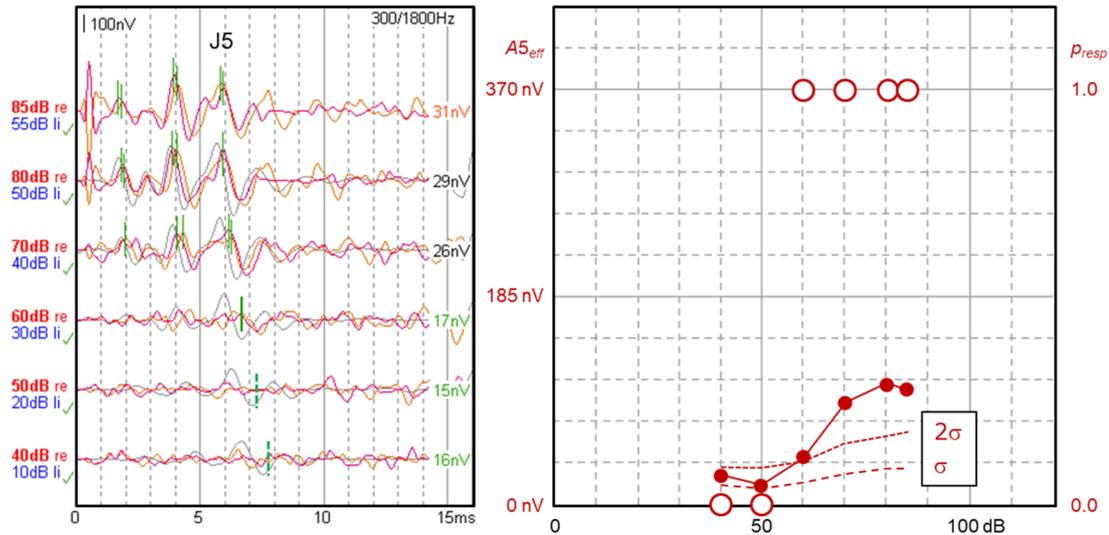
Time differential correlation analysis (gliding reproducibility)

Most of the physiological responses registered in objective audiometry are only present in a limited fraction of the time window. Nevertheless, the correlation coefficient or signal-to-noise ratio (often referred to as "repro") is usually calculated from the data in the *entire* time window. This can lead to a situation where there is little or no correlation between the result of this calculation and the response. In the worst case, a response can be overlooked, but in any case, considering integral reproducibility alone at least does not exhaust the potential of the parameter. However, with the help of the local, temporally differential evaluation of individual sections of the time window, even responses that are limited in time and variable in their position (latency) do not escape detection.

Figure 11 shows the result of a TEOAE measurement which, according to the visual evaluation, contains a clear physiological response. However, since the duration of this response is limited to the first part of the time window, the integrally calculated reproducibility is just 48.2%. According to the usual rules, this leads to the classification as "OAE-negative", although the limit of 60% is consistently exceeded in the interval 3 to 9 ms – with a peak value of 94.9%. The gliding time-differential calculation of the correlation coefficient (gliding reproducibility) enables the unambiguous OAE detection [16].

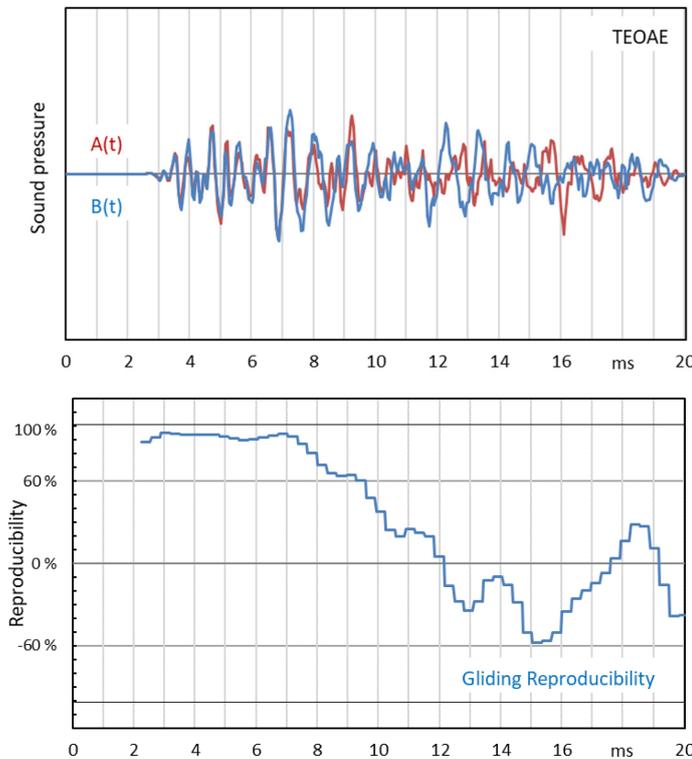
For the evaluation by the user, the quantitative graphical representation of the reproducibility in an additional coordinate system is not optimal. More intuitive and more directly graspable for the eye of the observer in comparison to the two-dimensional graph is the coding shown in Figure 12 in which the local correlation coefficient is presented in graded shades of gray [19]. The degree of blackening varies from white (correlation coefficient $r \leq 0\%$) to black ($r = 100\%$). Since the reproducibility as a correlation coefficient is a statistical quantity calculated from many numerical values, the shorter the underlying time window, the more it underlies the influence of chance. This explains that large contrasts can occur in the degree of shading of neighboring fields.

