Head Pulsation Signal Analysis for 3-Axis Head-Worn Accelerometers

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Abstract

Previously, using single-axis accelerometers, it has been proposed that in conditions such as traumatic brain injury (TBI) the brain pulsation signal characteristics change, potentially due to changes induced by the impact to the brain. In this paper, we aim to validate the use of a custom built embedded measurement system towards the analysis of the head pulsation signals. The system comprises of several synchronized high sampling rate 3-axis accelerometers and a simultaneous chest ECG. In our case three accelerometers on the surface of human head are used (in left temple, forehead and right temple), while the subject were in supine position. To illustrate that a proper signal quality may be extracted, we derive heart rate (HR) and heart rate variability (HRV) from each sensor and each axis for each of five healthy male volunteers. The results are reported against ECG as the ground truth. This study will build ground for further clinical trial utilizing multi-axial accelerometers to study both healthy and diseased subjects (e.g. TBI patients).

1. Introduction

The physiological monitoring of the brain is a challenge since the brain is sealed inside the skull-bone to protect it from impacts. Brain of a living human being is also highly vulnerable, which makes obtaining reliable measurement data through any invasive procedure a challenge. Current non-invasive standard imaging modalities for the brain (MRI, PET, CT) [1–3] typically require large infrastructure and/or cannot be used in very high imaging rate continuous-time mode. Development of non-invasive means to study of the brain to estimate intracranial pressure and other brain dynamics would possess an extemely high value both in scientific and clinical viewpoints [4]. For instance, patients suffering from traumatic brain injury (TBI) or patients suffering from intracranial hemorrhage or stroke would largely be benefited by a continuous-time non-invasive modality enabling fast and automatic reaction to a worsening in the patient's condition.

Transcranial Doppler (TCD) ultrasound a feasible alternative to monitor the brain, but its acquisition and interpretation are usually conducted by a doctor and thus the method is not automatic [5]. Invasive procedures to study the brain can provide continuous-time output (which can be processed automatically), but these methods typically require drilling a hole through the skull bone and placing a catheter to the brain causing infection risk. Attaching single-axis accelerometers on the surface of the skull have been proposed as a novel non-invasive modality to measure some of the changes in a patient with abnormal brain function [6–8]. A new continuous-time operation (e.g. 24 h) capable modality enabling evaluation and prediction of patient's condition automatically would be highly beneficial for patients suffering from severe TBI, for example, since even moving these patients to the imaging facilities (MRI, PET, CT) [1–3] in hospital causes a significant risk for recovery.

Related fields in the area of physiological monitoring to this study include seismocardiography (SCG) [9] and ballistocardiography (BCG) [10] which measure the heart induced mechanical motion in chest and in overall body (e.g. using a bed attached sensor), respectively. In this study, we evaluate the feasibility of using 3-axis accelerometers for head pulsation signal acquisition from five healthy male volunteers. To our knowledge, multi-axes accelerometers have not been used to this purpose previously in the clinical settings. This study can be seen as a pre-clinical assessment of the feasibility of using a similar set-up to measure brain diseased patients (e.g. TBI).

2. Methods

The measurement set-up was such that three accelerometers (ADXL355) were attached to the head of the test subjects who were in supine position (to the left and right temple and to the forehead). The measurement system characteristics have been previously described in [11] in detail. The accelerometer measurement units were attached using a double-sided tape to the head. In [7] a special helmet was designed to attach the sensors. Simultaneous refer-

ence ECG was also recorded using the system. The sampling rate of the sensors can be tuned manually, and we used a fixed 1 kHz sampling rate for all the sensors. The output signals were all synchronized in time to the same temporal locations by the system, including all accelerometer axes (X, Y and Z) and ECG. The original length of the signals was 5 minutes. However, despite of very good signal quality, in order to remove some of the motion artifacts we used only 3 minutes length uniform signal segments in the analyses. Prior to all analyses the signals i.e. each accelerometer axis and ECG were filtered with a Butterworth bandpass filter to obtain the frequency band between 1 Hz - 45 Hz.

2.1. Visual analysis of signal quality

In Fig. 1 examples of signals obtained from test subject no. 2 are shown. It can be observed, that according to visual inspection the signals are of very good quality potentially enabling the estimation of cardiac time intervals and amplitudes. However, due to limited number of participants in this study we did not perform any statistical evaluation of these intervals.

