Wolters Kluwer

Print ISSN 2542-629X
Online ISSN 2226-8561

Medknow

Digital Medicine

Issue 4 | Volume 5 | October-December 2019

Spring Media Publishing

Medicine (basic & clinical) | Mathematics | Informatics | Biomedical & Mechanical Engineering

Understanding the design rules for a nonintrusive, textile, heart rate monitoring system

Theodore Hughes-Riley[1]*, Fraser Hill-Casey[1,2], Carlos Oliveira[1], Arash Shahidi[1], William Hurley[1], Tilak Dias[1]

[1]Advanced Textiles Research Group, School of Art and Design, Nottingham Trent University, Nottingham, [2]Department of Chemistry, University of York, York, UK

ABSTRACT

Background and Objectives: Nonintrusive heart rate (HR) monitoring can be a useful tool for health monitoring. By creating capacitively coupled textile electrodes, a comfortable monitoring system can be integrated into seating or bedding that can monitor HR through clothing. This work empirically studied two factors for a system of this type: the electrode size and the material worn by the subject. **Materials and Methods:** HR measurements were taken using six different sizes of the rectangular textile electrode with four subjects and the signal-to-noise ratio (SNR) of the signals were analyzed. A further set of experiments were conducted with a single subject and a fixed electrode size where different materials were worn. **Results:** Electrode size was seen to have a statistically insignificant effect on the collected signal quality. The SNR was also largely unaffected by the worn material type. **Conclusion:** This study provided empirical data relating to two important factors for nonintrusive, textile, and HR monitoring systems. This data will be helpful for designing a seat-based HR monitoring system or to understand the operational limitations of a system of this type.

Keywords: Capacitively coupled electrode, electrocardiogram, electronic textile, heart rate, textile electrode

INTRODUCTION

This study focused on the characterization of an array of textile electrodes suitable for noninvasive heart rate (HR) measurements from a textile seat cover. HR and heart rate variability (HRV) can provide useful insights into the physiological state of a person, with HR providing an indication of physical exertion[1] or in some cases pathology.[2] HRV has been shown to provide indications of both stress[3,4] and drowsiness.[5] While the project that this study was part of focused on creating a solution for monitoring the stress levels of pilots, this type of system could have wider applications for health monitoring. For example, such a system could be employed in residential care homes to monitor the HR of residents and help provide a prediction of a sudden cardiac arrest[6] or allow staff to intervene in a timely manner if a patient suffers a cardiac arrest, as intervention time is critical to the rate of survival.[7]

Cardiac monitoring is generally conducted using electrocardiography (ECG), which records the electrical

***Address for correspondence:**
Dr. Theodore Hughes-Riley, Advanced Textiles Research Group, School of Art and Design, Nottingham Trent University, Nottingham, UK.
E-mail: theodore.hughesriley@ntu.ac.uk

Access this article online	
Quick Response Code	**Website:** www.digitmedicine.com
	DOI: 10.4103/digm.digm_27_19

How to cite this article: Hughes-Riley T, Hill-Casey F, Oliveira C, Shahidi A, Hurley W, Dias T. Understanding the design rules for a nonintrusive, textile, heart rate monitoring system. Digit Med 2019;5:162-9.

Submitted: 29-Nov-2019 **Revised:** 19-Feb-2020
Accepted: 03-Mar-2020 **Published:** 13-Apr-2020

activity of the heart as a function of time. Electrical signals are measured using electrodes, which are normally mounted directly onto the skin.[8] While ECG is widely employed in medicine[9] and for health monitoring[10,11] the metal electrodes employed can be uncomfortable or inconvenient to wear or use. Textiles provide a comfortable and conformable substrate from which electrodes suitable for HR monitoring can be created.[12-17] The conventional measurement of the heart's electrical signals using any type of electrodes, including textile electrodes, requires direct skin contact,[12,14] necessitating the subject being monitored to wear a specially designed garment with the electrodes embedded as an integral component of the fabric of the garment (for example, the NuMetrex Heart Rate Monitor Sports Bra by Textronics Inc., Wilmington, DE, USA).[18]

An alternative solution is to use high input impedance electrodes that allow for the electrical activity of the heart to be monitored through clothing. Here, the body and skin act as one electrode, the clothing acts as an insulating dielectric layer, and a conductive element acts as a second electrode creating a capacitor.[19] The electrical signals in the body associated with heart activity will change the local magnetic fields within the body, which couples to the displacement current of the capacitor allowing for the signal to be collected.