The example shown in Figure 12 makes clear that detectable and normal OAE are not the same and that the distinction between the two is definitely possible and also clinically meaningful. The global reproducibility is the basis of the purely dichotomous signal detection, thus it does not say more than "signal detection was successful". On the other hand, its temporally differential observation allows differentiated statements about deviations from the normal appearance of the response and thus about possible pathological causes. Also in the frequency domain, the differential consideration of the spectrum and the coding of the local signal-to-noise ratio in a gray scale extends the information yield. Visual detection of empty time regions or frequency bands, aided by graphical pro-



Significant responses are identified by comparison of $A_{5_{eff}}$ (small filled circles) with the residual noise σ (dashed lines). The threshold is reached if the effective amplitude exceeds two standard deviations of the background ($A_{5_{eff}} \geq 2\sigma$) – visualized by an indicator p_{resp} and its transition from 0 to 1 (large hollow circles). The latencies are determined by the examiner at suprathreshold levels (0 to 85 dB) and extrapolated by the software below the threshold (50 and 40 dB).

Figure 10: Series of click evoked FAEP recordings (left panel) and level dependence of the effective amplitude $A_{5_{eff}}$ (right panel) extracted from the original recordings.



In spite of a clear response of short latency, the integral reproducibility deduced from the whole curve amounts to only 48.2%. The time differential correlation analysis (lower diagram) shows that the reproducibility fulfills the criterion of significance only within the time range from 3 to 9 ms, whereas the correlation of the signal components lying beyond 10 ms is lower than 60% and hence not significant. The intervals for the correlation coefficient were 1.28 ms (32 samples), gliding along the time axis in consecutive steps of 0.32 ms (8 samples).

Figure 11: Transitory evoked otoacoustic emissions (TEOAEs) elicited by click stimuli and represented by the two partial averages $A(t)$ and $B(t)$ (upper diagram).

