On the Way to a Cable Free Operating Theater: An Operating Table with Integrated Multimodal Monitoring

Tobias Wartzek[1], Robert Elfring[2], Arne Janssen[2,3], Benjamin Eilebrecht[1], Marian Walter[1], Steffen Leonhardt[1]

[1]Chair of Medical Information Technology, RWTH Aachen University, Aachen, Germany
[2]Chair of Medical Engineering, RWTH Aachen University, Aachen, Germany
[3]SurgiTAIX AG, Aachen, Germany

Abstract

On the way to a cable free operating theater a new operating table is developed featuring into the table integrated sensors for patient monitoring, which do not necessarily need a direct contact to the naked skin, and a magnetic tracking system. The patient monitoring sensors are new multimodal sensors, which combine a capacitive electric field sensor with a reflective optical sensor and a temperature measurement. This multimodal measurement offers the opportunity of a reliable vital sign measurement by use of redundancy and plausibility checks. A novel artifact detection method for capacitive electrocardiogram measurements based on an optical measurement is presented. Together with an adaptive threshold based algorithm, intervals with artifacts can reliably be identified resulting in a robust estimation of heart rate.

1. Introduction

A common wish of surgeons and anaesthesiologists is the reduction or clearance of cables in an operating theater. A step towards this goal is the integration of a magnetic tracking system and sensors for patient monitoring into an operating table. The magnetic tracking allows to determine the position of medical instruments without any line of view and thus also inside the patient. This integration of tracking and monitoring reduces cost (disposables are avoided), reduces preparation time and increases patient safety.

In the following, we focus on the sensors for patient monitoring. Fig. 1 shows a sketch of the table in which the sensors are visible. To adapt to different patients an array of sensors is integrated into the table. The monitoring consists of a capacitive electrocardiogram (cECG), a novel optical artifact detection and a temperature sensor. The cECG allows the measurement of the ECG without a conductive contact and thus can also measure an ECG through clothes [1]. Although the body core temperature is the important value for physicians, the temperature sensors integrated measure only the skin or surface temperature. But the skin temperature may give further information to physicians and can predict a beginning hypothermia even in low risk operations in which a temperature monitoring would otherwise not be applied. Additionally, assuming a well isolating operating table, the skin temperature will converge towards the core temperature, similar to zero-heat-flux sensors [2].

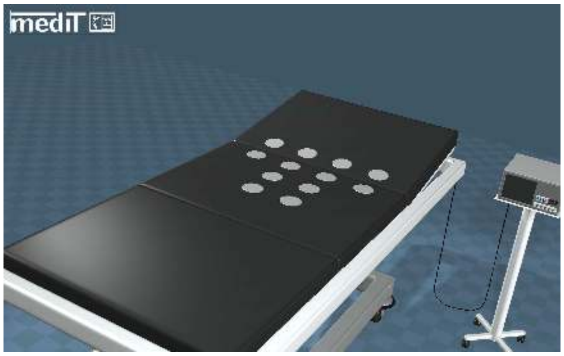

Figure 1. Operating theater table with integrated multimodal sensors.

However, since the electrodes are not tightly fixed to the patient, these measurements suffer from sever motion artifacts which may totally disrupt the signals of interest. To solve this drawback, a method to detect motion artifacts is presented in this paper. The reliable detection of artifacts is a main prerequisite for contactless sensors to achieve acceptance and provide a reliable deduction of further parameters (e.g. heart rate, heart rate variability or temperature trends).

2. Methods

2.1. Multimodal sensor

Fig. 2 shows a picture of the developed multimodal electrode. This sensor comprises a capacitive ECG circuit, a reflective optical measurement and a temperature sensor. It is 7 mm high with a diameter of 46 mm and integrates four integrated circuits.

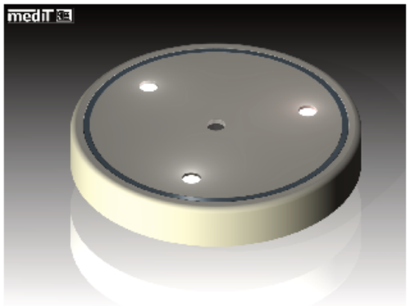

Figure 2. Picture of the multimodal electrode.

The capacitive sensor consists of a voltage follower with an input capacitance cancellation circuit [3, 4]. The optical reflective sensor consists of three infrared LEDs (880 nm) and a photodiode with a transimpedance amplifier. The photodiode is located in the middle of the sensor with a distance to the LEDs of 15 mm. The temperature sensor (ADT7420 from Analog Devices) measures the skin temperature.