Capacitively coupled ECG measurement using textile electrodes has previously been demonstrated in the literature: examples include their use for monitoring ECG during sleep,[13] and as part of a seat.[15] For this work, high input impedance knitted electrodes were created as part of a seat back cover, with a further electrode at the base of the seat acting as a ground electrode (driven right leg [DRL]). This meant that the HR and HRV of the aircrew could be monitored by having the aircrew simply sit in the seat.

This study focused on two design considerations for the employment of a textile electrode array for noninvasive HR measurements: electrode size, and worn clothing type, both of which would affect the capacitance of the capacitively coupled electrode. While others have investigated electrode size for dry textile electrodes for HR monitoring, these have been used for direct skin contact measurements,[20] electrode size may have a different effect for the capacitively coupled electrodes. This study focused on empirically investigating the effect of these parameters on the signal-to-noise ratio (SNR) of the signals collected by the textile electrodes. The new knowledge created in this work will inform future designs of textile-based systems for the monitoring of the electrical activity of the heart.

MATERIALS AND METHODS

Technical information

Measuring the electrical activity of the heart

The system used to measure the electrical activity of the heart comprised two main components; the knitted textile electrodes, and the recording and processing hardware. The recording hardware was a modified commercially available system provided by Plessey Semiconductors Ltd. (modifications were performed by Plessey Semiconductors Ltd.; Plymouth, UK) and was based on the capacitively coupled ECG method to measure heart activity: this technique enabled measurements to be taken without direct contact between the electrodes and the skin. Each knitted electrode was attached to an Electric Potential Integrated Circuit (EPIC™ sensor; Plessey Semiconductors Ltd.), which included a preamplifier, using a snap fastener. This placed the preamplifier very close to the surface of the textile electrode (~20 mm away). The array of electrodes, each attached to an individual EPIC™ sensor, and the DRL electrode were subsequently wired to a hardware unit that primarily comprised of bandpass and notch filters, an amplifier, and an analog-to-digital converter. The signals were subsequently read into a computer using a LabVIEW-based piece of proprietary software (LabVIEW; National Instruments, Austin, USA). The recording and processing hardware/software could be used to select the pair of electrodes that provided the best SNR (using an SNR optimization algorithm) from up to eight electrodes; however, only two electrodes (and the DRL) were used at any one time. Data were exported as a comma-separated values (CSV) file and processed using a bespoke MATLAB (MathWorks, Natick, USA) script.

Experimental testing procedure for electrode evaluation

Experimental validation of the system was conducted on four male volunteers, at rest, wearing a cotton T-shirt unless otherwise stated. All participants were members of the research team.

The experimental procedure saw each subject seated in a UK military aircraft pilot's seat (from a BAE Systems Nimrod MRA4; Martin-Baker Aircraft Co. Ltd., Uxbridge, UK), which had panels incorporating an array of textile electrodes attached to the back of the seat using pins. The pilot seat had a five-point harness that was worn by the volunteer while measurements were taken. The seat also had a lumbar support cushion, which was used throughout the study. It was noticed that the action of the volunteer sitting in the seat introduced noise into the recorded signal, this was due to the electrodes mechanically relaxing after being deformed and possibly due to static build-up between the volunteers clothing and the textile electrodes: this noise would typically dissipate after about 1 min and did not reoccur. Therefore, during this study, data were stored after the noise in the signal introduced by the volunteer sitting had subsided. After this, a single 2-min-long sample of data was collected. When possible readings from different volunteers were taken concurrently.

A 1 kHz sampling rate was used throughout this work. Figure 1 shows an example set of data.

The peak of the Q-R-S complex (the R-wave) was used to determine HR by measuring the time interval between R-waves and averaging this interval over 60 s. Despite the noise in the signal, the QRS complex (corresponding to the depolarization of the left and right ventricles of the heart) was discernible within the recorded data, allowing for HR to be extracted.

For evaluation purposes, actual HR values were not evaluated, as the focus of the work was to determine the functionality of the electrodes, and the results are presented as the ratio between the R-wave signal amplitude and baseline noise; this is referred to as SNR within the text. The system output HR and SNR values once per second. SNR values were only averaged when the system output a nonzero or nonnegative HR value to prevent spikes in the ambient noise from significantly influencing the results. Generally, an SNR of seven or higher provided consistent HR results.