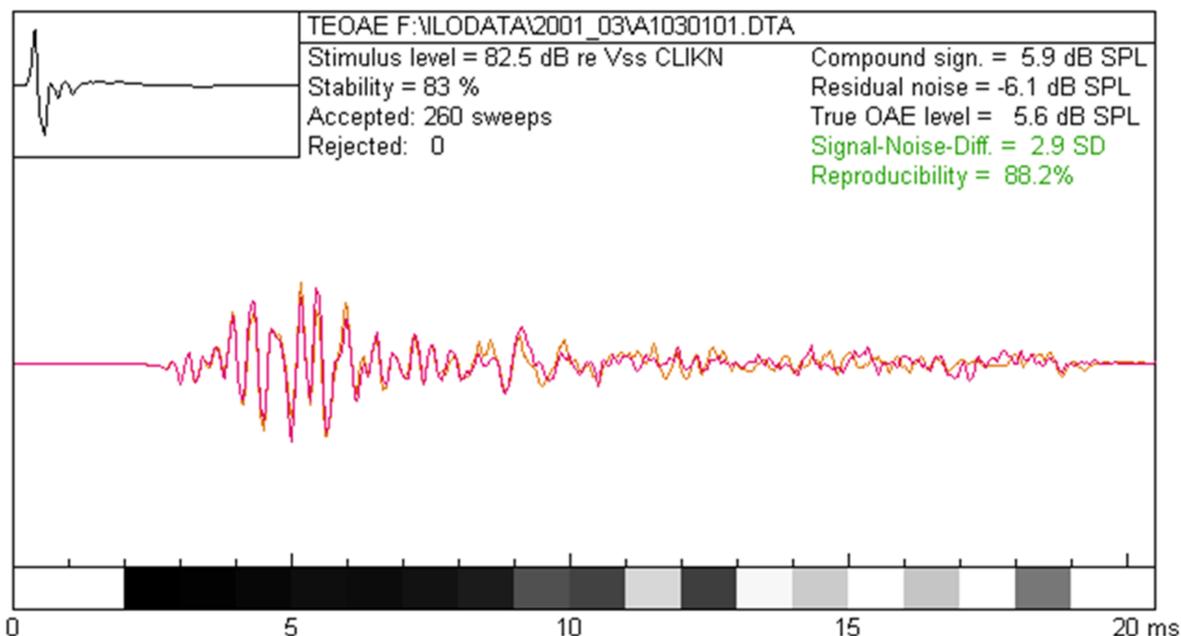


Figure 12: TEOAEs elicited by click stimuli in an ear with low frequency hearing loss. The reproducibility of the two partial averages is displayed in a time differential manner by the bar at the lower border, consisting of areas whose intensity is deduced from the local correlation coefficient (gray scale reaching from white for reproducibility $\leq 0\%$ to black for $r=100\%$). The duration of the underlying time slices is 1 ms (25 samples).

cessing, facilitates the derivation of clues to the frequencies affected by possible hearing loss.

Among the physiological signals used in objective audiometry, TEOAEs occupy a special position in that the response normally extends over the entire duration of the registered time window. In contrast, for transient AEPs (e.g., FAEPs and SAEPs), the response occupies only a limited portion of the time window and its location within this window is variable because the latency depends on stimulus parameters, physiological conditions, and pathological changes. It is in this situation that the display of temporally differential correlation, similar to the polarity average, contributes to the identification of the response and to the assessment of its significance (Figure 13).

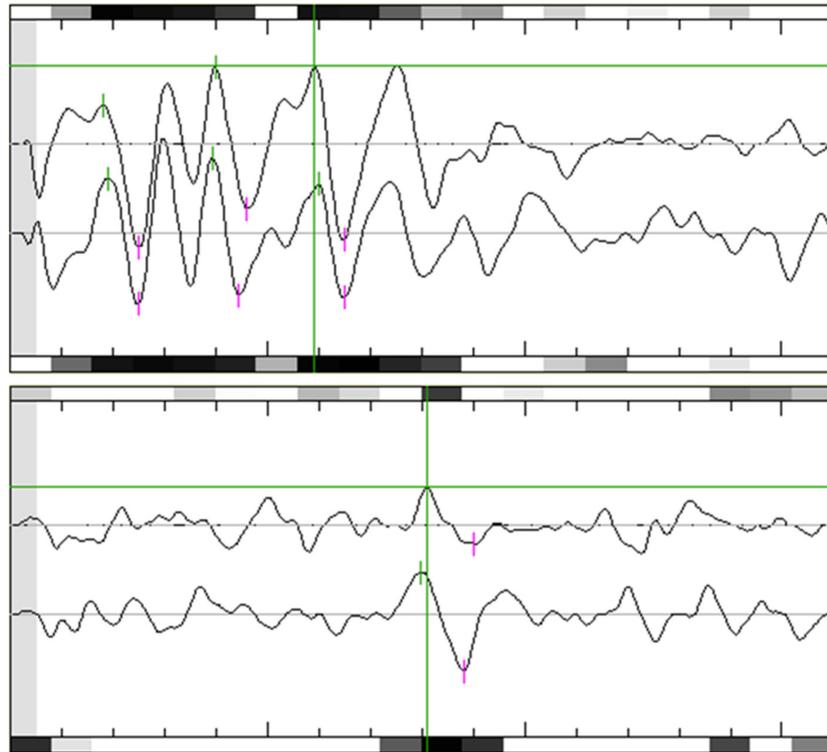
The correlation considered in this and the previous sections emerges from the amplitude of the curves involved and their time course. In addition, the correlation of the slope of the curves can be determined from the first time derivative of the curves. The interest in this arises from the fact that the latencies of the individual response components, which are so important for diagnosis, are characterized as local maxima and minima by a horizontal tangent. Thus, the slope must be equal to zero in both partial average curves. This equality is accompanied by an intensive blackening of the gray scale display (Figure 13). If the maxima are offset in time from each other, the display is more gray than black and the latency determination is less reliable.

Amplitude histograms

Detecting the signal against the background of residual noise is the key to the use of OAEs and AEPs. As tools in accomplishing this task, all features that can help to distinguish between signal and noise are useful. These features include, first, the elementary parameters of frequency, phase, and amplitude, by which all time-dependent processes are fully described. However, each of these quantities not only has an instantaneous value, which is included, for example, in the polarity average and the correlation coefficient, but also underlies a statistical distribution, which is characteristic and different for deterministic and stochastic processes. These frequency distribution densities or histograms of the mentioned parameters are specific for the process underlying the generation of the signal. In general, signal and noise will have different distribution densities. When signal and noise occur as a mixture, their specific distribution functions are overlaid to an integral histogram.