2.2. Heart rate and heart rate variability estimation

Heart rate (HR) and heart rate variability (HRV) were estimated from each subject, each sensor and axis. The method used was single-axis autocorrelation presented previously in [12, 13]. Both HR and HRV can be extracted by the method. The method outputs the RR interval lengths. These are obtained by autocorrelation between short (2.5 s) length subwindows which are repeated in 1 second intervals (overlap 1.5 seconds) through the whole signal. As the output of the method is RR interval series, it needs to be converted to related HR and HRV values. The average beat duration (RR interval) is simply calculated as the median of the RR intervals divided by the sampling rate (RR intervals are in number of samples at 1 kHz). The HR in beats-per-minute is calculated as 60 divided by the median RR interval (in seconds). Similarly, HRV is estimated as the median between the absolute values of the successive differences of the elements in the RR time series [12]. Finally, HRV is divided by the sampling rate (1 kHz) to obtain it in time units (seconds) instead of samples.

3. Experimental results

The quantitative results are shown in Table 1. In this study there were five subjects and three accelerometers for each subject. For each accelerometer there were 3 axes (X, Y, Z) of which the mean is calculated to the Table 1. Thus, in total there were 15 recordings and the single axis autocorrelation was applied thus 15*3 (the number of axes) times in total. It can be observed, that for all recordings (15 of 15) the deviation of the mean HR against ECG was below or equal to 1 bpm and for 11 best recordings the deviation in mean HRV against ECG was below or equal to 10 ms.

HEART RATE AND HEART RATE VARIABILITY vs. ECG					
	Deviation (E)	**E_HR (ECG HR)**	**E_HRV duration**	**E_SDNN (best)**	**E_RMSSD (best)**
	from ECG	**(in bpm)**	**(in ms)**	**(in ms)**	**(in ms)**
SUBJ1	LT	0.1 (38.8)	10	1.1	0.8
	FO	0.1 (38.8)	4	6.2	2.3
	RT	0.2 (38.8)	2	0.7	3
SUBJ2	LT	0.8 (61.4)	53	1.5	2.9
	FO	0.7 (61.4)	54	2	5.2
	RT	0.6 (61.4)	8	0.6	1.2
SUBJ3	LT	0.1 (78.9)	0	5.2	12.3
	FO	0.2 (78.9)	1	3.2	2.8
	RT	0.1 (78.9)	1	1.9	1
SUBJ4	LT	0.1 (86.6)	1	11.3	20.4
	FO	0.1 (86.6)	44	1.6	3.1
	RT	0.2 (86.6)	2	1.3	3
SUBJ5	LT	0.5 (73.9)	32	3.6	1.4
	FO	0 (73.9)	3	3.4	1.6
	RT	0 (73.9)	3	3	1.7

Table 1. Results from the head measurements. The deviations from ECG (error) is only reported. Three measurement locations LT, FO and RT were the left temple, forehead and the right temple, respectively. Test subjects (SUBJ) are indicated with numbers 1-5. The average HR (in bpm) was derived as the mean of all three accelerometer axes. The HRV duration's value (first from the left) was calculated (in Matlab) as the median of the differences between the beat intervals (beat intervals were obtained with single axis autocorrelation algorithm). The deviation between the HRV duration and HRV duration estimated from ECG is reported. For remaining two HRV measures - SDNN and RMSSD (extracted using the Kubios software) - the best axis's deviation from ECG's estimate is shown.

A sampling rate parameter of the autocorrelation, which generally determines the observable HR range was adjusted separately for subject no. 1 due to very low HR, within the lower limit of the operation range of the autocorrelation method [12]. Otherwise the same parameter values were used for all the subjects. This exception was seen justified, as the visually observable signal quality was very good for all the subjects. In Table I also the best axis's SDNN (standard deviation of the beat-to-beat time intervals) and RMSSD (root mean square of successive differences) for each individual and sensor position are shown. These measures were extracted using the Kubios software [14] based on beat intervals obtained with autocorrelation.

