Brief Communication

Using photoplethysmography in heart rate monitoring of patients with epilepsy

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ABSTRACT

Heart rate is a useful neurophysiological sign when monitoring seizures in patients with epilepsy. In an ambulatory setting, heart rate is measured with ECG involving electrodes on the skin. This method is uncomfortable which is burdensome for patients and is sensitive to motion artifacts, which decrease the usability of measurements. In this study, green light photoplethysmography, an optical technique arising from the fitness industry, was evaluated for usefulness in a medical setting. Simultaneous overnight measurements of HR with a commercially available optical heart rate (OHR) sensor and with ECG (HRECG) were performed in 7 patients with epilepsy. Overall, there was no significant difference between OHR and HRECG in random 10-minute periods during wakefulness (p = 0.69) and sleep (p = 1.00). The Bland–Altman analysis showed negligible mean differences. Limits of agreement were higher during wakefulness and during the occurrence of two seizures possibly because of less reliable HRECG measurements due to motion artifacts. Optical heart rate seems less sensitive to these motion artifacts, and measurements are more user-friendly. The optical heart rate sensor may fill the gap of systems for ambulatory heart rate monitoring and can be especially useful in the context of seizure detection in patients with epilepsy.

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1. Introduction

Wearable devices for ambulatory monitoring of heart rate (HR) are of interest in a medical setting. Heart rate monitoring can, for instance, be used to facilitate safe and effective exercise in patients with a history of cardiovascular disease. A more specific example of ambulatory monitoring is the automatic detection of seizures in patients with epilepsy. Though most seizures are self-limiting, seizures can be harmful when unnoticed. The occurrence of sudden unexpected death in epilepsy (SUDEP) is usually related to nocturnal, unmonitored seizures [1]. The vast majority of seizures are associated with a heart rate increase [2], which can, in turn, be used for seizure detection [3].

In a clinical setting, heart rate is mainly obtained through automatic analysis of an electrocardiography (ECG) signal measured with at least two electrodes attached to the skin. This is uncomfortable and can lead to skin irritation. Electrocardiography signals also tend to become noisy when patients move, which is often the case during seizures. The fitness industry faces a similar problem; heart rate needs to be assessed during exercise in a user-friendly and precise manner. A

revolution in this industry is the use of green light reflection photoplethysmography (PPG) which can be measured with a watch worn on the wrist. With PPG, heart rate is detected through measurement of changes in blood volume using either transmission or reflection of specific wavelengths of light [4].

In this study, we investigated the usability of a readily available optical heart rate sensor from the fitness industry for heart rate monitoring in a medical setting. We compared the performance of the optical heart rate sensor with that of the ECG in ambulatory monitoring of patients with epilepsy.
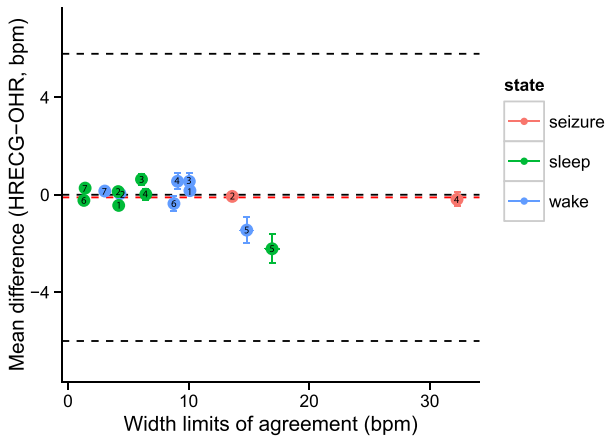
2. Methods

2.1. Measurements

Between September 1st 2013 and September 30th 2013, 7 patients in the Epileptic Monitoring Unit (EMU) of epilepsy center Kempenhaeghe (Heeze, The Netherlands) undergoing continuous video-EEG monitoring were asked to wear the MIO alpha heart rate monitor (Physical Enterprises Inc.)© sports watch. This sensor system contains an optical heart rate (OHR) sensor based on green plethysmography. The watch was placed around the upper arm with a custom-made elastic textile strap. Written informed consent was obtained. Optical heart rate (in beats per minute, bpm) was sent to an Ipod 5© via Bluetooth© 4 (frequency:

Fig. 1. Scatter plot of mean differences plotted against the width between limits of agreement (4 times SD of difference) for each patient (numbered) during wakefulness and sleep and during the occurrence of two seizures. Error bars show precision of the estimated mean differences (SE). The red dashed line indicates the overall mean difference. Overall limits of agreement based on the Bland–Altman analysis for repeated measures are shown at the upper black dashed line and at the lower black dashed line.