The statistical distribution density contains the answer to the question, with which frequency the individual values of the considered parameter occur. Among the parameters whose histograms could be useful, only amplitude is considered here, although frequency and phase spectra also play a role in statistically based signal detection [3], [2].

The amplitude distribution of a stochastic process is known to have the shape of a Gaussian curve: The value zero occurs most frequently, and with increasing amplitude the probability of occurrence becomes smaller and smaller. The amplitude distribution of a harmonic oscillation is less popular: Because the sine wave is flat at its



The gray scale covers the range from white (for correlation coefficients at or below 0%) to black (for 100%). In both diagrams, the bar at the lower border refers to the amplitude of the traces, whereas the upper border is derived from their slope. The duration of the gliding time window is 0.8 ms (16 samples).

Figure 13: Gliding reproducibility for the visualization of the correlation between the two partial averages of ABRs recorded at click levels of 80 dB nHL (upper frame) and 30 dB nHL (lower frame).

extreme values and steep in between, it has the shape of a trough with two poles (infinities or singularities). In graphical terms, this is due to the fact that the vertical probability density of the system is given by the inverse of the slope, if the samples of electric voltage are equidistantly distributed in the horizontal dimension (time axis). This observation makes clear that the amplitude distribution density of the additive superposition of a stochastic process and a sinusoidal signal differs in a characteristic way from the bell-shaped curve characterizing the random distribution by peaks at large absolute values of the amplitude (Figure 14). The height of these lines at the margins of the spectrum depends on the signal-to-noise ratio; in the case of isolated residual noise, the peaks disappear altogether.

In clinical practice, the observation of the amplitude histograms has proved particularly useful for the N1-P2 complex of the SAEPs, since here an evoked signal is present that resembles a pure harmonic oscillation (corresponding to the time interval of 100 ms between maximum N1 and minimum P2, the frequency of this oscillation is approx. 5 Hz). The comparison of the frequency distribution of the sum of signal and noise with that of the isolated residual noise makes the presence of a response convincingly clear (Figure 15). Beyond a single prototype realized so far in the author's laboratory, optimization of this tool is certainly still possible, for example with respect to the length of the analyzed time window.

With increasing length, the influence of the noise increases and the difference in the amplitude histograms loses clarity.

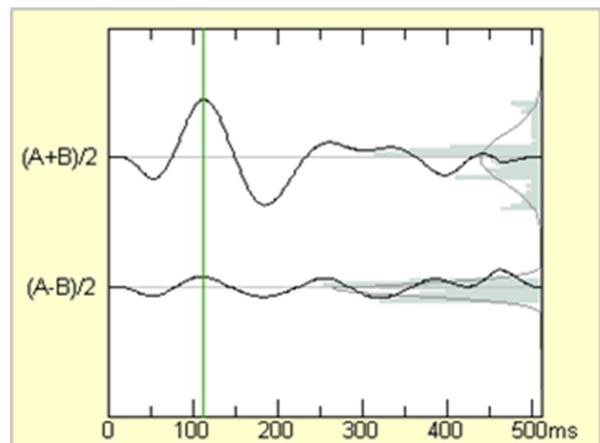


Figure 15: Cortical response N1-P2 and the amplitude density distributions of composite signal (superposition of evoked potential and residual noise, above) and isolated residual noise (below). The amplitude histograms are the shaded gray areas at the right border (with the amplitude axis directed upwards and the relative frequency of occurrence towards the left).

At present, the practical use of the amplitude histograms is limited to the observation and evaluation of their graphical representation; a quantitative description of

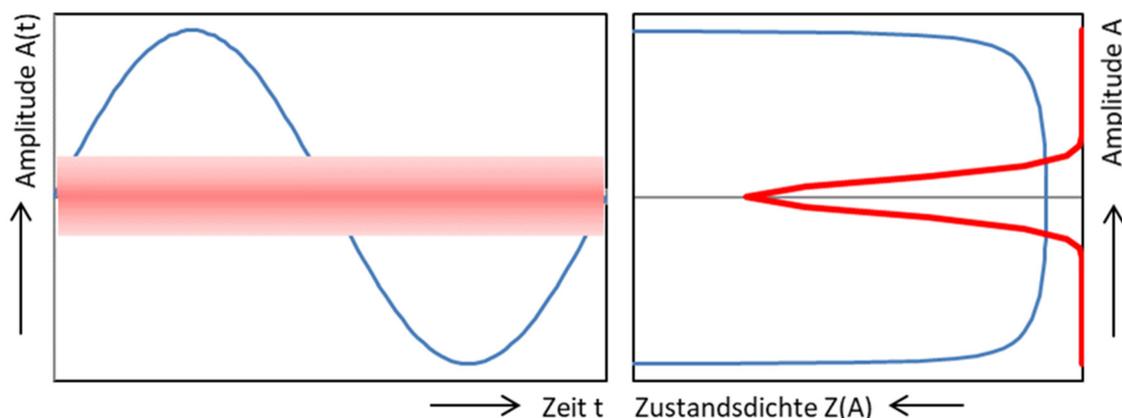


Figure 14: Schematic illustration of the effect of the superposition of a harmonic oscillation (blue) and a stochastic signal (red) on the amplitude distribution; left: time dependent oscillograms (stochastic signal only outlined by the range of values), right: amplitude histograms of both signal components whose summation results in a gaussian distribution with additional offset and satellite peaks.