Since several physiological parameters are measured in one sensor, redundancy and plausibility checks can be performed which increases the reliability of the measured parameters. The heart rate, for example, can be calculated from both the cECG and the optical signal. Since the temperature measurement requires contact to the body, it can be validated by verifying if the corresponding electrodes can measure an ECG or an optical pulse signal and thus are in close contact to the body. If no ECG is measured, the electrodes may have lost contact to the skin and the measured temperature is not reliable. Thus, with the knowledge that the thermal behavior of the electrode can roughly be approximated with a first order linear time invariant (LTI) system, it can be estimated after which time a temperature measurement is reliable if an ECG is detected. The advantage of integrating several sensors into one sensor device is also shown in [5].

2.2. Artifact detection

A severe problem in capacitive ECG measurements is the high sensitivity to motion artifacts. Since these artifacts are mainly caused by triboelectric effects on the electrode body interface, they are independent of the common mode ratio rejection as they are a differential signal [6, 7]. In this paper a novel artifact detection method for capacitive ECG measurements based on an optical measurement of the electrode body interface is presented.

Several methods such as acceleration sensors [8], capacitance [9], impedance [8, 10, 11] or pressure measurements [12] exist to detect motion and the resulting artifacts to discard these intervals. However, all these methods have several disadvantages: an acceleration sensor integrated into the electrode does not measure movements of the body, the capacitance, impedance and the pressure measurement are more suited to measure the interface but do not detect lateral movements of the body relative to the electrode. Additionally, in case of a very good coupling, changes in coupling impedance may be very low, whereas possible artifacts may still be arbitrary large. Thus, these parameters may be not perfect as a predictor for artifacts.

A new concept for motion detection (and thus artifact detection), which does not exhibit the drawbacks of the previous mentioned methods, is shown in Fig. 3. Light from the LEDs penetrates the textiles and skin and is received by a photodiode. The penetrated areas usually look like banana shapes and has been intensively studied in case of reflective pulseplethysmography (PPG), see e.g. [13]. Here, the artifact detection is basically based on the detection of changes in the optical path, similar to the sensitivity of PPG Sensors to motion artifacts: any motion or pressure at any interface or material will change the optical path and hence the received signal x_{opt} at the photodiode. Since artifacts due to triboelectric effects are very difficult to predict (depends on humidity, resitances, etc. [6]) the optical signal is not used to compensate the artifact but to discard intervals in which motion occurs since it is very likely, that motion results in an artifact. This is a kind of worst case consideration.

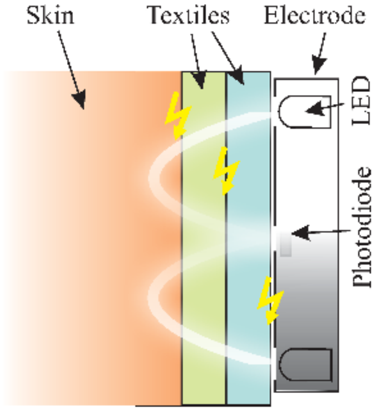

Figure 3. Principle of optical artifact detection.

However, in intervals with no artifacts the optical signal also shows variations. Further analyses showed that these variations correspond to the pulse signal. Surprisingly, this is even detectable through a shirt and a pullover without any special placement of the electrodes at the back. These results are in concordance with [14, 15], but complicate the detection of motion since variations due to the pulse should not be classified as an artifact. An algorithm which is capable to distinguish between motion and artifacts is presented in the following.

The optical signal is defined as

$$\vec{x}_{opt,i} = [x_{opt,i,1}, \dots, x_{opt,i,N}] \quad (1)$$

in which $i = 1, 2$ corresponds to the number of the optical signal and N denotes the total number of samples.

Since a PPG signal usually has small variations (or slopes) and a periodic behavior, but artifacts result in fast and random signal changes, the artifact detection is based on the time derivative of the optical signal and the standard deviation over a specific time period. Hence, similar to a moving window function a time interval of L samples is extracted

$$\vec{w}_{i,n} = [x_{opt,i,n-L}, \dots, x_{opt,i,n}] \quad (2)$$

whereas n follows to $n = L, L+s, \dots N$ in which s corresponds to the step width of the moving window. For each extracted signal $\vec{w}_{i,n}$ a confidence measure CM based on the standard deviation (SD) of the squared time derivative is calculated

$$CM_{i,n} = SD\left(|w_{i,n,k} - w_{i,n,k-1}|^2\right), \; k = 1, \dots, L \quad (3)$$

An interval is classified as an artifact if the sum of both confidence measures reaches a specific threshold θ

$$CM_n = \vec{CM}_1 + \vec{CM}_2 > \theta \quad (4)$$

θ is automatically calculated similar to eq. (1) to eq. (4) but from an interval which is probably free of artifacts. Analyses showed that artifacts usually result in several peaks in the cECG and the optical signals which do not have any specific correlation. Hence, artifact free intervals are estimated by finding those intervals in which at least 4 consecutive R-peaks are in each case followed by only one optical pulse in each optical signal.