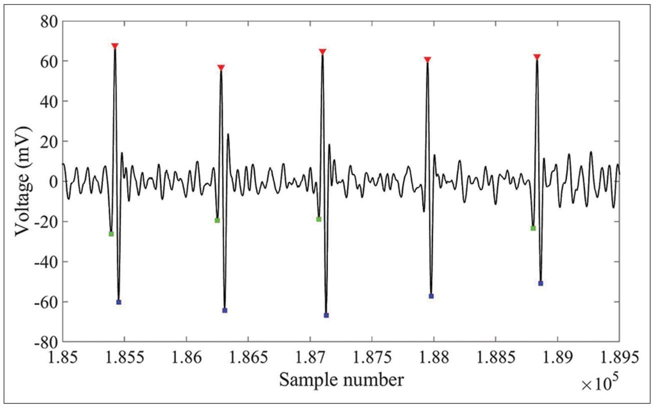

Figure 1: Example heart rate data collected using the knitted textile electrodes. Data have been plotted as a voltage against sample number, which corresponds to time. The QRS complex has been highlighted via an automated peak picking algorithm with 🟩 showing the Q-wave, ▼ showing the R-wave, and ■ showing the S-wave

Knitted rectangular electrodes used in the study

Rectangular knitted electrodes where explored in this work as a rectangle allowed for many different sizes of electrode to be explored without altering the physical placement of the electrodes. For the electrode size optimization experiments, six textile panels with an array of electrodes were knitted to cover the back of the seat: the panels were 0.54 m × 0.48 m, which was informed by the size of the seatback. When placed on the seat, this meant that all of the electrodes used in this study were located over the lower back, with the exception of the largest electrodes [where the used electrodes covered the whole back, Figure 2f]. As all of the participants were a similar height, this meant that the electrodes covered the lower half of their backs (the lumbar region and thoracic region up to around T9).

The conductive elements of the knitted electrodes were fabricated using silver-plated polyamide multifilament yarn (part number 20012223534HCB; Shieldex®, Statex GmbH, Bremen, Germany), with the nonconductive base fabric made from polyester yarns (164 dtex, 48 filaments). To create a stable fabric with minimum extensibility of the electrodes (which would change possibly their electrical properties), a knitted spacer structure with the basic knitted structure interlock was selected for the base fabric: a knitted spacer structure was chosen due to its

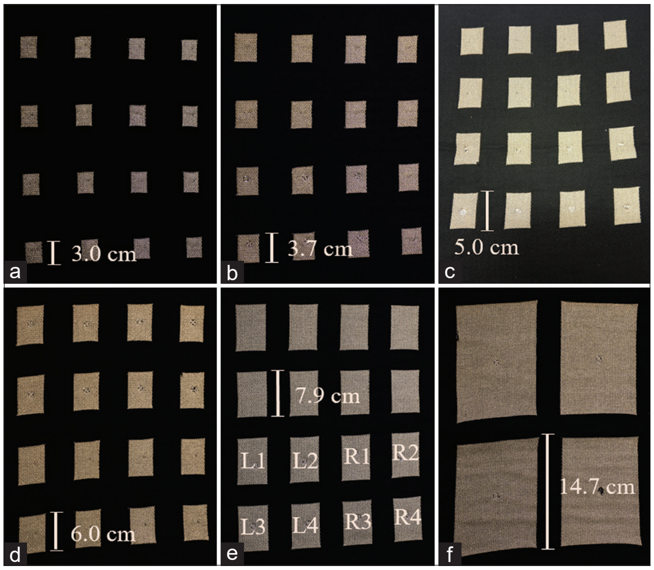

Figure 2: Photographs seatback designs used in this work. Electrode would stretch slightly when pressure was applied. (a) 6.9 cm^2 rectangular electrodes. (b) 11.1 cm^2 rectangular electrodes. (c) 18.0 cm^2 rectangular electrodes. (d) 24.0 cm^2 rectangular electrodes. (e) 44.2 cm^2 rectangular electrodes, with annotation showing the naming scheme for the electrodes. (f) 161.7 cm^2 rectangular electrodes

compressibility to ensure the comfort of the user. All of the samples were produced with a computerized flat-bed knitting machine (Stoll ADF 3 E14; Stoll, Reutlingen, Germany) using an intarsia knitting technique to craft the electrodes using the silver-plated yarn. Each seat back cover included multiple electrodes. The DRL electrode was also a knitted as a spacer structure covering the total area of the seat cushion with one face of the structure being knitted from the silver-plated conductive yarn.

