MULTI–CHANNEL WIENER FILTERING BASED AUDITORY STEADY-STATE RESPONSE DETECTION

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ABSTRACT

The detection of auditory steady–state responses (ASSRs) provides an objective and frequency specific technique to assess reliable hearing thresholds at audiometric frequencies. Unfortunately, the duration of ASSR measurements can be long, which is unpractical for wide scale clinical application. Therefore, we propose a multi–channel Wiener filtering (MWF) based technique with a priori knowledge through LQ factorisation as a tool to improve the ASSR detection in recorded multi–channel electroencephalogram (EEG) data obtained at intensities above hearing threshold. We conclude that this technique is able to reduce measurement duration significantly. For a multi–channel data set and implementation, near–optimal performance is obtained with five–channel recordings.

***Index Terms*—** multi–channel Wiener filtering, auditory steady-state response, electroencephalogram, hearing threshold estimation

1. INTRODUCTION

Children with hearing problems have to be diagnosed and supported by hearing aids as early as possible [1]. Therefore, the hearing capability of 96.5 % of all newborns in Flanders (Belgium) is effectively screened by the Flemish child care organisation since 1998. For the fitting of a hearing aid and for general diagnostic purposes, it is however necessary to quantify the hearing thresholds of these children in an objective way. In the late eighties, several techniques using *Auditory Steady–State Responses* (ASSRs) were developed to perform objective and frequency specific threshold measurements. ASSRs are faint evoked electrical responses of the brain [2]. These responses can be elicited by amplitude– and/or frequency–modulated (AM/FM) pure tones. If a tested carrier frequency is modulated with a lower modulation frequency, the appearance of this modulation frequency in the monitored electroencephalogram (EEG) is a strong indication that the subject has effectively perceived the carrier. Several studies have shown that modulation frequencies above 80 Hz lend themselves well to audiometry, especially with young children [3], [4].

Unfortunately, due to the duration of a complete hearing threshold assessment (about one hour), the ASSR technique is impractical in a clinical environment. The main reason for this long measurement duration is the low signal–to–noise ratio (SNR): with magnitudes in the order of nanovolts, the ASSRs are significantly smaller than the surrounding noise originating from the brain and e.g. muscle artefacts. Present day single–channel measurement setups [5] do not seem to allow for a further performance improvement by using more advanced signal processing techniques. However, a newly–built multi–channel setup opens up a new array of techniques to improve the SNR and thus reduce the measurement duration. One such technique is presented here, based on multi–channel Wiener filtering (MWF), which is extended with a priori knowledge incorporation through an LQ factorisation procedure.

This paper is organised as follows: In Section 2, we describe the extended multi–channel Wiener filtering technique in detail. In Section 3, we survey the used experimental setup and analysis methods. Section 4 is dedicated to the obtained results. In Section 5, a comparison is made with independent component analysis applied to the same data set. Finally, conclusions are given in Section 6.

2. MULTI–CHANNEL WIENER FILTERING WITH LQ FACTORISATION

This section describes a multi–channel Wiener filtering (MWF) technique that incorporates a priori knowledge through LQ factorisation. The sinusoidal nature of the ASSR makes it possible to search for a specific frequency, equal to the modulation frequency used in the stimulus.

Without loss of generality, we first focus on the detection of one modulation frequency. When an $n \times m$ measurement matrix $\mathbf{M}$ with n EEG channels of m samples is available,

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this matrix can be transformed to a new signal–plus–noise representation with an orthogonal basis. Each channel of $\mathbf{M}$ contains the scaled version of the same ASSR at the known modulation frequency f_{mod}. This is referred to as the signal space component. The components of the noise space are assumed to be orthogonal to the signal space, which consists of a plane formed by a sinusoid $\mathbf{s}_1$ and cosinusoid $\mathbf{s}_2$ with known frequency f_{mod}. This condition is imposed using an LQ factorisation. $\mathbf{L}$ is lower triangle, $\mathbf{Q}$ orthogonal ($\mathbf{Q}^T\mathbf{Q} = \mathbf{I}$).

