Research Article

Vestibular Evoked Myographic Correlation

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ABSTRACT

This work started from the hypothesis that the physiological processes giving rise to the vestibular evoked myogenic potential (VEMP) can be induced not only by transient sounds but also by a continuous stimulation with a stochastic signal. The hypothesis is based on the idea that the number of motor unit action potentials (MUAPs) decreases after a momentary amplitude increase of the effective stimulus, whereas a momentary amplitude decrease has the opposite effect. This concept was theoretically analyzed by assuming that the effective stimulus is closely related to the envelope of the stimulus actually presented. The analysis led to the prediction that the cross-correlation function of the effective stimulus and the measured electromyogram (EMG) has VEMP-like properties. Experiments confirmed this prediction, thus providing evidence of a novel electrophysiological response: the vestibular evoked myographic correlation (VEMCorr). The methodological approach corresponded to a conventional VEMP study, except that the stimulus (delivered with a hand-held minishaker) comprised not only a series of 500-Hz tone pulses (classical VEMP measurement, for comparison) but also sequences of narrow-band noise with a center frequency of 500 Hz (VEMCorr measurement). Each of the 12 test persons showed a clear VEMCorr. Moreover, VEMP and VEMCorr largely resembled each other, as predicted. Apparently they are two different expressions of a more general mechanism that leads to a roughly linear relationship

between stimulus envelope and expectation of the EMG. Future applications of the VEMCorr could exploit that a continuous-stimulation paradigm allows for varying the center frequency of the stimulus without changing the relative bandwidth.

Keywords: vestibular evoked myogenic potential, VEMP, VEMCorr, electromyogram, stimulus envelope

INTRODUCTION

Bickford et al. (1964) proved that a certain class of click-evoked potentials recordable from the human scalp, hitherto assumed to be cortical in origin (Geisler et al. 1958), arises mainly from the neck muscles inserting near the inion. The fact that this so-called inion response was found also in deaf patients suggested that the underlying reflex originates from vestibular receptors (Bickford et al. 1964; Cody et al. 1964). Subsequent work by Townsend and Cody (1971) led to the more precise conclusion that “the saccule is the end organ receptor for the inion response.” At that time, the inion response was seen mainly as a problem for investigations of the auditory system using evoked potentials (Cody and Klass 1968). Although the possible benefit for examinations of the vestibular system was recognized (Cody and Bickford 1969), no systematic research was done in this direction. A turning point was the work of Colebatch et al. (1994). They found, over the sternocleidomastoid muscle, a potential similar to the inion response, for which the term *vestibular evoked myogenic potential* (VEMP) was coined shortly afterwards (Halmagyi et al. 1994). It was concluded also in this case that the underlying reflex is triggered by sound stimulation of the saccule.

The VEMP generally has a stereotyped waveform: A positive peak with a latency of about 13 ms (p13) is

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followed by a negative peak with a latency of about 23 ms (n23). Possible later peaks, with latencies around 34 and 44 ms (n34 and p44), are presumably caused by the activation of cochlear hair cells (Colebatch et al. 1994). By recording the potentials of single motor units using needle electrodes, Colebatch and Rothwell (2004) traced the p13-n23 response back to a short inhibition of tonic muscle activity. Their findings not only explain why a successful VEMP recording critically depends on a sufficient contraction of the target muscle but also provide the basis for developing a mathematical model. Starting from the fact that the electromyogram (EMG) represents a superposition of motor unit action potentials (MUAPs) occurring at irregular time intervals, Wit and Kingma (2006) developed a VEMP model by assuming that MUAP generation is shortly inhibited after the presentation of an intense brief sound. This basic idea has led to a refined model (Lütkenhöner et al. 2010, 2011), in which the VEMP is explained by two mathematical functions: one specifying the mean number of MUAPs occurring per unit time (MUAP rate), and the other one describing the time course of a typical MUAP. Although the latter (from now on briefly denoted as the MUAP) is not related to the vestibular system, it largely determines the waveform of the VEMP. For sufficiently short inhibitions, VEMP and MUAP are even virtually identical, except that the polarity and the scaling differ. More generally, the VEMP may be considered as a band-pass filtered MUAP rate, where the impulse response of the filter corresponds to the MUAP (Lütkenhöner and Basel 2011). First attempts have been made to undo this filtering by deconvolution (Lütkenhöner 2015; Lütkenhöner and Basel 2012).

The present study was motivated by the hypothesis that the model just outlined is valid beyond the narrow context for which it was originally devised. Putting aside mathematical details for now, the basic idea is that the stimulus dependence of the MUAP rate is at least qualitatively predictable, such that a momentary increase of the stimulus amplitude causes, with some delay, a reduced MUAP rate, whereas a momentary decrease of the stimulus amplitude has the opposite effect. In this generalized framework, a VEMP measurement corresponds to the special case that brief pulses presented at relatively long (typically regular) intervals serve as the stimulus. The experimental investigation of other stimulation paradigms is impeded by the fact that the EMG provides only a filtered version of the MUAP rate. For example, long-lasting inhibitory periods leading to a constant (although reduced) MUAP rate would leave almost no signature in the measured EMG, except that the beginning and the end of the inhibitory period would be marked by an On and an Off response, respective-

ly.¹ A completely different situation would arise if the stimulus amplitude, although being constant on average, exhibited random fluctuations. In what follows, this scenario is first analyzed theoretically. The analysis leads to the prediction that the cross-correlation of the envelope of the stimulus with the measured EMG should be closely related to the VEMP. To indicate the affinity of the predicted new kind of vestibular “response” with the VEMP, the term vestibular evoked myographic correlation (VEMCorr) is suggested. The theoretical section is followed by the presentation of experimental results, which confirm that the VEMCorr really exists, ipso facto suggesting that the overall modeling concept is correct.

THEORY

This section begins with a greatly simplified explanation of the kinship of VEMP and VEMCorr (“[Simplified Theory in a Nutshell](#)”). The explanation is mainly addressed to readers with a limited theoretical background, for whom the detailed theory presented later may be difficult to comprehend. Those readers may continue thereafter with “[Materials and Methods](#).” Mathematically skilled readers, by contrast, may immediately proceed to “[EMG Model](#)” now.