The six electrode designs used in this work are detailed in Table 1. The resistances over the electrodes were measured using a multimeter (model 34410A, Keysight Technologies, Santa Rosa, USA), with the standard deviation provided from the average of four measurements.

Photographs of six of the seatback designs are shown in Figure 2.

Testing the mechanical and electrical properties of the electrodes

Uniaxial tests were conducted on electrode samples using the zwickiLine tensile testing machine (model = Z2.5; Zwich/Roell, Ulm, Germany) following BS EN 14704-1:2005 (determination of the elasticity of fabrics – part 1: strip tests). During these experiments, resistance was recorded over the electrodes (electrode area = 24 cm^2) using a custom Wheatstone bridge fed into a voltage input on the zwickiLine tensile testing machine.

RESULTS

Mechanical and electrical properties of the electrodes

It was important to understand how the electrodes would behave under different mechanical loads, which would exert stress and strain on the electrodes. Not only would users with different weights and body shapes apply different mechanical loads to the electrode but also transient effects, such as breathing would slightly alter the load applied at any given time. This meant that it was important to understand the deformation of the electrode fabric under mechanical loads and observe the change in the electrical properties. The force required to apply different strain percentages to the electrode, and the corresponding electrical resistance are shown in Figure 3.

Unilateral loading and unloading of the knitted electrode showed the force required to apply different levels of strain on the electrodes, with a force of approximately 17 N required to apply a 25% strain to the sample. This showed the inextensibility of the electrodes, which was due to their interlock knit structure. Strain introduced some variation in the electrical resistance over the electrode. This was due to greater or poorer contact between the loops of the silver conductive yarn under different strains. The greatest difference was observed during unloading; this could be attributed to material hysteresis where the previously stretched yarns would need to relax back to their original length to make good contact with neighboring yarns. Under all of the strains explored, the resistance, at most, changed by about a factor of four. For the high input impedance system used in this work, this would not affect the electrode performance; however, this would potentially effect systems where direct skin contact was necessary. Therefore, the application of strain on the electrodes would not affect the functionality of the textile HR monitoring system.

Electrode size investigation

The SNR for four subjects where each of the different sizes of electrodes were used is shown in Figure 4.

The average of the normalized SNR for the subjects is shown in Figure 5.

Average SNR values ranged between 4.0 ± 1.6 and 15.2 ± 1.6. For most of the subjects, the SNR was marginally higher when smaller electrodes were used; this was statistically significant for Subjects 1 and 2. For Subject 4, a significantly lower SNR was observed when smaller electrodes were used.

Table 1: Textile electrode designs explored to determine the optimal size of the electrodes

Name	Electrode area (cm^2)	Electrode length (cm)	Electrode width (cm)	Number of electrodes	Resistance across electrodes (Ω)
6.9 cm^2 rectangular electrode [Figure 3a]	6.9	3.0	2.3	16 (8 used)	0.80±0.05
11.1 cm^2 rectangular electrode [Figure 3b]	11.1	3.7	3.0	16 (8 used)	0.73±0.04
18.0 cm^2 rectangular electrode [Figure 3c]	18.0	5.0	3.6	16 (8 used)	0.88±0.06
24.0 cm^2 rectangular electrode [Figure 3d]	24.0	6.0	4.0	16 (8 used)	1.02±0.18
44.2 cm^2 rectangular electrode [Figure 3e]	44.2	7.9	5.6	16 (8 used)	0.93±0.02
161.7 cm^2 rectangular electrode [Figure 3f]	161.7	14.7	11.0	4 (4 used)	1.06±0.03

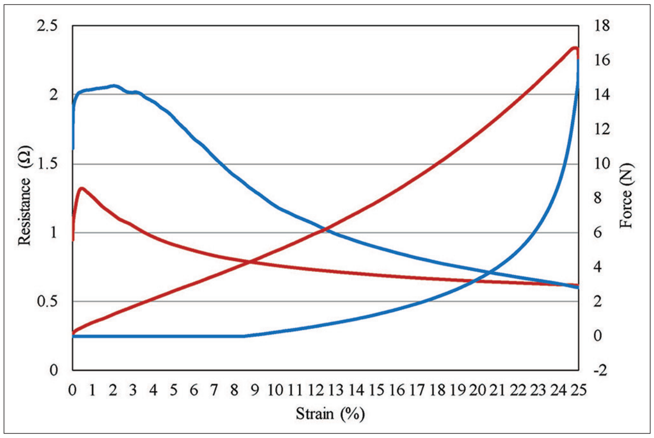

Figure 3: Unilateral tensile tests on a single knitted electrode. Results are the average of five loading and unloading cycles. The force (lines with values increasing left-to-right) and resistance (lines with values decreasing left-to-right) are both shown. Data for when the sample was loaded are shown in red and for when the sample was unloaded is shown in blue