$$\begin{bmatrix} \mathbf{s}_1^{1\times m} \\ \mathbf{s}_2^{1\times m} \\ \mathbf{M}^{n\times m} \end{bmatrix} = \mathbf{L}^{(n+2)\times(n+2)}\mathbf{Q}^{(n+2)\times m} \tag{1}$$

with

$$\mathbf{L} = \begin{bmatrix} 1 & 0 & \mathbf{0} \\ 0 & 1 & \mathbf{0} \\ \mathbf{d}_1 & \mathbf{d}_2 & \mathbf{L}_*^{n\times n} \end{bmatrix}, \qquad \mathbf{Q} = \begin{bmatrix} \mathbf{s}_1 \\ \mathbf{s}_2 \\ \mathbf{Q}_*^{n\times m} \end{bmatrix} \tag{2}$$

In this way, $\mathbf{M}$ can be written as:

$$\begin{aligned} \mathbf{M} &= \begin{bmatrix} \mathbf{d}_1 & \mathbf{d}_2 \end{bmatrix} \begin{bmatrix} \mathbf{s}_1 \\ \mathbf{s}_2 \end{bmatrix} + \mathbf{L}_*\mathbf{Q}_* & (3) \\ &= \mathbf{DS} + \mathbf{N} & (4) \end{aligned}$$

It is seen that $\mathbf{M}$ has a contribution of signals $\mathbf{s}_1$ and $\mathbf{s}_2$, with corresponding steering vectors $\mathbf{d}_1$ and $\mathbf{d}_2$, plus a noise contribution $\mathbf{N}$.

A Minimum Mean Squared Error (MMSE) criterium for weighting vector $\mathbf{w}$ can be designed as

$$\min_w \|\mathbf{w}^T\mathbf{M} - \mathbf{D}(1,:)\mathbf{S}\|_2^2 \tag{5}$$

with $\mathbf{w}^T\mathbf{M}$ as the 'filter output' and $\mathbf{D}(1,:)\mathbf{S}$, the useful signal in the (arbitrary) first electrode, as the 'desired response'.

The solution of this MMSE problem can be written as an output–SNR optimisation problem. This corresponds to the knowledge that an MWF can be written as the product of the weight vector of the Minimum Variance Distortionless Response (MVDR) beamformer, and a real–valued scalar factor [6]:

$$\begin{aligned} \max_w \frac{\|\mathbf{w}^T\mathbf{DS}\|_2^2}{\|\mathbf{w}^T\mathbf{N}\|_2^2} &= \max_w \frac{\mathbf{w}^T\mathbf{DSS}^T\mathbf{D}^T\mathbf{w}}{\mathbf{w}^T\mathbf{NN}^T\mathbf{w}} & (6) \\ &= \max_w \frac{\mathbf{w}^T\mathbf{DD}^T\mathbf{w}}{\mathbf{w}^T\mathbf{L}_*\mathbf{L}_*^T\mathbf{w}} & (7) \\ &= \max_{\overline{w}} \frac{\overline{\mathbf{w}}^T\mathbf{L}_*^{-1}\mathbf{DD}^T\mathbf{L}_*^{-T}\overline{\mathbf{w}}}{\overline{\mathbf{w}}^T\overline{\mathbf{w}}} & (8) \end{aligned}$$

A solution for $\overline{\mathbf{w}} = \mathbf{L}_*^T\mathbf{w}$ can be found as the eigenvector corresponding to the largest eigenvalue of $\mathbf{L}_*^{-1}\mathbf{DD}^T\mathbf{L}_*^{-T}$ in (8). The multi–channel measurement matrix $\mathbf{M}$ can now be transformed to a single–channel measurement vector $\mathbf{m} = \mathbf{w}^T\mathbf{M}$ with

$$\mathbf{w} = \mathbf{L}_*^{-T} * (largest\ eigenvector\ of\ \mathbf{L}_*^{-1}\mathbf{DD}^T\mathbf{L}_*^{-T}) \tag{9}$$

The weight vector $\mathbf{w}$ thus recombines the rows of measurement matrix $\mathbf{M}$ into one channel that maximises the SNR for the signal component with the specified modulation frequency f_{mod}.