Simplified Theory in a Nutshell

A key concept of this work is that the expectation of the EMG recorded in a VEMP or VEMCorr experiment is, at least to a first approximation, linearly related to a so-called normalized effective stimulus (“linear model”). The normalized effective stimulus is derived from the envelope of the stimulus actually presented by subtracting the mean value and applying a normalization. A basic intuitive understanding of the analogy between the VEMP and the proposed VEMCorr can be obtained by approximating the effective stimulus in a VEMCorr experiment (usually a continuous signal) by a sequence of short rectangular pulses. The assumption of a linear model implicates that each rectangular pulse contributes a VEMP-like response to the recorded EMG, with an amplitude proportional to the amplitude of the respective pulse (the amplitude values can be positive or negative). The VEMCorr (being a cross-correlation function) is, in essence, the sum of time-shifted and rescaled copies of the recorded EMG. The time-shifting has the effect that one complete set of VEMP-like responses (one response for each rectangular pulse) becomes perfectly synchronized, and the

¹ A possible Off response was recently observed by Ciardo et al. (2016).

rescaling (with a factor corresponding to the amplitude of the associated rectangular pulse) ensures, among others, that those VEMP-like responses have a consistent polarity. Altogether this means that one set of VEMP-like responses can eventually be added without any mutual cancelation. Two conclusions can be drawn from this consideration. First, the existence of the VEMCorr crucially depends on how realistic the proposed linear model is. Second, if the VEMCorr exists, its physical appearance can be expected to be similar to that of the VEMP.

EMG Model

Taking up the line of thought developed in the “[Introduction](#),” the EMG, $y(t)$, is described by an equation of the form

$$y(t) = n(t) + \int_{-\infty}^{\infty} x(t-\tau) h(\tau) d\tau. \quad (1)$$

The second term on the right-hand side, which will be referred to as the expectation of the EMG, represents a convolution of the MUAP rate, $x(t)$, with the MUAP, $h(t)$. Since the MUAP rate specifies the *expected* number of MUAPs per unit time, but not the factual number and by no means the exact occurrence times of the MUAPs, the EMG recorded in a real experiment differs, of course, in each repetition. The difference between measurement and expectation, which comprises both the stochastic component of the EMG and measurement noise, is denoted as $n(t)$. A detailed characterization of $n(t)$ is not required here, except that this signal is assumed to be uncorrelated with $x(t)$ and to have zero mean. The MUAP is assumed to have the property

$$\int_{-\infty}^{\infty} h(\tau) d\tau = 0 \quad (2)$$

so that increasing or decreasing the MUAP rate by an arbitrary additive constant does not affect the expectation of the EMG.

A key idea of the present work is that the MUAP rate, $x(t)$, is not completely unknown, but can be partially predicted from the stimulus that was presented. More specifically, $x(t)$ is assumed to be linearly related to a function $u(t)$, called the normalized effective stimulus. The question as to how the latter can be derived from the stimulus will be considered later. The proposed relationship between $x(t)$ and $u(t)$ is

$$x(t) = x_0 + \alpha \int_{-\infty}^{\infty} u(t-\tau) v(\tau) d\tau. \quad (3)$$

The parameter x_0 in this equation represents the MUAP rate that would ensue without stimulation, α is a

proportionality constant (being negative if the effect of stimulation is inhibitory), and the convolution integral describes filtering of $u(t)$. The impulse response of the filter, $v(t)$, as well as the parameters x_0 and α are unknown. Conceptually, all three may depend on the experimental condition and the test person.

Plugging the above model for $x(t)$ into Eq. (1) yields, after a minor rearrangement,

$$y(t) = n(t) + \alpha \int_{-\infty}^{\infty} u(t-\tau) \tilde{h}(\tau) d\tau \quad (4)$$

with

$$\tilde{h}(t) = \int_{-\infty}^{\infty} h(t-\tau) v(\tau) d\tau. \quad (5)$$

The function $\tilde{h}(t)$ represents a filtered MUAP (for convenience, MUAP and filtered MUAP are not consistently distinguished in this article). Note that the parameter x_0 dropped out of the equation, owing to Eq. (2).

Equation (4) was formulated for an experiment of infinite duration. A finite duration is obtained by assuming that $u(t)$ is zero outside a certain time frame. In the frequency domain, Eq. (4) reads

$$Y(f) = N(f) + \alpha U(f) \tilde{H}(f), \quad (6)$$

where $Y(f)$, $N(f)$, $U(f)$, and $\tilde{H}(f)$ are the Fourier-transforms of $y(t)$, $n(t)$, $u(t)$, and $\tilde{h}(t)$, respectively (as to the frequency-domain representation of the convolution integral see, e.g., Brigham 1988; Press et al. 2007).

Cross-correlation of EMG and Normalized Effective Stimulus: VEMCorr

The cross-correlation of a function $u(t)$ with a function $y(t)$ is defined as

$$c_{uy}(\ell) = \int_{-\infty}^{\infty} u(t) y(t+\ell) dt, \quad (7)$$

where ℓ denotes the time lag (see, e.g., Brigham 1988; Papoulis 1962; Press et al. 2007; Yarlagaadda 2010). In the frequency domain, this equation reads

$$C_{uy}(f) = U^*(f) Y(f), \quad (8)$$

where $C_{uy}(f)$ is the Fourier-transform of $c_{uy}(\ell)$, and $U^*(f)$ is the complex conjugate of $U(f)$ (see, e.g., Brigham 1988; Press et al. 2007). Plugging Eq. (6) into (8) results in

$$C_{uy}(f) = U^*(f) N(f) + \alpha U^*(f) U(f) \tilde{H}(f). \quad (9)$$

Back transformation into the time domain then yields

$$c_{uy}(\ell) = c_{un}(\ell) + \alpha \int_{-\infty}^{\infty} c_{uu}(\ell - \tau) \tilde{h}(\tau) d\tau, \quad (10)$$

where c_{un} and c_{uu} are defined analogously to c_{uy} (in Eq. (7), y has to be replaced by n and u , respectively). Since $n(t)$ and $x(t)$ were assumed to be uncorrelated (implying that $n(t)$ and $u(t)$ are uncorrelated as well), the expectation of $c_{un}(\ell)$ is zero. While $c_{uu}(\ell)$ is the autocorrelation function of the normalized effective stimulus, $c_{uy}(\ell)$, representing the cross-correlation of the normalized effective stimulus with the measured EMG, is obviously the VEMCorr proposed in the “Introduction.”