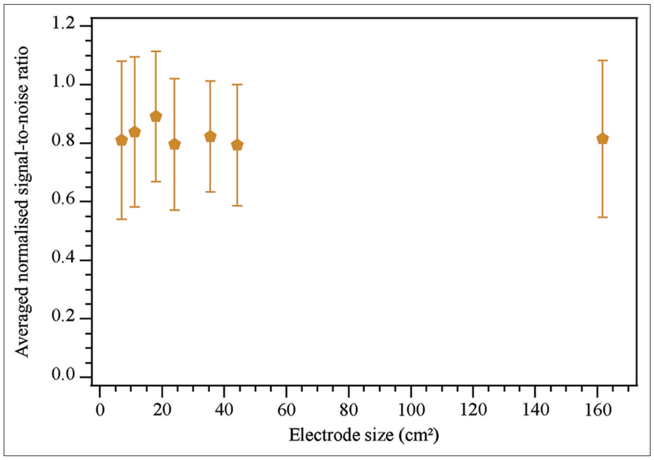

Figure 5: Signal-to-noise ratio for the four subjects compared to the electrode size. The datapoints represent a normalized average of the data shown in Figure 4 (normalized to the highest signal-to-noise ratio obtained for each subject); the error bars were produced using error propagation of the (normalized) standard deviations shown in Figure 4

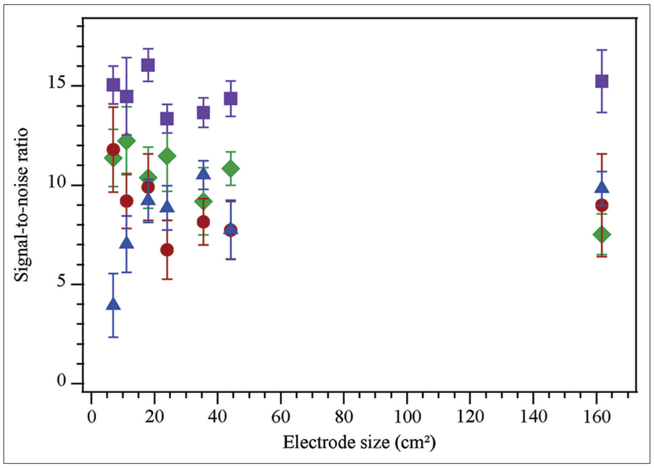

Figure 4: Average signal-to-noise ratio measurements for four subjects as a function of electrode size. Measurements using a rounded rectangular electrode (area = 35.5 cm^2) have also been included. The averages presented are of the SNR for each detected QRS complex, with the associated standard deviation shown by the error bars. Each datapoint was obtained from 2 min of recorded data. Subject 1: ●, Subject 2: ■, Subject 3: ◆, Subject 4: ▲

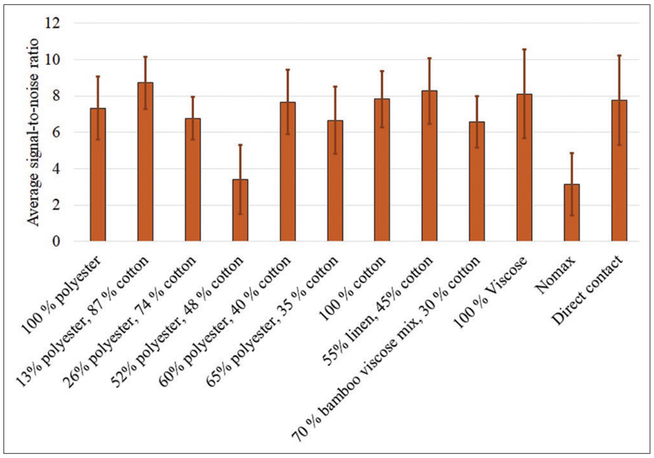

Figure 6: Averaged signal-to-noise ratio compared to worn material type. The averages presented are of the SNR for each detected QRS complex, with the associated standard deviation shown by the error bars. Each datapoint was obtained from 2 min of recorded data

Study of the effects of clothing

Given that the electrodes were not in direct contact with the skin, it was believed that the clothing interface between the human body and the electrodes would affect the quality of the collected signal, as different materials would have different dielectric properties, which would influence the capacitive coupling. To better understand this behavior experiments were conducted with a volunteer wearing clothes made from different materials [Figure 6]. For these experiments, the 4 cm × 6 cm (24 cm^2) electrodes were used.