the benefit for diagnostic certainty is not possible. Of course, the deviation of the positive response from the normal distribution which is typical for pure noise, can be described with the help of statistical tests; however, the visualization rather meets the needs of the users, who are supported in this way by a machine and without additional effort in judging numerical parameters.

Documentation

For some of the commercially available devices for measuring OAEs or AEPs that are in practical use, the final presentation of the examination results could still be improved in terms of expediency and ergonomics. It is not expedient and even less ergonomic to follow the precept of completeness, which may be quite appropriate in many other areas of life. Even if it is undoubtedly correct to document all parameters concerning the stimulation, the signal acquisition, the measurement conditions and the results, the completeness of the information on the (hard copy) report intended for the patient's file or for passing on to colleagues is rather disadvantageous than purposeful. A rationally determined and purpose-oriented documentation should be limited to the information with diagnostic or therapeutic consequences. The amplification factor or the filter limits are certainly not among them; since their setting values are changed only for rare special questions, they need not be indicated on the daily report. Just like the overloading with irrelevant parameters, the omission of substantial data is one of the frequently encountered deficiencies. The deepness of anaesthesia, which may be fundamentally important for interpreting the measurement result, and a quantity describing the residual noise, which is indispensable according to the DIN EN 60645-6 and 60645-7 standards [5], [6], cannot be found at all in many measurement systems, or only after a lengthy search and only on the screen. Avoiding these deficits was the goal of the development of a soft-

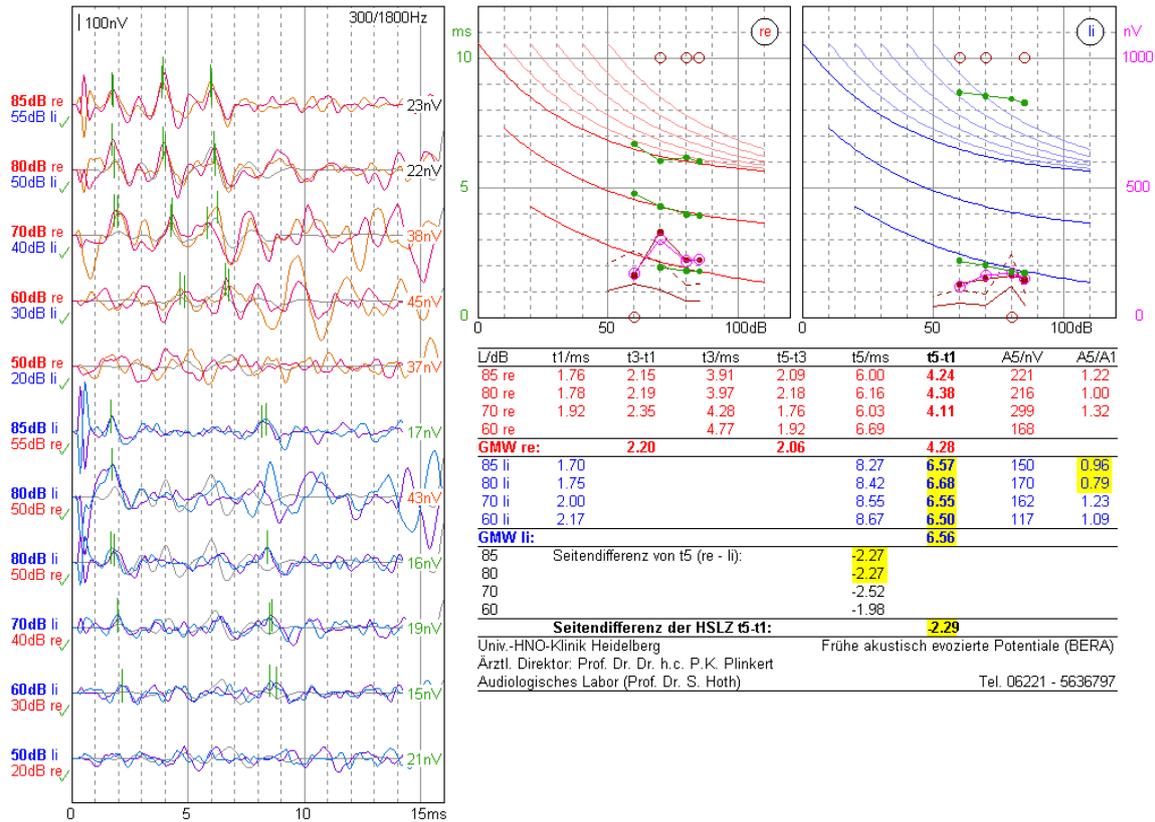
ware environment at the Audiology Laboratory of the Heidelberg University ENT Clinic, with the help of which the results of OAE and AEP measurements can be displayed completely and compactly at the same time. For the FAEPs, the compilation of graphical displays and numerical data, which has been optimized over decades, is presented here in more detail (Figure 16). The advantage for the viewer is the presentation of primary and secondary examination results at a glance – without the need to turn pages or to jump between different screen windows.