This method can be extended to more than one modulation frequency. By introducing extra $\mathbf{s}_i$ with other frequencies in equation (1), the output channel can be optimised in SNR–sense for all defined $\mathbf{s}_i$ (where (9) is recomputed for each individual modulation frequency).

3. EXPERIMENTAL SETUP

Eight normal–hearing subjects were tested in a sound–proof Faraday cage. The auditory stimuli consisted of four carriers at audiometric frequencies (0.5, 1, 2 and 4 kHz) for each ear, which were separately 100% amplitude modulated with a modulation frequency from 82 up to 110 Hz. This resulted in eight different modulation frequencies that have to be detected in the EEG. The stimuli were applied at four different intensities (60, 50, 40 and 30 dBSPL respectively) for a period of 48 sweeps (± 13 minutes) per intensity. Seven–channel EEG measurements were conducted by placing nine electrodes on the surface of the skull on the following positions, in accordance with the international 10–20 system [7] (active electrodes 1 to 7, common and ground electrode respectively): Oz, P4, P3, Cz, F4, F3, Pz, forehead, left mastoid. The single–channel reference method is created by the Cz–Oz difference.

The EEGs were amplified, and recorded using a multi–channel sound card. Processing was conducted offline: the signals were downsampled and artefacts were rejected. The n-channel recordings ($n \leq 7$) were divided in sweeps and each sweep was averaged with all preceding sweeps from the same channel to raise the SNR–level. The algorithm took such an averaged n-channel sweep as its input and linearly combined these channels to produce a one–channel output (see Section 2). An F–test was conducted on each modulation frequency using its F–ratio and calculating its F–value [5].

In order to evaluate the single–channel and multi–channel technique, receiver operating characteristic (ROC) curves were calculated from 8 subjects [8], [9]. The ROC–curves were constructed using the data at 16 modulation frequencies of which 8 were used as control frequencies, as in these frequencies it was known that only noise was present. The area under the curve was used as a measure of detection accuracy. The larger the area, the higher the chance a decision by the algorithm is correct. A point of reference was created by determining the behavioural hearing thresholds of the subjects. The above calculations were carried out each time an additional sweep was collected and averaged with previous sweeps, so that the performance could also be analysed on a time based scale. The ROC areas where finally compared statistically using the Z–test [10].

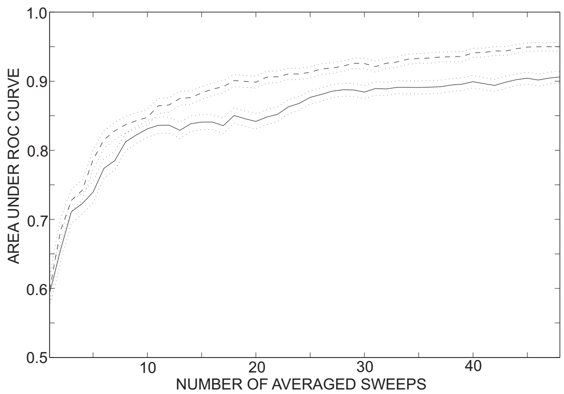

Fig. 1. 'area under ROC–curve' versus 'number of averaged sweeps': MWF procedure applied to a seven–channel data set (dashed), single–channel reference method (solid). The dotted lines denote two standard deviations.