Figure 6 shows that there was some variation in the SNR for different material types, however, the variation in the recorded SNR made it difficult to draw any clear conclusion. Of the materials tested, the use of Nomax (the material that many flight suits are made of) and a 52% polyester, 48% cotton mix, were the only materials to provide notably lower results, with average SNRs of 3.1 ± 1.7 and 3.4 ± 1.9, respectively.

The low SNR when using the 52% polyester, 48% cotton mix (SNR = 3.4 ± 1.9) was particularly curious given that the similar 60% polyester, 40% cotton mix fabric gave a much higher SNR (SNR = 7.7 ± 1.8).

To try and better understand the observed behavior, the ability for the system to obtain HR values was examined. The system would output data once per second (therefore 120 values in each 2 min of recorded data), however, SNR values were only averaged when a nonzero or nonnegative SNR value was output, as SNR values of zero or less would represent when the system was unable to identify the QRS complex. The working value [Figure 7] represents the percentage of the data points obtained where the QRS complex could be successfully identified.

While Figure 7 shows poor performance for the 52% polyester, 48% cotton mix (working value = 29%) and Nomax (working value = 42%), this level of successful data collection would be adequate for the systems intended application of monitoring air crew, where data would be collected over many hours. For example, a working value of 29% still represents the successful identification of 35 QRS complexes over 2 min.

The highest working value was obtained for the 13% polyester and 87% cotton mix (96%). This was notably higher than when direct skin contact was made with the electrodes (74%).

As different materials have different dielectric properties, it was possible that certain material combinations would improve the capacitive coupling between the electrodes and the body, improving the signal. As such, the electrical permittivity of each of the materials was determined by applying a strip of conductive copper tape onto either side of the material and using an impedance analyzer (Agilent 4192A LF Impedance Analyzer; Agilent Technologies, Santa Clara, USA) to measure the capacitance: electrical permittivity was subsequently calculated. The relationship between the SNR, working value, and relative electrical permittivity is shown in Figure 8.

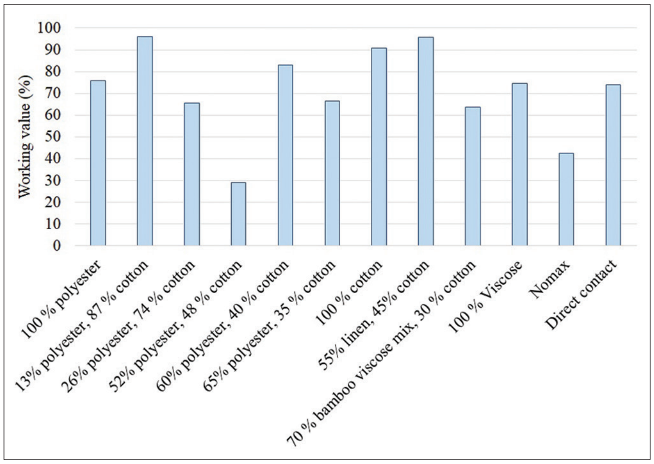

Figure 7: Working value compared to the worn material type. The working value is the percentage of the datapoints obtained from 2 min of recorded data where the QRS complex could be identified correctly

There was no relationship between the material's relative electrical permittivity and the SNR [Figure 8a] or working value [Figure 8b].

DISCUSSION

Electrode size investigation

The electrode size experiments provided a useful empirical result, as in principal larger electrodes would result in a greater collected signal intensity, as the area of the electrode is directly proportional to the coupling capacitance. However, the results showed that, practically, using a larger electrode area did not lead to an increase in the SNR in most cases. This may have been attributed to two factors: an increase in the noise, or body morphology. Capacitively coupled electrodes are prone to picking up signals from other external electrical sources, such as power lines, due to their large impedance (external sources, such as power lines, can capacitively couple to the electrodes and the body creating interference).[21] It is possible that external electrical interference may not have remained consistent throughout the experiments. The noise collected by capacitively coupled electrodes is also known to be very sensitive motion, such vibrations:[22] despite efforts to minimize the movement of the volunteers while sitting, such as the use of the harness, it is possible that the volunteers were not completely still during each experiment.