With special emphasis we would like to advocate the display of two partial averages for each take. In the usually shown total average, a peak can practically always be found at the points where the evaluator looks for it. Only by consideration of the partial averages a decision about the reliability of this maximum is possible. If necessary, the two parts can be fused with ease to form the total average, but the benefit is merely an aesthetic enhancement – at the cost of a loss of information.

In principle, all components of the documentation can already be displayed during data acquisition. After completion of the measurement series, evaluations for finding the response threshold can follow, as described in more detail in the next section.

Threshold determination

Many of the procedures and tools described in this review aim to reliably identify the response and thus contributing to the determination of the response threshold. The audiological interest in the response threshold – i.e. the lowest stimulus level at which a response is detectable – is based on the close relationship between this quantity and the hearing threshold as the target quantity actually relevant for hearing diagnosis. Hearing threshold and response threshold are not identical, but they can be very close to each other. The hearing threshold is generally



The panel with the primary results (left panel) contains the gender and level specific normal traces (gray). From the comparison of the partial averages it becomes clear that the component J5 derived from the left side is not unequivocal at all stimulus levels. Each of the traces comes along with two important data: the level of contralateral masking (at the left border directly below the stimulus level) and the residual noise (at the right end of each curve). The two square diagrams (right top) show the level dependence of the response latency times t1, t3 and t5 (green), the amplitude A5 (violet), the normalized effective response amplitude A5_{eff} (brown), the residual noise (brown line) and an indicator for the significance of the response (brown hollow circles). In order to support the interpretation of measured latencies, the normal functions of t1(L), t3(L) and t5(L) (solid lines, based on the age and sex specific normal values given in [22]) and the latencies of t5(L) expected for conductive hearing loss of 10 to 50 dB (thin lines) are included. The table with latencies and differences thereof (right bottom) contains the (weighted) mean values (GMW) of quantities which do not depend on the stimulus level. The data for the central conduction time (HSLZ) t5-t1 as well as the interaural differences of t5 and t5-t1 are highlighted if they lie aside of the normal range.

Figure 16: Example for the comprehensive documentation of click-evoked FAEPs in the case of a vestibular schwannoma on the left side.

below the response threshold, because the auditory system is definitely activated if a response has been detected.

The distance between auditory and response thresholds is variable and depends on many factors (Figure 17). The starting point for a closer look is the undoubtedly correct observation that the disappearance of the response is never observed, but only its sinking into the noise. To be recognized with sufficient certainty as a significant signal, the (effective) amplitude of the response must exceed the (effective) amplitude of the residual noise by a specified minimum amount. In signal processing, it is common practice to require a margin of 6 dB. This amount is distinguished by the fact that it corresponds to a factor of two of the amplitudes of signal and noise – or somewhat more precisely: the effective amplitude N of the

noise increased by the signal amplitude S is twice as large as N alone, or $(S+N) > 2N$.

Responses whose amplitude falls below the mentioned limit escape detection. At stimulus levels above the response threshold L_1 , the amplitude of the signal increases, below which it continues to decrease until it reaches zero. The zero crossing L_0 can be determined approximately by (linear) extrapolation of the suprathreshold measured amplitude characteristic or growth function (AGF). Its (horizontal) distance ΔL_1 from the response threshold depends on the strength of the residual noise, on the absolute amplitude of the response, on the slope of its growth function, and on the nature of any pathological event that may be present (e.g., recruitment or loss of neuronal synchronization). Also, the stimulus level associated with L_0 is generally still above the (physiological) auditory threshold L_{HS} . The distance between the