Comparison of VEMCorr and VEMP

In a typical VEMP experiment, the stimulus consists of a series of identical short pulses. Thus, the corresponding normalized effective stimulus can be represented by the function

$$u(t) = \sum_{k=1}^K u_0(t - t_k), \quad (11)$$

where $u_0(t)$ is the contribution of a single pulse occurring at time $t = 0$. The VEMP is usually obtained by stimulus-triggered averaging of the recorded EMG. For the present model, this yields (after plugging Eq. (11) into (4))

$$\text{vemp}(t) = \frac{1}{K} \sum_{k=1}^K n(t - t_k) + \alpha \int_{-\infty}^{\infty} u_0(t - \tau) \tilde{h}(\tau) d\tau. \quad (12)$$

A comparison of this equation with Eq. (10) reveals a close similarity: The expected value of the first term is zero in both cases, and the second terms have exactly the same mathematical structure. What differs is that the VEMP is derived from the normalized effective stimulus itself, whereas the VEMCorr is derived from the autocorrelation function of the normalized effective stimulus. A second and potentially critical difference becomes evident by considering how the VEMCorr and the VEMP are calculated from measured data: The estimation of the VEMCorr requires knowledge of the normalized effective stimulus, whereas the estimation of the VEMP only requires knowledge of the trigger times t_k ($1 \leq k \leq K$).

Normalized Effective Stimulus

From the above comparison between VEMCorr and VEMP, it can be concluded that the concept of a VEMCorr lacks practical relevance as long as the normalized effective stimulus is unknown. Thus, the all-important question to be answered is whether or not it is possible to derive at least a rough guess of the normalized effective stimulus from the stimulus presented.

A good starting point could be the stimulus envelope. The envelope of a real-valued signal $x(t)$ can be obtained by calculating the magnitude of the associated analytical signal $x(t) + ix_H(t)$, where $x_H(t)$ is the Hilbert transform of $x(t)$ (see, e.g., Bracewell 1986). Although this approach is convenient from a mathematical point of view, the question arises, of course, as to how known physiological mechanisms could lead to a signal closely related to the stimulus envelope. Rectification at the level of the vestibular hair cells (Eatock and Songer 2011; Songer and Eatock 2013; Soto et al. 2002) appears to be a plausible possibility. This is exemplified in Fig. 1. The black solid curve in Fig. 1a shows rectified narrow-band (third octave) filtered noise with a center frequency of 500 Hz; the dotted curve shows what was suppressed by the rectifier. The envelope of the narrow-band noise is represented by the gray curve. A convolution of the latter curve with the MUAP displayed in the inset resulted in the gray curve in Fig. 1b, which represents the expectation of the EMG in this model simulation. Almost the same result (black curve) is obtained by convolving the rectified noise rescaled by a factor of π (exactly this factor would be required for a rectified sine). Thus, convolution with the MUAP essentially removes all differences between rectified signal and envelope. For the sake of completeness, the convolution with the MUAP was also applied to the narrow-band noise itself (before rectification). The result (dotted curve in Fig. 1b) basically corresponds to the zero line, which demonstrates that rectification (or some other operation by which the original stimulus is transformed into a signal related to the stimulus envelope) is critical here.

To be consistent with the idea of a *normalized* effective stimulus, the stimulus envelope has to be transformed. For stimuli such as the narrow-band filtered noise considered in Fig. 1, a suitable definition of the normalized effective stimulus is

$$u(t) = \frac{s(t) - \bar{s}}{\sigma_s}, \quad (13)$$

where $s(t)$ is the stimulus envelope, and $\bar{s}$ and σ_s denote the mean and the standard deviation of $s(t)$ in the time

FIG. 1. Model simulation illustrating that the stimulus envelope and the rectified stimulus lead to basically the same expected EMG. **a** The stimulus corresponded to narrow-band noise with a center frequency of 500 Hz. Rectification of the stimulus resulted in the black solid curve; the dotted curve shows what the rectifier rejected. The envelope of the stimulus (derived from the associated analytical signal) is represented by the gray curve. **b** Expected EMG. The gray

curve and the almost identical black curve were obtained by convolving the corresponding curves in **a** with the MUAP shown in the inset (in the case of the black curve: additional scaling by a factor of π). Convolution of the original stimulus (no rectification) with the MUAP resulted in the dotted curve, which basically corresponds to the zero line

range of interest. Owing to the normalization, $u(t)$ depends only on the temporal structure of the stimulus, but not on the stimulus amplitude. The effect of the latter is accounted for by the proportionality factor α in Eq. (4). Even when noise serves as stimulus, the normalized effective stimulus is a deterministic signal: $u(t)$ is derived in a predetermined way from the stimulus that is actually presented in the experiment considered.

Model Simulation Illustrating the Kinship of VEMP and VEMCorr

To complete the theoretical part of this work, the close relationship between the VEMP and the hypothesized VEMCorr is illustrated with a model simulation. The two curves in Fig. 2a exemplify the autocorrelation function of the normalized effective stimulus (scaled such that the autocorrelation at zero lag is 1). The same model as in Fig. 1 was used, except that the stimulus durations were 1 (black curve) and 100 s (gray curve). The curves

are basically identical, apart from the fact that the black one appears much noisier. A convolution of these curves with the MUAP shown in the inset of Fig. 1 resulted in the corresponding curves in Fig. 2b. According to Eq. (10), these curves represent the expected VEMCorr. For comparison, the MUAP (appropriately rescaled) is superimposed as a dashed curve. The time courses of MUAP and VEMCorr are quite similar in this example.

After a redefinition of the axes, Fig. 2 can also serve as an example illustrating the calculation of the VEMP using Eq. (12): If the abscissa represented time rather than lag and the curves in Fig. 2a represented a normalized effective stimulus rather than an autocorrelation function, the corresponding curves in Fig. 2b would show the expected VEMP. Under this new perspective, Fig. 2 resembles previous VEMP simulations (see, e.g., Fig. 1 of Lütkenhöner and Basel (2011)). Thus, without knowing the context it may be impossible to decide whether a given curve shows a VEMP or a VEMCorr. This conclusion depends, of

Fig. 2. Kinship of VEMCorr and VEMP. **a** The thick gray curve exemplifies an autocorrelation function of a stimulus envelope. As in Fig. 1, the stimulus corresponded to narrow-band noise with a center frequency of 500 Hz. The stimulus duration was 100 s. The much noisier black curve was obtained for a stimulus duration of only 1 s. **b** VEMCorr obtained by convolving the autocorrelation functions in **a** with the MUAP displayed in the inset of Fig. 1. A properly rescaled version of this MUAP is shown as a dashed curve, for comparison. After redefining the axes, the figure can be reinterpreted as an example illustrating the relationship between effective stimulus and VEMP (for details see text)

course, on the supposition that the underlying model is sufficiently realistic. It remains to be proven by means of dedicated experiments that the hypothesized VEMCorr actually exists.