1 Hz) and was recorded with the Wahoo© app. Electrocardiography data were obtained with two supraclavicular ECG leads (left and right). The ECG, OHR sensor, and iPod 5 were synchronized manually. Patients were asked if the OHR sensor had caused discomfort.

2.2. Signal preprocessing

The MIO sensor emitted the OHR signal at 1 Hz. The method for derivation of this OHR from the raw PPG signal was not made public by the manufacturers. Heart rate was extracted offline from the ECG data using a MATLAB implementation of Afonso et al. [5]. Outliers (heart rate $> \pm 30\%$ of previous heart rate) were replaced by the average of the previous two heartbeats. Heart rate from the ECG (HRECG)

was smoothed using a 10-sample moving average, interpolated, and resampled at 1 Hz (to match the sample rate of the OHR). Heart rate from the ECG and optical heart rate were aligned using cross-correlation to remove time lags introduced during Bluetooth transmission. For each patient, a 10-minute fragment of OHR was shifted along the HR derived from the ECG. The signals were time-locked where the minimum of the root mean square of the difference between the signals was found.
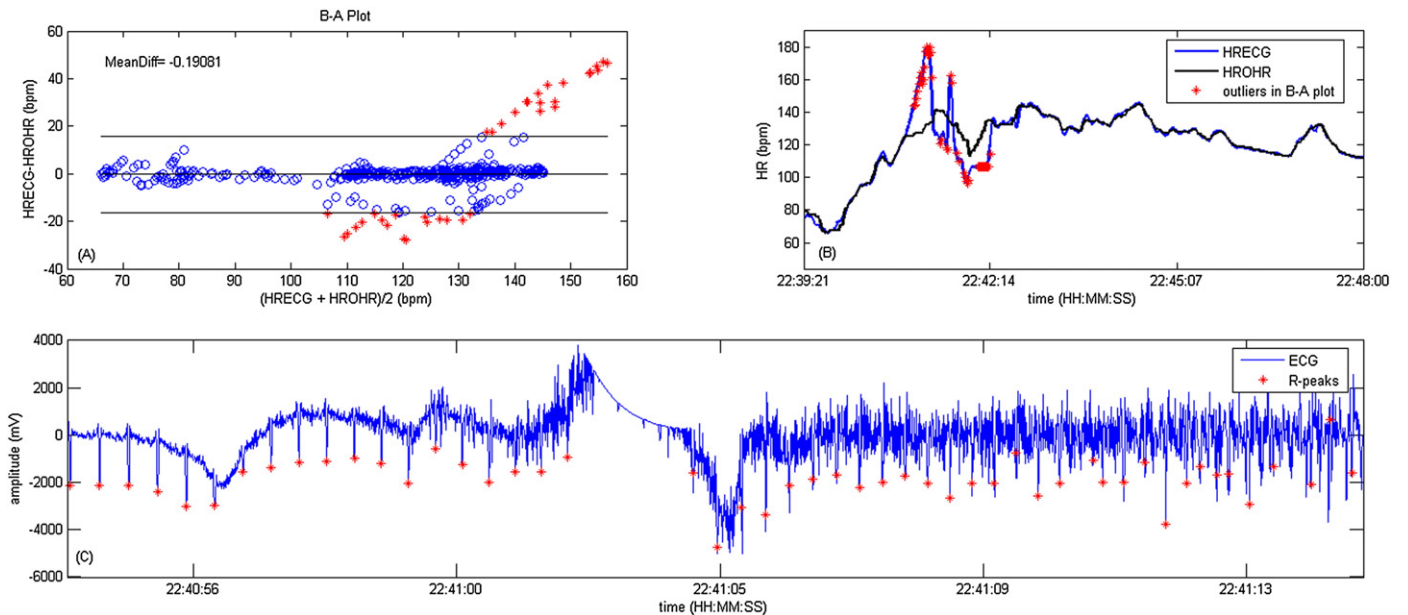
2.3. Analysis