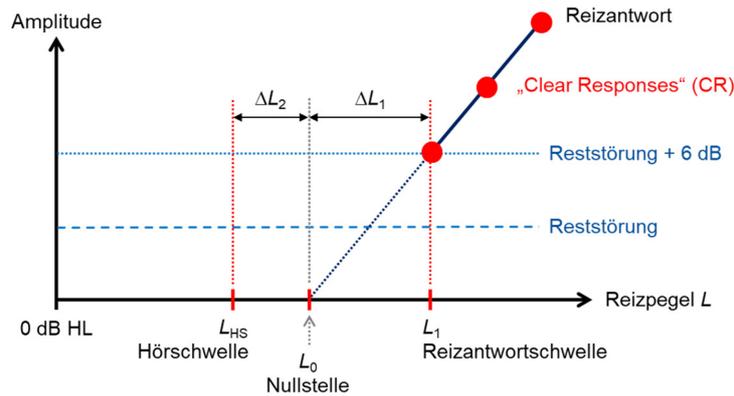


Figure 17: Illustration of the relation between hearing threshold L_{HS} and response threshold L_1 . The linear extrapolation of the amplitude growth function (AGF) yields its zero crossing L_0 , which is in general different from the hearing threshold L_{HS} (i.e. $\Delta L_2 \neq 0$). The difference ΔL_1 can be reduced by minimizing the residual noise. This does not hold for the difference ΔL_2 , which can depend on the type and origin of the signal to be detected, on the detection method and eventually on underlying pathologic processes.

thresholds L_{HS} and L_0 depends, among other things, on the type and origin of the signal to be detected (e.g., it is very large in the case of the stapedius reflex).

The impact of noise amplitude, slope of the growth function, and number of averages on the response threshold can be described by an exact mathematical expression using simple reasoning and based on realistic assumptions:

[Formula 8]

$$L_1 = L_0 + \sqrt{3} \cdot \frac{A_N(m=1)}{g} \cdot \frac{1}{\sqrt{m}}$$

Here,

- L_1 designates the response threshold,
- L_0 is the zero crossing of the amplitude growth function,
- $A_N(m=1)$ designates the (effective) amplitude of the unaveraged noise,
- g is the slope of the (linearly approximated) amplitude growth function (for FAEPs typically 10 nV/dB) and
- m is the number of averages.

If only the effect of the number m of averages is considered in the given equation, it can be seen that the response threshold L_1 approaches the zero crossing L_0 with increasing accuracy. However, the two quantities agree with each other only asymptotically (i.e., for $m \rightarrow \infty$).

The other parameters appearing in the equation have the following effect (Figure 18):

- The stronger the (unaveraged) noise $A_N(m=1)$, the larger the discrepancy between L_1 and L_0 .
- The steeper the amplitude characteristic curve (large slope g), the smaller this deviation.

Under unfavorable but realistic conditions, the supposed response threshold L_1 may deviate from the real response threshold L_0 by up to 30 dB. In practice, the reliability of threshold estimation can be estimated by graphically displaying the obtained data set several times after completion of the measurement series (consisting of several derivations performed at different stimulus levels),

for example for the averaging numbers m , $m/2$, $m/4$, and $m/8$ (this assumes that either all individual sweeps or at least the intermediate averages were stored during data acquisition). The investigator then determines the visual response threshold from each of the four series, compares the values with each other, and derives an estimate for the “true threshold”.

According to theory, the difference $L_1 - L_0$ between the supposed and real threshold decreases by a factor of $\sqrt{2}$ if the number of averages is doubled:

[Formula 9]

$$\frac{L_1(m) - L_0}{L_1(2m) - L_0} = \sqrt{2}$$

This equation can be solved for L_0 and yields the following expression for the “true threshold”:

[Formula 10]

$$L_0 = \frac{\sqrt{2} \cdot L_1(2m) - L_1(m)}{\sqrt{2} - 1}$$

For example, if the observed response threshold L_1 drops from 40 to 30 dB when the averaging number is doubled, then the true response threshold L_0 is 6 dB.

Unfortunately, no statement can be made about the practical usefulness of the described approach, since no measuring device with a corresponding implementation exists yet. It may be assumed that the described rules are only approximately valid in practice for at least two reasons: First, the considerations are based on the assumption of a steady-state EEG noise, which only a few patients are willing and able to produce, and second, the stimulus levels within a measurement series are usually varied in a grid with steps of 10 dB, which limits the accuracy of the results. In general, the relevance of the threshold correction decreases as the slope of the amplitude growth function increases. For very efficient stimuli like the CE chirp, it is probably dispensable.