MATERIALS AND METHODS

Subjects

Twelve healthy adults (three male and nine female) aged 20–34 years (median 24 years) with no history of auditory or vestibular disorder volunteered in this study. All subjects gave written consent after the nature of the experiment had been explained to them. The study was approved by the ethics committee of the General Medical Council Westfalen-Lippe and the Medical Faculty of Münster.

Stimulation

The stimulus design was inspired by the modeling example considered in Figs. 1 and 2. Using a custom MATLAB script, third-octave filtered noise with a center frequency of 500 Hz was generated by passing a sequence of normally distributed random numbers through a second-order digital Butterworth filter with cutoff frequencies of $500/2^{1/6} \approx 445$ Hz and $500 \cdot 2^{1/6} \approx 561$ Hz. The stimulus level was, in the first place, defined in terms of the input voltage provided to the power amplifier, with 0 dB corresponding to a root mean square (RMS) value of 1 V. Four different levels (−10, −5, 0, and 5 dB) were investigated in immediate succession (no break in between). Each level was kept constant for 8 s, so that the total duration of the noise sequence was 32 s. The different levels were studied in increasing order rather than in a randomized way, to preclude the possibility that a stronger stimulus partially masks a subsequent weaker stimulus. The noise sequences were immediately followed by a series of short 500-Hz tone bursts, which served as a reference stimulus. The plateau duration as well as the rise and fall time of each tone burst was 2 ms, the peak level was 15 dB re. 1 V,² and the repetition rate was 4 Hz. Tone bursts were presented for 8 s so that the total duration of the composed stimulus (noise sequences followed by tone burst sequence) was 40 s. The stimulus was generated anew for each experimental run. Thus, the noise sequences in the stimulus were statistically independent in different runs.

A digital representation of the 40-s composed stimulus (40.96-μs sampling interval) was sent to a TDT RP2 real-time processor (Tucker-Davis Technologies, Alachua, FL, USA), which handled the digital-to-analog (D/A) conversion. The resulting analog signal was fed to a Brüel & Kjær power amplifier (type 2718), which finally provided the voltage driving a Brüel & Kjær minishaker (type 4810). To allow for an easy and safe application of the stimulus to the subject's head, the shaker was fitted with a short plastic rod (length of 60 mm, diameter of 25 mm), similar to what was described by Rosengren et al. (2009b). During a measurement, the shaker was held in the hand of the examiner such that the plastic rod was oriented normal to the head. To ensure a good contact between rod and head, the examiner exerted a force of about 15 N. Such an external force substantially influences the intrinsic properties of the shaker (Lütkenhöner 2017), so that variations of the

² A peak level of 15 dB re. 1 V corresponds to a peak amplitude of 5.62 V. This amplitude is comparable to what has been used by other authors in previous studies. For example, the stimuli used by Rosengren et al. (2009b) had peak amplitudes of 5 V, and in a study by Todd et al. (2010) the peak-to-peak amplitude was 20 V.

force could have unforeseeable side effects. Thus, to help maintaining a constant force, the base of the shaker (opposite to the tube) was fitted with a spring (embedded in a kind of damper), which had to be compressed in a predetermined way. Stimulation points were the left and the right mastoid of the subject.

To get an idea of how the shaker transforms the electrical input into mechanical action, an exemplary measurement of the acceleration along the longitudinal axis of the tube was done, while the shaker was pressed (as describe above) against the mastoid of a subject. The acceleration was recorded with a Brüel & Kjaer accelerometer type 4535-B-001 (operated through a Brüel & Kjaer signal conditioner type 1704-A-002), which was screwed to the tube. The result of the measurement is shown in Fig. 3. Owing to the time scale used, the figure does not reveal the fine structure of the stimulus, but it gives a good impression of its overall composition: Noise stimulation at four different levels is followed by a series of tone bursts. The axis on the left provides the measured acceleration in m/s^2 , whereas the axis on the right uses the unit $g=9.81 \text{ m/s}^2$. The noise at the highest level and the tone bursts yield similar peak amplitudes, being of the order of $15 g$.³ The RMS values of the different noises are indicated by white horizontal lines.

Experimental Procedure

Electrodes (BlueSensor T, Ambu A/S, Ballerup, Denmark) were bilaterally placed over the middle portion of the sternocleidomastoid muscle (active electrode), the upper sternum (reference), and the forehead (ground). The electrodes were connected to a gUSBamp biosignal amplifier (g.tec medical engineering GmbH, Schiedlberg, Austria), which continuously sampled the input at a rate of 4800 Hz (high-pass filtered with a cutoff frequency of 2 Hz). Custom MATLAB software running on a PC stored the digitized data, the associated trigger information, and the stimulus in a computer file.

The subject lay supine on an examination couch, the head section of which was raised by about 25° . For a measurement run, each lasting well over 40 s (duration of the stimulus plus some overhead time), the examiner applied the stimulus, as described above, while the subject slightly lifted their head, so that it was not supported anymore by the examination couch. After each run, the subject had the opportunity to relax, while the examiner had a quick look at

the data and prepared the next run. Each experiment comprised 12 runs (six for each stimulation side).

Data Analysis

All data analysis was done using custom MATLAB software. In a first step, the archived stimulus had to be synchronized with the recorded data. This step was necessary because only a trigger channel, but not the stimulus itself was recorded synchronously with the EMG channels. Synchronization was achieved by matching the recorded trigger channel with the trigger information associated with the archived stimulus. A complicating factor was that the actual sampling rates of the D/A and A/D converters apparently differed slightly from their nominal values. As a result the stimulus generated by the D/A converter (nominal duration 40 s) appeared to be 40.00124 s in duration in the recorded data. The problem was solved by assuming that the actual sampling rate of the D/A converter was equal to its nominal value (24,414.0625 Hz), but that the actual sampling rate of the A/D converter was 4800.1489 Hz instead of its nominal value (4800 Hz). This adjusted rate was finally used for resampling both the stimulus and the stimulus envelope. The latter was calculated from the archived stimulus by means of a Hilbert transform, as explained in the theoretical part of this article. Resampling was done by linear interpolation using the MATLAB function *interp1*.