The Bland–Altman analysis for repeated measures on 10-minute time windows in the awake state (around 7 PM) and during sleep (around 12 AM) was used to compare OHR and HRECG [6]. The Mann–Whitney U-test was used to compare the average HRECG and OHR of all patients in a 10-minute window in both activity states. All signal preprocessing and statistical analyses were performed with MATLAB 2011a [7]. The study was approved by the local research ethics committee of the Academic Centre for Epileptology Kempenhaeghe.
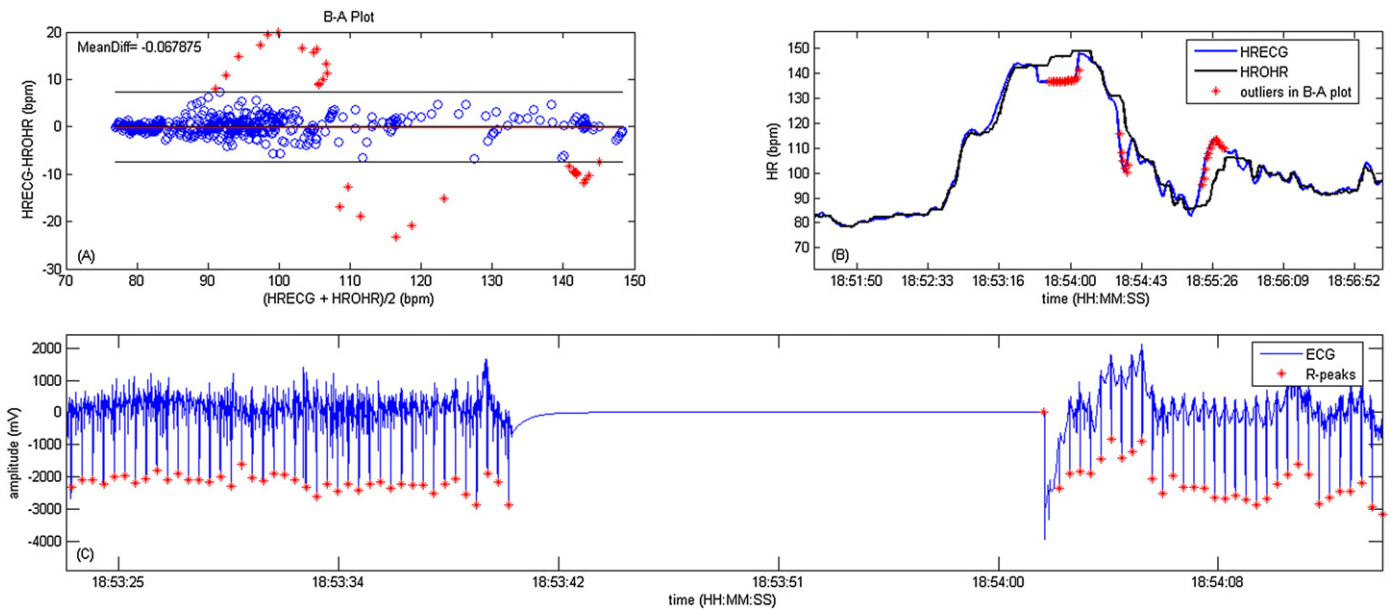
3. Results

All 7 patients (mean age (SD): 33 (13), 4 males) reported that the OHR sensor embodiment was comfortable to wear.

3.1. The Bland–Altman analysis for awake and sleep states

Fig. 1 shows the mean differences between HRECG and OHR and the width between limits of agreement (4 times SD of mean difference) for each patient during wakefulness and sleep and during the occurrence of two seizures. Overall mean difference and overall limits of agreement are also shown. In general, the mean differences during sleep and wakefulness are negligible; however, the variance is higher during the day (significant Levene's test with $p < 0.001$). In patient five, the ECG data were very noisy, which probably explains the high limits of agreement for both time periods.