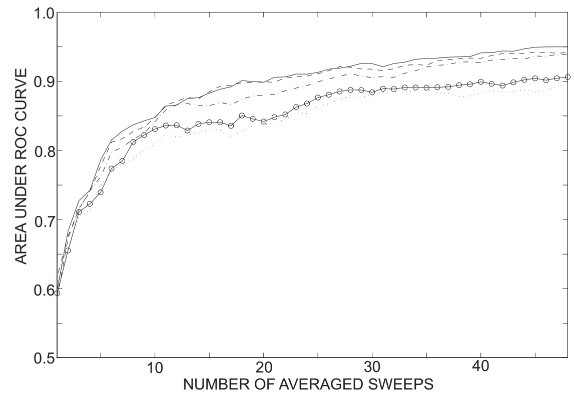

Fig. 2. 'area under ROC–curve' versus 'number of averaged sweeps'. MWF procedure applied to an n–channel data set, without the use of extra channels: $n = 7$ (solid), $n = 5$ (dashed), $n = 4$ (dashdot), $n = 2$ (dotted), reference method (solid–circle).

Table 1. Measurement time reduction (in %) per subject, intensity and carrier frequency for the MWF procedure applied to a seven–channel data set. Figures relative to the single–channel reference method. A response is considered present if it is significant for 8 consecutive sweeps. Significance is reached at p = 0.050 for the single–channel reference method and p = 0.00025 for the MWF method. Sensitivity is equal to 95.0 % for both methods.

subj #	1	2	3	4	5	6	7	8
%	9	4	12	5	2	-9	62	22
int (dBSPL)	30		40		50		60	
%	9		14		14		37	
carrier (Hz)	500		1000		2000		4000	
%	1		25		23		15	

4. RESULTS

Fig. 1 shows the results of the MWF procedure on a seven–channel EEG data set. For a sufficiently large number of sweeps, a significant performance increase is obtained compared to the single–channel approach. A problem with this configuration is its poor performance for a small number of sweeps (left–hand side of Fig. 1). The calculation of the eigenvectors is found to be sensitive to noise.

When the results per subject are considered in Table 1, we observe an average measurement duration reduction of 13.4 %, varying between a 9 % measurement duration increase and a possible 62 % decrease. Higher intensities and frequencies with physically larger responses (e.g. with 1 and 2 kHz carriers) are more prone to faster detection. The one case with the measurement duration increase lies closely to the noise floor of 6.6 % (2σ around mean measurement duration reduction of 0 % for noise frequencies).

While measurement duration can be reduced significantly with the multi–channel approach, connecting many electrodes to the subject's head is not very practical. To reduce preparation time, it is considered an advantage if a method requires fewer measurement electrodes. Fig. 2 shows that performance is only marginally compromised by using five instead of seven channels for this data set. Adding extra (i.e. more than five) simultaneous EEG channels does not further reduce measurement duration. It is expected that a more efficient (non–symmetric) electrode placement may increase performance beyond five channels.

5. COMPARISON WITH INDEPENDENT COMPONENT ANALYSIS

The application of independent component analysis (ICA) on ASSR measurements already showed multi–channel techniques provide performance benefits over single–channel methods [11], [12]. In contrast with the MWF procedure, that exploits knowledge of the known modulation frequency, ICA does not use any prior information except for the assumption of independence of the underlying non–Gaussian sources. Nevertheless, it can not be proven with this data set that the MWF procedure performs significantly better than ICA. However, a benefit of the MWF technique is that a considerably smaller number of code lines is needed and thus less operations per processed sweep. This is appealing for realtime processing.

As both techniques are essentially source separation techniques, they are rather similar. In literature, the link between (generalised) eigenvalue decomposition and blind source separation is already known [13]. The poor performance for lower

SNR values can be observed from the results of both methods and is confirmed by ICA theory [14]. The saturation effect for five channels is also present in the same, ICA processed, data set [11], [12]. The cited works additionally show that performance can be further increased by introducing extra artificial channels. Similar behaviour is observed with the MWF procedure.

6. CONCLUSION

A significant improvement in detection speed with a mean of 13.4 % is possible when recorded multi–channel EEG data at intensities above hearing threshold are preprocessed using a multi–channel Wiener filtering (MWF) based technique with a priori knowledge through LQ factorisation. However, the MWF technique is performing poor when applied to low SNR data. For this data set, a saturation level is reached at five channels (seven electrodes). Results are similar to those from the same, ICA processed, data set [11], [12].

7. REFERENCES

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