Different body morphologies would also effect the collected signal, as different body shapes would make better or worse mechanical contact with the electrodes. Poor contact could result in only part of the electrode making mechanical contact with the subject, reducing the effective area of the electrode.

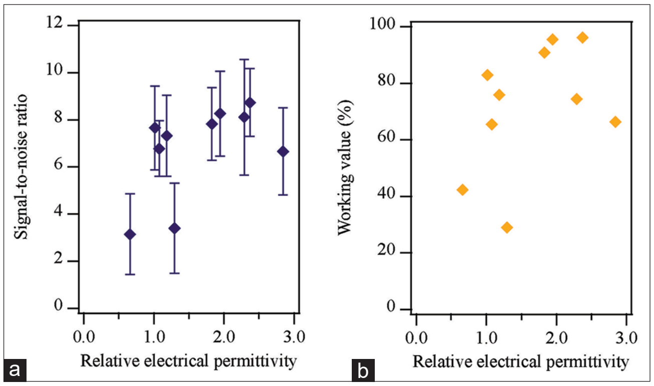

Figure 8: Relative electrical permittivity of different materials compared to: (a) Average signal-to-noise ratio, with the associated standard deviation shown by the error bars. (b) Working value. Please note that the 70% viscose, 30% cotton and the 70% Bamboo-Viscose mix, 30% cotton datapoints have not been included

For this study, the processing and recording hardware/software was used to select the optimal pair of electrodes for each participant (based on SNR). Of the 16 possible combinations of electrodes [for the electrode arrays in Figure 2a-e] only four combinations were selected from twenty experiments. L1-R1 (selected five times), L1-L3 (selected 11 times), L2-R3 (selected three times), and L2-R4 (selected once). Generally speaking, the electrode selection did not change for a given participant for different electrode sizes. This suggested that the functionality of electrodes of different sizes was not effected by their position on the back.

No clear relationship between the electrode area and SNR could be identified. While the results in Figure 4 did not suggest a favorable electrode design, the poor performance seen with small electrodes for Subject 4 [Figure 4] suggested that in some cases, small electrodes were undesirable. Therefore, an intermediate electrode size was most desirable (18.0 cm^2, 24.0 cm^2, or 44.2 cm^2).

Discussion on the effects of clothing

There was no observable relationship between the material's relative electrical permittivity and the SNR (or working value) and direct skin contact resulted in a lower working value than with various worn interface materials: this suggested that the dominant factors in the quality of the obtained signal were the mechanical contact between the subject and the seat, and motion artifacts.

CONCLUSION

The purpose of this study was to understand the design rules for a textile, seat-based, HR monitoring system, where skin contact with the subject was not necessary, by using capacitively coupled electrodes. This empirical study focused on optimizing the textile electrode size and understanding the effect of clothing between the subject and electrodes, both of which are factors that affect the capacitive coupling.

The results of the experiments demonstrated that electrode size had little effect on the quality of the signal obtained for electrodes with areas ranging between 7 and 162 cm^2. The interface clothing between the body of the subject and electrodes was, in most cases, seen to have little effect in the quality of the obtained signal (SNR). It was proposed that the dominant factors in the quality of the obtained signal were mechanical contact between the subject and the seat, and motion artifacts.

Figure 9: A photograph of textile heart rate monitoring system, utilizing an array of 24 cm^2 rectangular electrodes, design for the monitoring of aircrew. Photograph by Jim Boxall

Overall, this study has presented important empirical data, testing two key parameters in the design optimization of a textile HR monitoring system. Ultimately, this knowledge was implemented in designing, and better understanding the boundary conditions for operating, a seat for monitoring the HR and HRV of the aircrew [Figure 9].

Acknowledgment

The authors would like to thank Achala Satharasinghe for his assistance with this work.

Financial support and sponsorship

This project has received funding from the European Union's Horizon 2020 research and innovation program under grant agreement no. 755652.

Conflicts of interest

There are no conflicts of interest.