The stimulus envelope was transformed into a normalized effective stimulus using Eq. (13). Then the VEMCorr was estimated by calculating the cross-correlation of the recorded EMG (a possible offset was removed beforehand by a demeaning operation) and the normalized effective stimulus. This was done by invoking the MATLAB function *xcorr* with the scaling option “coeff.” The effect of the option is that both input sequences are normalized, so that their auto-correlations at zero lag have the value 1.

A VEMCorr was estimated not only from the data recorded during continuous narrow-band noise stimulation but also from the data obtained for tone burst stimulation. The latter estimation did not require any change of the algorithm, because the fundamental differences between continuous noise and tone burst stimulation are accounted for by the normalization induced by the option “coeff” (see above). For the sake of clarity, the VEMCorr obtained with transient stimuli such as tone bursts will generally be referred to as the t-VEMCorr, whereas the term VEMCorr will generally be reserved for the case of continuous noise stimulation. Of course, tone bursts were primarily presented to elicit a VEMP. The VEMP was calculated, as usual, by stimulus-triggered averaging of the recorded EMG. The result was subsequently normal-

³ Comparable peak accelerations (20 g) were used, for example, by Iwasaki et al. (2008).

FIG. 3. The stimulus consisted of narrow-band noise sequences of increasing intensity (each lasting 8 s) and a series of tone bursts. The signal displayed is the acceleration along the longitudinal axis of the tube attached to the minishaker

ized by dividing by the root-mean-square (RMS) value of the EMG.

For the subsequent analyses, the results for the single runs were averaged across the 12 runs. This required a relabeling of the channels: The recording from the side of stimulation became the ipsilateral channel, whereas the recording from the other side became the contralateral channel. Averaging was done separately for the ipsilateral and the contralateral channel. More explicitly, the ipsilateral response was obtained by averaging the right-side signals recorded during right-side stimulation and the left-side signals recorded during left-side stimulation, whereas the contralateral response was obtained by averaging the left-side signals recorded during right-side stimulation and the right-side signals recorded during left-side stimulation. As the ipsilateral and the contralateral responses were generally very similar, some analyses were finally performed for the mean of ipsilateral and contralateral recording.

Peak detection was done without human intervention by a simple MATLAB script: Peak p13 was identified as the maximum in the latency range from 11 to 19 ms, whereas peak n23 was identified as the minimum in the latency range from 20 to 28 ms. In the case of the VEMCorr, the latency was determined only for the highest stimulus level. That latency was subsequently used to determine the peak amplitudes for the other stimulus levels. The VEMP peak amplitudes were measured against a pre-stimulus baseline (latencies between -100 and 0 ms). A corresponding baseline was defined for the VEMCorr (latency replaced by time lag). The signal-to-noise ratio of a peak was estimated by dividing the peak amplitude by the RMS value determined for the pre-stimulus baseline.

MATLAB was also used for the statistical analysis of the estimated peak amplitudes and latencies. Differences between two conditions were investigated using a paired *t*-test. The MATLAB function *ttest* provided not only the value of the test statistic *t* and the corresponding *p* value but also the 95 % confidence interval. Correlation coefficients were estimated using the MATLAB function *corrcoef*, which provided the correlation coefficient *r* as well as the probability *p* of getting a correlation coefficient as large as the observed value by random chance. Linear least-squares fitting was done using the MATLAB function *polyfit*.

Adjustment of the Zero Points for Time and Time Lag

In contrast to the carefully balanced model simulation presented in Fig. 2, experimental data are not expected to show exactly the same peak latencies for VEMP and VEMCorr. A crucial prerequisite for a meaningful interpretation of latency differences is a meticulous adjustment of the zero points for time and time lag, respectively. A dedicated experiment was carried out to achieve this: With the shaker positioned on the floor, the acceleration along the longitudinal axis of the tube was recorded. Apart from the fact that an accelerometer (already specified above) was connected to the biosignal amplifier rather than EMG electrodes and that the stimulation level was reduced by 40 dB (to avoid overdriving the biosignal amplifier), the procedure corresponded to the investigations of VEMP and VEMCorr. The measured data was used to adjust the zero point of the time axis, so that the tone burst recorded by the accelerometer optimally corresponded to the theoretical tone burst. Figure 4a documents the match achieved. The thick gray curve

represents the theoretical tone burst, i.e., the signal that was sent to the D/A converter of the stimulus generator, whereas the solid black curve represents the measured acceleration (averaged over all tone burst presentations). The correspondence between the two curves is evidently quite good. The fact that the gray curve is mostly lagging the solid black curve has no physical meaning, but is due to the sampling interval of roughly 0.2 ms, which did not allow for achieving a better temporal match. For comparison, the dashed curve represents the output voltage of the power amplifier driving the shaker (signal provided by the voltage monitor of the amplifier). The difference between dashed and solid black curve indicates that the measured acceleration is slightly lagging the driving voltage, as to be expected.

Figure 4b corresponds to Fig. 4a, except that the envelope of the stimulus rather than the stimulus itself is considered and that a different time scale is used (see bottom of the figure). According to the ideas developed in the theoretical part of this article, the stimulus envelope is closely related to the effective stimulus that gives rise to the VEMP. Thus, the latencies of the VEMP peaks are expected to reflect the fact that the plateau of the stimulus begins 1 ms after stimulus onset and has a duration of 2 ms. Figure 4b will, indeed, be crucial for discussing latency differences between VEMP and VEMCorr.

The bottom two panels of Fig. 4 can serve as a reference for discussing latency differences between t-VEMCorr and VEMCorr. Figure 4c is based on the same data as Fig. 4b, but the result of a correlation analysis is shown. The thick gray curve represents the autocorrelation function of the tone burst envelope (i.e., the envelope shown as a gray curve in Fig. 4b), whereas the solid black curve represents the cross-correlation between the tone burst envelope and the envelope of the measured acceleration (i.e., the envelope shown as a solid black curve in Fig. 4b). The two curves are almost identical, and both show a maximum at zero lag. Very similar curves were obtained for narrow-band noise (Fig. 4d), owing to the chosen bandwidth. In particular, the peak of the black solid curve is again found at zero lag. Thus, t-VEMCorr and VEMCorr should have identical peak latencies unless the ideas developed above are too simple in some respect.

