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Real-time EEG-based mental fatigue estimation model

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INTRODUCTION

Mental fatigue is a serious problem in many professions, such as commercial drivers, pilots, and train operators. It affects cognitive functions, which may lead to reduced attention or episodes of microsleep.

Current fatigue monitoring methods are often subjective and unreliable. There is a need for an objective, real-time method of fatigue estimation from biological signals, such as electrical brain activity.

The aim of this study was to develop an appropriate methodology for real-time EEG-based mental fatigue estimation and a suitable Matlab Simulink model.

MATERIALS, METHODS & IMPLEMENTATION

Six healthy volunteers, including one female and five males, aged 33±9 years, participated in the experiment. EEG signals were measured using the g.NAUTILUS RESEARCH device, the g.SAHARA model, developed by g.tec. It is a wireless 64-channel EEG device using dry electrodes. Preset filters for the measurement were set to 0.5 – 30 Hz to specify the frequency band of brain activities. A suitable fatigue index, beta/(theta+alpha) ratio, was estimated from the periodogram, calculated by Welch's method. The Fz electrode, positioned in the middle of the frontal region, was selected as the electrode of interest. A suitable model was conducted in Matlab Simulink. The brain activity is highly individual for each person. Our solution was to conduct three measurements: two calibration measurements for calculating the Alert and Tired state thresholds, and the main measurement for testing the developed model. 6 minutes long videos were chosen for the calibration measurements, and the fatigue index was calculated only from the sixth minute. 30 minutes long video of monotonous highway driving was selected for the main measurement, and the fatigue index was calculated every 10 seconds of the measurement. The fatigue index was recorded five minutes after the start and five minutes before the end of the video. For the comparison of objective and subjective results, the subjective Stanford Sleepiness Scale (SSS) was completed from all participants.

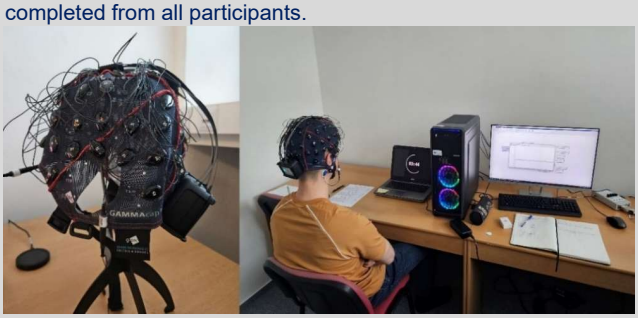

Fig. 1: The EEG device from g.tec used for measurements

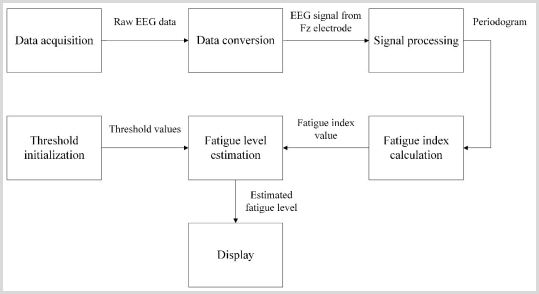

Fig. 2: Block diagram for real-time fatigue monitoring and estimation

RESULTS

The mental fatigue level was presented in the form of three text messages. The messages were displayed based on a comparison of the estimated fatigue index with threshold values. Alert is evaluated if: (fatigue index ≥ Alert threshold), Tired is evaluated if: (fatigue index ≤ Tired threshold), Partially alert is evaluated if: (Tired threshold < fatigue index < Alert threshold). The fatigue index decreased in all participants except Participant 1, and Participant 6. The increase could have been caused by the high fatigue index variability due to the short time window from which the fatigue index was evaluated. While the fatigue index results were similar with small deviations in each participant, subjective perception of fatigue led to a large variation in SSS results. Although the fatigue index results and the chosen SSS level are partially correlated, it is not possible to determine universal thresholds due to the subjective perception of mental fatigue.

Table 1: The participants results for the fatigue index and SSS levels; *C1* and *C2* – Calibration measurements, *M* – Main measurement

Participants	Alert threshold	SSS level *C1*		Tired threshold	SSS level *C2*		Five minutes after start	Five minutes before end	SSS level *M*	
1	0.600	1	2	0.194	2	3	0.539 (Partially alert)	0.841 (Alert)	1	4
2	0.904	2	3	0.839	1	3	0.820 (Tired)	0.764 (Tired)	2	4
3	1.025	1	2	0.799	2	4	0.974 (Partially alert)	0.802 (Partially alert)	3	5
4	0.816	1	1	0.763	3	4	0.713 (Tired)	0.711 (Tired)	3	6
5	0.902	2	2	0.792	3	3	0.890 (Partially alert)	0.637 (Tired)	3	4
6	0.819	1	3	0.783	4	5	0.673 (Tired)	0.680 (Tired)	2	4

CONCLUSION & IMPLEMENTATION PATH

A method for real-time mental fatigue estimation and a model for detecting and determining mental fatigue levels were developed. A key advantage of the model is its flexibility in setting initial conditions, estimation speed, and simplicity. Our model is suitable for measurements in laboratory conditions. Future work should involve the model upgrade for motion artifacts or any other adverse effects removal and testing the model on a larger group of volunteers. A cross-group comparison (sex, age, health status) should also be included. Experimentation with the window length for PSD calculation is also the object of future research. Simplifying the device to use only a single selected electrode should also be considered. It may reduce robustness and increase the convenience of measurements. This approach could facilitate the development of a final product suitable for commercial use in high risk professions requiring active alertness.

ACKNOWLEDGEMENTS

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