3. Results

Preliminary measurements showed that the magnetic fields of the tracking system do not interfere with the patient monitoring. Here, in this paper we focus on cECG measurement results and the optical artifact detection.

3.1. Optical artifact detection

Fig. 4 shows the measured cECG with automatically detected R-peaks (marked with an asterisk) and the corresponding optical signals $x_{opt,1}$ and $x_{opt,2}$ of a subject laying with the back on the electrodes. Intervals disrupted by artifacts are highlighted in red, and it is obvious that artifacts occur in the cECG signal if the optical signal exhibits strong variations. The small variations due to the pulse signal in intervals without artifacts are also clearly visible. Although the subject was wearing a T-shirt and a pullover the ECG and the optical pulse signal are reliably measured.

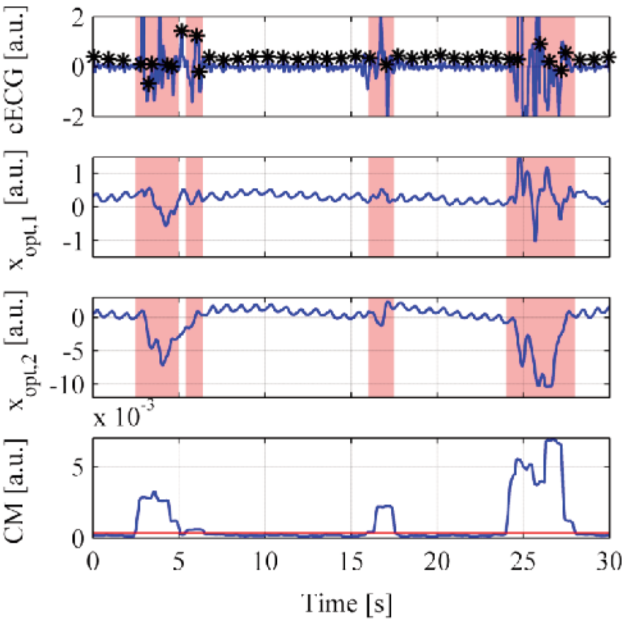

Figure 4. Capacitive measured ECG and the corresponding optical signals and the CM parameter.

The lowest panel of Fig. 4 shows the calculated confidence measure $\vec{CM}$ and the automatically calculated threshold. The moving window has a width of 2 s and a step width of 0.1 s. Since the moving window inherently results in a small time delay, $\vec{CM}$ is shifted by an offset of 0.5 s backwards. The effectiveness of the proposed algorithm is clearly visible as $\vec{CM}$ significantly increases in case of artifacts.

The derived heart rate (HR) from the original signal and the signal after artifact detection are given in Fig. 5. Falsely detected peaks, i.e. peaks which occurred inside

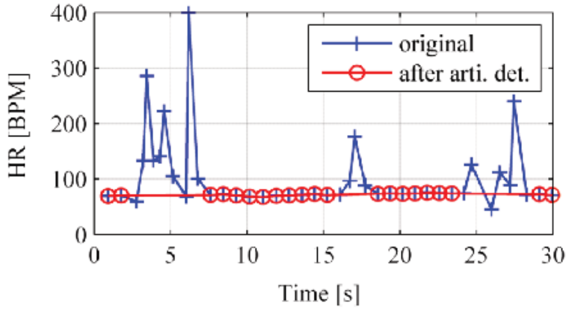

Figure 5. Extracted heart rate before and after artifact detection.

intervals which are marked as artifacts, are discarded and the HR can reliably be estimated.

4. Discussion and conclusions

The combination of a capacitive electric field sensor with an optical reflective measurement and a temperature sensor offers a lot of advantages for performing reliable vital sign measurements, e.g. a plausibility check, or for the calculation of further parameters such as e.g. the pulse transit time. In addition, the integration of these sensors into an operating table offers the opportunity to reduce cables and to provide monitoring without the need for extra devices even in case of small and short interventions. No interference was found between the magnetic tracking system and the sensitive electric field sensor.