RESULTS

Figure 5 provides an overview of the results of the main experiment. Three subjects were exceptional in so far as a VEMCorr could be found at all investigated levels. The results for these subjects are presented in

Fig. 4. Experimental adjustment of the zero points of time and lag. **a** Theoretical tone burst (gray curve), measured acceleration (solid black curve) and output voltage of power amplifier (dashed curve). **b** Envelopes of the signals in **a**. **c** Autocorrelation function for the envelope of the theoretical tone burst (gray curve), and cross-correlation function for the envelopes of theoretical tone-burst and measured acceleration (solid black curve). **d** Corresponding correlation analysis for narrow-band noise rather than a tone burst

the first three columns. In the fourth column, grand-averaged results for the remaining subjects are presented. The upper five rows show the VEMCorr for the four different noise levels and the tone burst,

FIG. 5. Measured VEMCorr and VEMP. First three columns: results for three individual subjects. Fourth column: grand-averaged results for the other subjects. Upper four rows: VEMCorr for narrow-band noise stimulation (four different stimulus levels). Fifth row: VEMCorr for tone-burst stimulation (t-VEMCorr). Bottom row: VEMP. Both the ipsilateral recording (black curve) and the contralateral recording (gray curve) are shown. The dotted horizontal lines indicate the RMS

values determined from the pre-stimulus baseline (polarity inverted for contralateral recording). The zero level is indicated by a short horizontal line on the right of each panel. Both the VEMCorr and the VEMP are dimensionless here: The VEMP was normalized by dividing by the RMS value of the EMG, whereas in the case of the VEMCorr a value of 1 corresponds to a perfect correlation between normalized effective stimulus and recorded EMG

respectively. The bottom row shows the VEMP calculated from the tone burst data. The ipsilateral recording is represented by a black curve, the contralateral recording by a gray curve. The two curves are generally similar, particularly in the case of the grand-averaged results displayed in the right-most column. The dotted horizontal lines allow for assessing the signal-to-noise ratios. These lines indicate the RMS values estimated for the ipsilateral and the contralateral recording, respectively (in the latter case: values multiplied by -1 , for better visibility). The zero level is indicated by a short horizontal line on the right of each panel.

A comparison of corresponding curves in the bottom two rows of Fig. 5 reveals a high degree of similarity, apart from scaling and a minor time shift. The similarity is not surprising because the curves were obtained by analyzing the same data in two closely related ways: A correlation analysis yielded the t-VEMCorr shown in row 5, and averaging of stimulus-related epochs yielded the VEMP shown in row 6. Corresponding curves in rows 4 and 5 resemble each other as well, although to a lesser degree. The resemblance corroborates the theoretical idea that the time courses of VEMP and VEMCorr largely reflect the time course of the MUAP. In view of this

resemblance, the VEMCorr peaks will be labeled using the nomenclature established for the VEMP. Dotted vertical lines indicate the latencies of the peaks p13 and n23 of the VEMCorr at the highest stimulus level (row 4). The latencies at lower stimulus levels are roughly the same, whereas the t-VEMCorr and the VEMP generally have longer latencies. As expected, the VEMCorr decreases in amplitude when the stimulus level is reduced. As a consequence, most subjects showed a clear response only at the two highest levels. Nevertheless, in the grand average, there is a clear response also at the second lowest level and probably a small response even at the lowest level.

Although the main focus of this article is on the VEMCorr, the tone burst stimulation plays a pivotal role as a reference condition, because both a t-VEMCorr and a VEMP can be estimated from the data. The comparison between rows 5 and 6 in Fig. 5 already revealed a great similarity. A more detailed investigation of this similarity is provided in Fig. 6. The upper panel (a) shows an analysis of the p13 amplitude and the bottom panel (b) an analysis of the n23 amplitude. Abscissa and ordinate represent the amplitudes of t-VEMCorr and VEMP, respectively. Three symbols are provided for each subject: A cross represents the ipsilateral recording, a plus sign the contralateral recording, and a filled circle the mean of ipsilateral and contralateral recording. The results for the three exceptional subjects considered in the first columns of Fig. 5 are indicated by larger circles. Likewise, the two subjects with the worst signal-to-noise ratio are indicated by smaller circles. As expected, the amplitudes of t-VEMCorr and VEMP are highly correlated: The correlation coefficient, r , is 0.992 for p13 and 0.995 for n23 ($p < 10^{-9}$). A solid line represents the result of a linear least-squares fit, whereas the dotted line represents the best fit under the assumption that the amplitudes of t-VEMCorr and VEMP are proportional to each other. The two lines are roughly consistent, for both p13 and n23. The slopes of the dotted lines are 8.69 (p13) and 8.24 (n23), respectively.

The analysis presented in Fig. 7a–b is analogous to the one presented in Fig. 6, except that the t-VEMCorr (abscissa) is compared with the VEMCorr at the highest stimulus level (ordinate). The correlation coefficient, r , is 0.919 for p13 and 0.896 for n23 ($p < 10^{-4}$). The dotted lines, again representing proportionality, have slopes of 0.88 (p13) and 1.11 (n23), respectively. In combination with the previously revealed very high correlation between t-VEMCorr and VEMP, the present results imply that VEMCorr and VEMP are correlated as well.

The analysis presented in Fig. 7c corresponds to those in the upper two panels, except that the amplitude ratio of the peaks p13 and n23 is consid-

Fig. 6. Amplitude of t-VEMCorr (abscissa) versus amplitude of VEMP (ordinate). **a** Analysis for peak p13. A cross represents the ipsilateral recording, a plus sign the contralateral recording, and a filled circle the mean of the two recordings. Larger circles identify the three exceptional subjects considered in the first columns of Fig. 5, smaller circles identify the two subjects with the worst signal-to-noise ratio. A linear least-squares fit resulted in the solid line (dotted line if proportionality was assumed). **b** Corresponding analysis for peak n23

ered instead of the absolute amplitudes. There is only a weak correlation ($r = 0.582$, $p < 0.05$). While the solid line represents the result of a linear least-squares fit, the dashed line represents the situation that the amplitude ratio is identical for VEMCorr and t-VEMCorr. Almost all data points are located below this line. Thus, the VEMCorr tends to exhibit a smaller amplitude ratio than the t-VEMCorr.