Fig. 2. (A) The Bland–Altman plot showing the mean difference and limits of agreement for the heart rate fragment containing a tonic–clonic seizure. (B) Heart rate signals acquired with both methods for the same seizure fragment. The depicted continuous lines are based on 1-Hz measurements (one datapoint per second). The red points are the outliers in Fig. 2A. Both techniques show a rising heart rate before and during the tonic phase of the seizure with a plateau during the clonic and postictal phases. During the tonic, preictal, and postictal phases, both heart rate signals are similar. Fig. 2(C) shows the ECG signal during the clonic phase of the seizure. The ECG is distorted and difficult to analyze because of a high intensity of motion.

Fig. 3. (A) The Bland–Altman plot showing the mean difference and limits of agreement for the heart rate fragment containing a partial complex seizure in patient two. There is a negligible bias, and limits of agreement are much lower than what were measured during the tonic–clonic seizure. (B) Heart rate signals acquired with both methods for the same seizure fragment. The depicted continuous lines are based on 1-Hz measurements (one datapoint per second). The red points are the outliers in Fig. 3A. Panel C shows the ECG signal with an electrode connection problem during the seizure leading to a flat line in the ECG. The interpolated heart rate generated outliers in the Bland–Altman plot.

3.2. Heart rate during seizures

Measurements during a tonic–clonic and a complex partial seizure are presented in Figs. 2 and 3. Especially during the tonic–clonic seizure, limits of agreement are high, possibly because the ECG signal is distorted.

3.3. Differences in measurements over all patients

Both in wakefulness (U = 29, p = 0.69) and during sleep (U = 32, p = 1.00), no significant difference in HRECG and OHR over all patients was found.

4. Discussion

In ambulatory monitoring of patients with epilepsy, the heart rate measured with the OHR sensor seems to be equivalent to the heart rate derived from automatic ECG analysis, with mean differences mostly below two beats per minute. The variation in differences between both methods, however, is high during wakefulness and during the occurrence of two seizures included in the data.

A possible explanation for this higher variation is that movement occurring during wakefulness and during seizures leads to artifacts in the ECG and, therefore, less reliable derivation of HRECG. This would be in line with previous studies proposing that green light PPG is less sensitive to motion artifacts compared with ECG [8,9]. When visually appraising the OHR signal during the seizures, we found that it seems to show physiologically realistic measurements. However, as we only had access to the processed heart rate by the OHR sensor, we cannot conclude that these measurements are correct. Comparison with a reliable gold standard is necessary to prove this hypothesis.

Though the OHR sensor is designed to be used as a watch on the wrist, all data were recorded at the upper arm. A previous study showed that measuring green light PPG at the upper arm gave less motion artifacts compared with other locations [8]. Also, a sensor on the upper arm is less obtrusive as it can easily be concealed with clothing.

Considering ambulatory seizure monitoring, we found that the optical heart rate sensor has several advantages over the standard ECG. The lack of electrodes and wires reduces potential skin irritation, risk of losing signal due to electrodes falling off, and unwanted incidents due to wires. These advantages are especially prominent when a patient has a seizure involving high-frequency motion and/or increased sweating.

A limitation of this study is that all results were reported for mean heart rate produced by algorithms embedded in the commercially available optical heart rate sensor. An analysis of the raw PPG signal is necessary to fully assess the usability of PPG for heart rate detection in epilepsy, for instance, by also including measures for heart rate variability. Mean heart rate is a suitable measure to detect tachycardia, bradycardia, and asystole, which are important markers of clinical relevance of seizures. However, in other medical settings, such as the evaluation of arrhythmias [10], more features of heart rate and ECG characteristics such as QT interval, which PPG currently cannot provide, are necessary for proper monitoring. Measurements in other patient groups are necessary to fully assess the usefulness of PPG for cardiac monitoring.

All in all, the optical heart rate sensors may fill the gap of systems for ambulatory monitoring in a setting where the primary measure of interest is the mean of the heart rate. As shown in our measurements of heart rate during the occurrence of two seizures, the OHR sensor can be especially useful in the context of seizure monitoring in patients with epilepsy.

Conflict of interest

None of the authors has a conflict of interest to disclose.

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