4. Discussion

As described, the signal quality in this study was very good, enabling the detection of HR and HRV very accurately. Even the identification of individual heart beats seems feasible based on our visual analysis. The subjects in this case were young healthy adults and the signal qual-

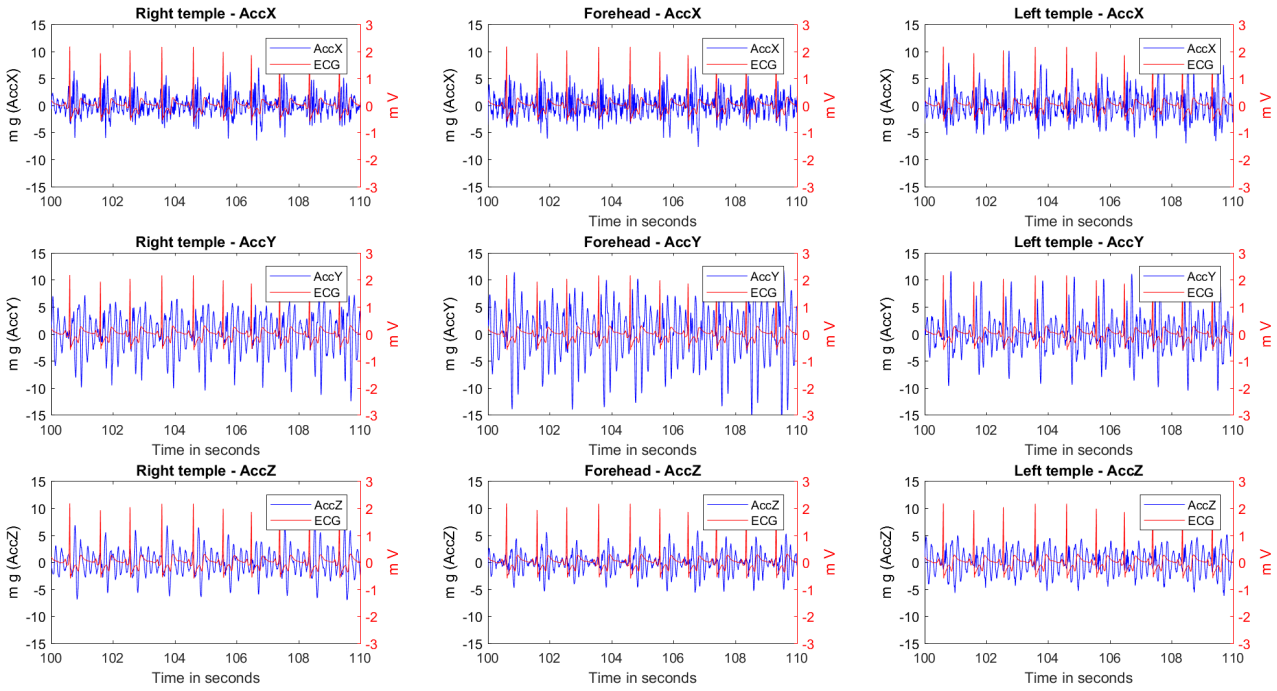

Figure 1. Example signals from subject no. 2. Signals from each three axis and each location of the head are shown. Simultaneously captured ECG which is the same in each signal is shown as a reference. The sampling rate of the system (accelerometer and ECG) is 1 kHz.

ity for older subjects, for instance, might be lower. Potentially in the actual clinical trial, if implemented in the future, it might be reasonable to initially exclude the patients with diseased heart (e.g. reduced ejection fraction) from the study. As TBI patients are often relatively young, this exclusion criteria should not limit the number of available study participants significantly. Another exclusion criteria could be severe injuries in the facial area, which could complicate the attachment of the sensors.

We also performed similar test for the subjects while they were standing. The signal quality in that case was very low for some of the persons. Therefore, those signals are not included to this study. In addition to the sensors attached to the head, we initially applied also two sensors (3-axis accelerometers) at the subject's neck (left and right side) and one sensor on the sternum. Also the signal quality of these sensors was very good.

As the patients were lying on a bed, the BCG signal from the bed may propagate significantly to the signal obtained from the head. It is also possible the patient's head's motion is at least partly induced from the chest through the body (and neck) to the head. A potential direction to study these issues further would be to obtain an additional signal directly from the subjects bed or fast/fix the test subject's head some way. Furthermore, whether the signal information we obtained with accelerometer is really related to the blood circulation at the surface of the skull or the cerebral circulatioin inside the skull is still an open issue. In [6–8] the assumption was that brain pulsations could be captured from the head surface. It has been recently observed that brain related abnormalities such as concussion may also affect to the autonomous regulation of the heart (e.g. HRV) [15]. How these affect to the autonomous regulation of the heart is, thus, an emerging research topic, which could be studied as part of our future research.

5. Conclusion

It is concluded that the used measurement system is capable of attaining the head pulsation signal from the individuals included to this pre-clinical study. Future work includes conducting a clinical trial with e.g. TBI patients. We expect, that potentially the non-invasive head pulsation signals could be utilized to find information indicative of the patient's condition.

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