As already mentioned in the context of Fig. 5, the peak latencies of VEMCorr, t-VEMCorr, and VEMP generally differ. Figure 8 presents a closer examination for peak p13. In contrast to the previous figures, only the mean of ipsilateral and contralateral recording is studied. Moreover, results are displayed only if the peaks compared have a signal-to-noise ratio greater than 3. The abscissa represents the p13 latency of the t-VEMCorr, whereas the ordinate

Fig. 7. a–b Amplitude of the t-VEMCorr (abscissa) versus amplitude of the VEMCorr at the highest stimulus level (ordinate). Otherwise corresponding to Fig. 6. c Comparison of the amplitude ratios p13/n23. A linear least-squares fit yielded the solid line. The assumption of identical amplitude ratios is represented by the dashed line

represents the p13 latency of the VEMCorr and the VEMP (data points displayed as circles and squares, respectively). Filled symbols are used for the three individual subjects considered in the first columns of Fig. 5, open symbols are used for the other subjects (the area of an open symbol is proportional to the

Fig. 8. Latency of t-VEMCorr peak p13 compared to the corresponding latencies of VEMCorr (circles) and VEMP (squares). The three exceptional subjects considered in the first columns of Fig. 5 are identified by filled symbols. Data points representing identical latencies would be found on the dashed line. Shifting this line by the mean latency difference of t-VEMCorr and VEMP (2.84 ms) resulted in the dotted line. A linear least-squares fit for the data points represented by squares yielded the solid line

number of symbols sharing the respective location). Data points with identical abscissa and ordinate values would be found on the dotted line. However, apart from one exception, all data points are found clearly above or below this line. The distribution of the symbols reflects the fact that the p13 latency of the t-VEMCorr is 1.41 ± 0.67 ms longer than the p13 latency of the VEMCorr (paired $t(7) = 5.97$, $p < 10^{-3}$; 95 % CI [0.85, 1.96] ms) and 2.83 ± 0.64 ms shorter than the p13 latency of the VEMP (paired $t(9) = 14.04$, $p < 10^{-6}$; 95 % CI [2.38, 3.29] ms). The dotted line runs parallel to the dashed line, at a distance corresponding to the

mean latency difference between t-VEMCorr and VEMP. The correlation coefficient for the data points represented by squares is $r=0.975$ ($p<10^{-5}$). A linear least-squares fit through these points yielded the solid line.

Figure 9 illustrates how the amplitude of the VEMCorr depends on the stimulus level. Peak p13 is considered in the upper panel, peak n23 in the bottom panel. All amplitudes were individually normalized by dividing by the amplitude at the highest level (thus yielding a value of unity at this level). The open symbols represent individual subjects where the respective peak had a signal-to-noise ratio greater than 3. The number of subjects fulfilling that requirement is displayed above the upmost symbol for the respective level. The three exceptional subjects considered in the first columns of Fig. 5 are, again, represented by especially large circles. The filled

Fig. 9. Level dependence of the VEMCorr amplitude (upper panel: peak p13; bottom panel: peak n23). Larger circles identify the three exceptional subjects considered in the first columns of Fig. 5. Filled squares represent the grand average of the other subjects. All amplitudes were normalized, so that an amplitude of unity was obtained for the highest stimulus level. The inserted numbers indicate for how many subjects a signal-to-noise ratio greater than 3 was obtained for the respective condition

squares, by contrast, represent the grand average of all other subjects. It can be concluded from the figure that decreasing the stimulus level by 10-dB reduces the response amplitude by about a factor of three. A consequence of this amplitude decrease is that, at the lowest stimulus levels, only few subjects fulfill the signal-to-noise ratio criterion.

DISCUSSION

Linear Approach for a Non-linear System

The most important result of this study is that the hypothesized VEMCorr really exists. This finding proves that the ideas developed in the theoretical part of this article are largely correct: In essence, VEMP and VEMCorr appear to be two different expressions of a more general mechanism that leads to a roughly linear relationship between stimulus envelope and expectation of the EMG. Such an outcome was not necessarily to be expected from the outset, because all theoretical considerations were based on a linear model. Thus, it was not at all clear whether non-linear effects not accounted for would reduce the predictive value of the model to such an extent that the idea of a VEMCorr would have to be abandoned.

The great success of the linear approach does not imply a lack of non-linearities. In fact, the whole theory builds on a signal that is the result of a non-linear transformation: Not the stimulus itself, but its envelope is considered relevant. The transformation from a signal closely related to the stimulus to a signal closely related to the stimulus envelope presumably takes place at the level of the vestibular hair cells, where tilting the hair bundle towards the kinocilium causes much larger voltage changes than displacements in the opposite direction (Soto et al. 2002). Owing to this directional sensitivity, the hair cells basically represent a rectifier, and a rectifier combined with a low-pass filter is an envelope detector (see, e.g., Proakis and Salehi 2002). In the present case, the rectified signal is presumably not smoothed in a single step, but by an entire cascade of processes having low-pass filter characteristics. Among these processes are, for example, diffusion mechanisms between different transmitter reservoirs (Zilany et al. 2009). All in all, the emergence of a signal resembling the stimulus envelope can be explained quite well by known physiological mechanisms, although the details are still sketchy.

Because it is the stimulus envelope that serves as the starting point for the linear modeling approach, the non-linear nature of the transformation from stimulus to stimulus envelope does not require further attention here. Non-linearities affecting only the

overall amplitude of the response, but not the time course, are not a matter of concern either: Such effects are handled by the amplitude factor α in Eq. (3), which is considered to be constant for a given experiment, notwithstanding that its value may fundamentally differ for experiments taken under altered conditions. For example, α non-linearly depends on stimulus intensity.⁴ The filtered MUAP, $\tilde{h}(t)$, is conceived to be invariant for a given experimental condition so that a possible non-linear dependence on experimental parameters is uncritical as well. Potentially problematic, however, would be short-term temporal effects such as those found for the post-auricular reflex. For example, in experiments with pairs (or longer series) of clicks, the second click elicited a smaller post-auricular response than the first click (Fox et al. 1989; Hackley et al. 2017). While no corresponding findings were obtained for the inion response (Fox et al. 1989), VEMP experiments with pairs of tone bursts (Welgampola and Colebatch 2001) also showed an amplitude reduction for the second response, apart from one experimental condition for which a paradoxical amplitude increase was reported. It was not clear from the outset whether such short-term effects would spoil the postulated close relationship between stimulus envelope and expectation of the EMG. The experimental proof of the VEMCorr eventually showed that this concern was unfounded.