REFERENCES

1. Noble BJ, Borg GA, Jacobs I, Ceci R, Kaiser P. A category-ratio perceived exertion scale: Relationship to blood and muscle lactates and heart rate. Med Sci Sports Exerc 1983;15:523-8.
2. Kannel WB, Kannel C, Paffenbarger RS Jr, Cupples LA. Heart rate and cardiovascular mortality: The Framingham Study. Am Heart J 1987;113:1489-94.
3. Hjortskov N, Rissén D, Blangsted AK, Fallentin N, Lundberg U, Søgaard K. The effect of mental stress on heart rate variability and blood pressure during computer work. Eur J Appl Physiol 2004;92:84-9.
4. Thayer JF, Ahs F, Fredrikson M, Sollers JJ 3rd, Wager TD. A meta-analysis of heart rate variability and neuroimaging studies: Implications for heart rate variability as a marker of stress and health. Neurosci Biobehav Rev 2012;36:747-56.
5. Vicente J, Laguna P, Bartra A, Bailón R. Drowsiness detection using heart rate variability. Med Biol Eng Comput 2016;54:927-37.
6. Abdelghani SA, Rosenthal TM, Morin DP. Surface Electrocardiogram Predictors of Sudden Cardiac Arrest. Ochsner J 2016;16:280-9.

7. Fan KL, Leung LP. Outcomes of Cardiac Arrest in Residential Care Homes for the Elderly in Hong Kong. Prehosp Emerg Care 2017;21:709-14.
8. Electrocardiography: Overview, ECG Indications and Contraindications, Preparation. Available form: https://emedicine.medscape.com/article/1894014-overview. [Last accessed on 2019 Nov 25].
9. Trzeciak S, Erickson T, Bunney EB, Sloan EP. Variation in patient management based on ECG interpretation by emergency medicine and internal medicine residents. Am J Emerg Med 2002;20:188-95.
10. Shin W, Cha YD, Yoon G. ECG/PPG integer signal processing for a ubiquitous health monitoring system. J Med Syst 2010;34:891-8.
11. Baek HJ, Chung GS, Kim KK, Park KS. A smart health monitoring chair for nonintrusive measurement of biological signals. IEEE Trans Inf Technol Biomed 2012;16:150-8.
12. Ishijima M. Cardiopulmonary monitoring by textile electrodes without subject-awareness of being monitored. Med Biol Eng Comput 1997;35:685-90.
13. Lim YG, Kim KK, Park KS. ECG recording on a bed during sleep without direct skin-contact. IEEE Trans Biomed Eng 2007;54:718-25.
14. Xu PJ, Zhang H, Tao XM. Textile-structured electrodes for electrocardiogram. Text Prog 2008;40:183-213.
15. Lim YG, Chung GS, Park KS. Capacitive driven-right-leg grounding in Indirect-contact ECG measurement. Conf Proc IEEE Eng Med Biol Soc 2010;2010:1250-3.
16. Chamadiya B, Mankodiya K, Wagner M, Hofmann UG. Textile-based, contactless ECG monitoring for non-ICU clinical settings. J Ambient Intell Humaniz Comput 2013;4:791-800.
17. Lee HJ, Hwang SH, Yoon HN, Lee WK, Park KS. Heart rate variability monitoring during sleep based on capacitively coupled textile electrodes on a bed. Sensors (Basel) 2015;15:11295-311.
18. NuMetrex Heart Rate Monitor Sports Bra. Available from: https://crunchwear.com/numetrex-heart-rate-monitor-sports-bra/. [Last accessed on 2019 Nov 25].
19. Higashi D, Imai T, Ueno A, Miyashita O. A wearable capacitive heart-rate monitor for controlling electrically assisted bicycle. In: 2009 International Conference on Electrical Machines and Systems. Tokyo, Japan, Piscataway: IEEE; 2010. p. 1-6.
20. An X, Tangsirinaruenart O, Stylios GK. Investigating the performance of dry textile electrodes for wearable end-uses. J Text Inst 2019;110:151-8.
21. Luna-Lozano PS, Pallas-Areny R. Power-line interference in the ECG obtained with capacitive electrodes. In: 5th European Conference of the International Federation for Medical and Biological Engineering. Budapest, Hungary, Berlin: Springer; 2011. p. 910-3.
22. Asl SN, Oehler M, Schilling M. Noise model of capacitive and textile capacitive noncontact electrodes for bioelectric applications. IEEE Trans Biomed Circuits Syst 2018;12:851-9.