The detection of artifacts works well with the proposed algorithm based on the standard deviation of the time derivative of the optical signals. The proposed algorithm even adapts to different physiological waveforms as it can use redundancy in the optical and electrical measurement. However, this methods only detects artifacts which are due to motion at the electrode and not artifacts which are due to common mode distortions. In this case, an additional post-processing is needed.

Acknowledgements

The authors gratefully acknowledge the support provided by the NRW Ziel2 Program funded by the State of North Rhine-Westphalia (Germany) and the European Union, as part of the European Fund for Regional Development.

References

[1] Spinelli E, Haberman M. Insulating electrodes: a review on biopotential front ends for dielectric skin–electrode interfaces. Physiological Measurement 2010;31(10):183–198.

[2] Teunissen LPJ, Klewer J, Haan Ad, Koning JJd, Daanen HAM. Non-invasive continuous core temperature measurement by zero heat flux. Physiological Measurement 2011; 32(5):559–570.

[3] Amatniek E. Measurement of Bioelectric Potentials With Microelectrodes and Neutralized Input Capacity Amplifiers. IRE Transactions on Medical Electronics 1958; 10(0):3–14.

[4] Harland CJ, Clark TD, Prance RJ. Electric potential probes - new directions in the remote sensing of the human body. Measurement Science and Technology 2002; 13(2):163–169.

[5] Chetelat O, Sola i Caros J, Krauss J, Dasen S, Droz S, Gentsch R, Koller JM, Luprano J, O'Hare A, Pilloud P, Theurillat P. Continuous multi-parameter health monitoring system. In World Congress on Medical Physics and Biomedical Engineering, volume 7 of IFMBE Proceedings. Seoul, Korea, 2006; 684–687.

[6] Wartzek T, Lammersen T, Eilebrecht B, Walter M, Leonhardt S. Triboelectricity in Capacitive Biopotential Measurements. IEEE Transactions on Biomedical Engineering 2010;.

[7] Chi YM, Jung TP, Cauwenberghs G. Dry-Contact and Noncontact Biopotential Electrodes: Methodological Review. IEEE Reviews in Biomedical Engineering 2010;3:106–119.

[8] Ottenbacher J, Kirst M, Jatoba L, Huflejt M, Grossmann U, Stork W. Reliable motion artifact detection for ECG monitoring systems with dry electrodes. In 30th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBS). Vancouver, BC, 2008; 1695–1698.

[9] Heuer S, Chiriac S, Kirst M, Gharbi A, Stork W. Signal Quality Assessment for capacitive ECG monitoring systems using body-sensor-impedance. In 4th International Joint Conference on Biomedical Engineering Systems and Technologies (Biostec). Rome, Italy, 2011; .

[10] Kim S, Yazicioglu RF, Torfs T, Dilpreet B, Julien P, van Hoof C. A 2.4μA continuous-time electrode-skin impedance measurement circuit for motion artifact monitoring in ECG acquisition systems. In IEEE Symposium on VLSI Circuits (VLSIC). Honolulu, HI, 2010; 219–220.

[11] Degen T, Loeliger T. An improved Method to continuously monitor the Electrode-Skin Impedance during Bioelectric Measurements. In 29th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBS). Lyon, France, 2007; 6294–6297.

[12] Schumm J, Setz C, Bächlin M, Bächler M, Arnrich B, Tröster G. Unobtrusive Physiological Monitoring in an Airplane Seat. Personal and Ubiquitous Computing Special Issue on Pervasive Technologies for Assistive Environments 2010;14(6).

[13] Rybynok VO, Kyriacou PA. Beer-lambert law along nonlinear mean light pathways for the rational analysis of Photoplethysmography. Journal of Physics Conference Series 2010;238:012061.

[14] Wong MYM, Pickwell-MacPherson E, Zhang YT. Contactless and continuous monitoring of heart rate based on photoplethysmography on a mattress. Physiological Measurement 2010;31(7):1065–1074.

[15] Hyun JB, Gih SC, Ko KK, Jung SK, Kwang SP. Photoplethysmogram Measurement Without Direct Skin-to-Sensor Contact Using an Adaptive Light Source Intensity Control. IEEE Transactions on Information Technology in Biomedicine 2009;13(6):1085–1088.

Address for correspondence:

Tobias Wartzek, RWTH Aachen University
Pauwelsstr. 20, 52074 Aachen, Germany
wartzek@hia.rwth-aachen.de