VEMCorr Versus VEMP

The experiments also confirmed the theoretical prediction that the VEMCorr is a phenomenon closely resembling the VEMP. In view of the observed similarities, it appeared even justified to use the same nomenclature for the peaks in the time courses. The correspondence was greatest between t-VEMCorr and VEMP. Not only were the time courses very similar (compare the bottom two rows of Fig. 5), but the peak amplitudes obtained for the individual subjects were highly correlated (Fig. 6). These findings are, of course, not surprising, because t-VEMCorr and VEMP were derived from exactly the same experimental data by closely related computations: Averaging as used for the calculation of the VEMP is basically equivalent to the calculation of the cross-correlation of a series of Dirac delta functions (representing the triggers for the tone bursts) with the recorded EMG. From a theoretical point of view, the main difference between VEMP and t-VEMCorr is that, in the latter case, the Dirac delta function is replaced by the envelope of the

transient stimulus (tone burst in the present case). This also explains why the latency of VEMP peak p13 is, on average, 2.83 ms longer than the corresponding peak of the t-VEMCorr (cf. Fig. 8): In the case of the VEMP, the trigger coincides with the very beginning of the stimulus, and it takes some time until the stimulus has maximum effect. In the case of the t-VEMCorr, by contrast, the calculation of the cross-correlation implicitly accounts for the shape of the tone burst envelope. Indeed, the observed latency difference roughly corresponds to half the duration of the tone burst (3 ms). The t-VEMCorr was considered here mainly to demonstrate its kinship with the VEMP. In practice, calculation of the latter will suffice whenever such a calculation is possible.

The real significance of the VEMCorr is that it can be calculated under more general conditions than the VEMP, for example with continuous narrow-band noise as the stimulus. Measurements were performed at four different levels of the noise stimulus. The time course of the obtained VEMCorr did not show an obvious dependence on the stimulus level, but the amplitude considerably decreased with decreasing level (Fig. 5). According to the more systematic analysis presented in Fig. 9, a decrease of the stimulus level by 10 dB reduced the VEMCorr amplitude by about a factor of 3. The effect is somewhat greater than was reported for the VEMP (Akin et al. 2003; Rosengren et al. 2009a).

The amplitudes of the VEMCorr peaks p13 and n23 (measured at the highest stimulus level) were correlated with the corresponding amplitudes of the t-VEMCorr (Fig. 7a–b), although not to the same degree as the amplitudes of t-VEMCorr and VEMP. The lower correlation can be partially attributed to the fact that VEMCorr and t-VEMCorr were derived from statistically independent data. Probably more important is that the simple model underlying the analyses does not account for physiological details and their interindividual variability. The latter reason could also explain why the p13/n23 amplitude ratio differs between VEMCorr and t-VEMCorr (Fig. 7c).

Peak p13 of the VEMCorr occurred, on average, 1.41 ms earlier than the corresponding peak of the t-VEMCorr. By contrast, the reference measurements with an accelerometer yielded identical latencies (Fig. 4). The reason for this discrepancy is that the reference measurements were made for a largely linear system, whereas the vestibular system can be assumed to be non-linear in several respects. From the modeling point of view, the latency difference between VEMCorr and t-VEMCorr can be interpreted to the effect that the filtered MUAP, $\tilde{h}(t)$, differs for continuous noise and tone burst stimulation. The physiological basis for the observed latency difference could be that the neural activation causing the t-

⁴ Note that a linear dependence on stimulus level (to some extent found in Fig. 9) corresponds to a non-linear dependence on stimulus intensity, owing to the logarithmic nature of the dB scale.

VEMCorr starts at zero and needs some time to build up, whereas the VEMCorr is caused by variations of a certain sustained activity.

Neither the VEMP nor the VEMCorr differed much for ipsilateral and contralateral stimulation. This suggests that the ipsilateral and the contralateral otolith organ were stimulated more or less symmetrically. By contrast, previous VEMP studies in which 500-Hz tone bursts were applied using a clinical bone vibrator yielded smaller amplitudes and longer latencies on the contralateral side (Rosengren et al. 2011; Welgampola et al. 2003). Smaller amplitudes on the contralateral side would also be expected with regard to the transcranial attenuation known from auditory investigations. The median attenuation level at 500 Hz was 2 dB in a study by Stenfelt (2012) and 4 dB in a study by Snapp et al. (2016). But the variability was huge in both studies, and for a certain percentage of the subjects a negative attenuation level was obtained, i.e., these subjects were more sensitive to contralateral stimulation than to ipsilateral stimulation. It should finally be noted that the head vibrations induced by a hand-held minishaker (as used in the present study) may substantially differ from those induced by a clinical bone vibrator, owing to the much larger mass of the minishaker and the different couplings to the head. Thus, the differences between ipsilateral and contralateral stimulation are possibly less pronounced if a minishaker is used rather than a clinical bone vibrator. This view is supported by acceleration measurements for 500 Hz tone bursts applied with a minishaker (Dennis et al. 2016), which showed basically the same amplitudes on the ipsilateral and the contralateral side.

Perspectives

While VEMP measurements require a series of transient sounds as the stimulus, the VEMCorr methodology developed in this article provides the possibility to obtain closely related results using continuous noise stimulation. With respect to future applications of the VEMCorr, it is of particular interest that a continuous-stimulation paradigm allows for varying the center frequency of the stimulus without changing the relative bandwidth. Thus, the VEMCorr appears to be ideally suited for investigating frequency tuning. That topic was already investigated in several VEMP studies (e.g., Govender et al. 2016; Todd et al. 2000, 2009; Wei et al. 2013; Welgampola and Colebatch 2001). However, the transient stimuli used in those studies naturally had a relatively broad frequency spectrum, at least at low frequencies, where the total duration of the stimulus sometimes corresponded to only a single cycle of a tone so that the window function used to limit the stimulus

duration had a significant effect as well. Such problems do not occur in VEMCorr measurements. Another potential advantage of the VEMCorr is that by using low-frequency stimuli, it might be possible to reduce the co-activation of the auditory system, so that the investigation becomes more comfortable for the test person or patient.

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