Correcting the observed response threshold for the influence of residual noise does not relieve the investigator of the necessity to derive the hearing threshold from the

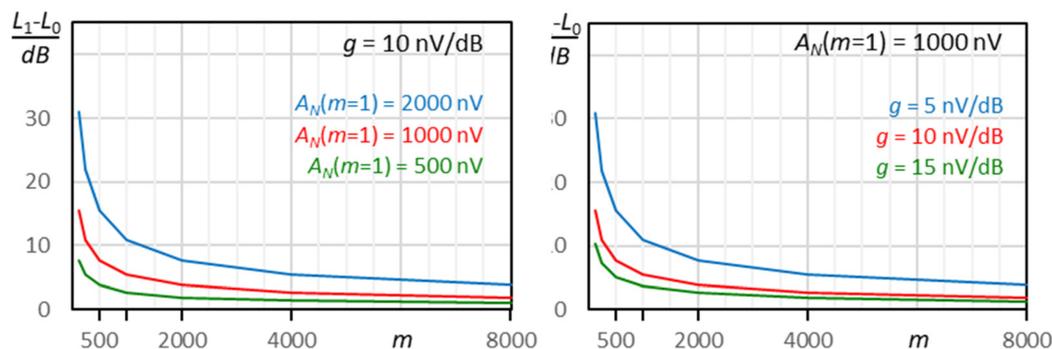


Figure 18: Development of the distance $L_1 - L_0$ between estimated and true response threshold with increasing number m of averages. The left panel shows the situation for variable noise amplitude $A_N(m=1)$ with fixed slope g , the right panel for variable slope with fixed noise amplitude. In the course of averaging, the estimation L_1 continuously approximates the true value L_0 without ever reaching it.

response threshold. As shown in Figure 17, the distance between these two quantities is composed of two contributions ΔL_1 and ΔL_2 . Since the contribution ΔL_1 is now under control, the only unknown remaining is the difference ΔL_2 . This term accounts for the fact that even under ideal conditions (free of interference), not every stimulation that leads to a perception of the stimulus is accompanied by a measurable response of the respective sensory, synaptic, or neuronal target structure. For example, with electrocochleography (ECoChG), the compound action potential (CAP) is detectable virtually directly at the threshold of hearing, whereas the cochlear microphonics (CM) are not visible until stimulus levels which exceed the hearing threshold by more than 50 dB [22]. For the consideration of effects of this kind, the only practicable procedure is the application of an empirically justified method-specific correction. However, as a result of the compensation of the insufficient noise liberation, the corrections will be smaller and the accuracy will increase as a result.

Conclusion and outlook

The devices offered today for practical use for the measurement of OAEs and AEPs undoubtedly have a high technical level, but they still do not exhaust the possibilities of signal processing. From the technical limitations in the pioneering days of electrical response audiometry, which date back about half a century, relics have been preserved which have lost their *raison d'être* today and have been partly adopted for the measurement of OAEs. A prominent example of unused technical resources is the discarding of signal sections with undesirably large amplitude (“artifacts”) immediately after their registration. Today, the availability of storage space and computational capacity make it possible to store all signal segments (sweeps) until the completion of data acquisition in order to sort them later and to be able to discard those sweeps in which the damage caused by the unfavorable signal-to-noise ratio outweighs the benefit of an additionally recorded physiological response [28]. It is generally known,

even in early life, that only at the end of the scholar year, when all school grades are available, there is a reliable basis for deciding which of the individual assessments should be excluded when calculating the average grade. The feasibility of sorted averaging and the resulting benefit have already been demonstrated [29].

The starting point for the present review was the conviction that the use of all signal processing options leads to a reduction of the certainly very high number of negative AEPs or OAEs results that cannot be attributed to pathology. Many of the tools presented in this overview are well-established basic instruments, the use of which should be – but is not – self-evident in the detection of small and noisy signals. Other approaches are new and so far only tentatively tested, so their benefit cannot be quantified. It seems appropriate to note that the algorithms of power-optimized signal processing run automatically in the background and do not require any additional knowledge or intervention from the examiner.

Only the derivation of the auditory threshold from the response threshold requires a minimal intervention by the user. However, the effort associated with the correction of the residual noise does not significantly exceed that of the previous application of questionable “extrapolation rules”. Since the majority of objective hearing tests performed in practice serve to determine the hearing threshold, and because incorrect determination of the threshold can have serious consequences particularly in children [23], this effort is certainly justified.

It will certainly also help to increase the quality of objective audiometry if manufacturers and users pay increased attention to the relevant standards [5] and [6] and recommendations [1], [21], [25]. Finally, it should be noted that the lack of a uniform nomenclature for the methods and parameters unnecessarily complicates their use [20]. In this respect, as in the more in-depth study of the contents of the article now coming to an end, the manufacturers are certainly challenged to a greater extent than the users of their products.

Notes

Competing interests

The author declares that he has no competing interests.

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