Investigating the Performance of Dry Textile Electrodes for Wearable End-Uses

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ABSTRACT

Textile electrodes have become popular in recent years for their good skin sensorial comfort and their good integration with clothing which offers great potential for sensing of signals for wearable end uses. However, in comparison with wet electrodes, dry textile electrodes have much higher and unstable skin-electrode impedance which could introduce differential noise in signals and cause difficulties in results and diagnosis. To solve this problem, this paper is focused on determining the reasons for this phenomenon and optimizing the performance of textile electrode. Several factors have been examined and the results indicate that the skin-electrode impedance performance is very sensitive to changes of electrode position, size and holding pressure. The fabrication of textile electrode and its optimum holding pressure and size are also described in this paper. Through the implementation of electrocardiogram (ECG) measurements, it was demonstrated that when the electrode size and holding pressure are optimized, the textile electrodes can achieve similar signal performance as wet electrodes.

Keywords: electrode holding pressure, electrode position, electrode size, skin-electrode impedance, textile electrode, ECG

Introduction

Electrodes for measuring biopotential have been studied since 1903, when the first string galvanometer for electrocardiogram (ECG) recording was invented. With advances of ECG recording techniques, many types of electrodes have been designed to meet different requirements. The textile-based electrode is one of these and it was first introduced for ECG monitoring by (Ishijima, 1997), who made part of a bed sheet with electrically conductive yarn to monitor ECG while the subject was sleeping. Many researchers have been working since then on many textile electrode types and much effort has been spent to improve their performance (Donnelly et al., 2013; Pola & Vanhala, 2007; Zheng, Zhang, Wu, & Zhang, 2007)

Compared with traditional silver/silver chloride (Ag/AgCl) disposable electrodes, textile electrodes are more flexible and breathable, which makes the wearer feel more comfortable during long-term ECG monitoring. Also, as they do not need any adhesive and conductive gel, textile electrodes are friendly to human skin and reusable. However, the absence of conductive gel also means that textile electrodes have much higher and more unstable skin-electrode impedance than the traditional Ag/AgCl disposable electrodes, which can introduce unwanted signals noise. This unwanted noise is usually transformed from the common-mode noise that exists in the monitoring system due to the presence of displacement currents in the leads and the human body (Huhta & Webster, 1973). This transformation would only happen when the skin-electrode impedances of the two electrodes are imbalanced.

This paper is focused on the study of the skin-electrode performance in order to minimize the unwanted noise and optimize the performance of the textile electrode. However, many challenges are associated with the measurement of skin-electrode impedance because the human skin has a highly nonhomogeneous multi-layered structure along the different body parts and it also changes its properties with time, which makes any measurement challenging due to very low consistency and repeatability. (Rosell, Colominas, Riu, Pallas-Areny, & Webster, 1988) found that the skin impedance had great variation between subjects using the same electrodes under "identical" conditions, which meant the results measured in different subjects have very low comparability. To exclude time induced impedance variation, some researchers place different electrode pairs on an adjacent skin area such as a subject's thigh or forearm (Rattfält, Lindén, Hult, Berglin, & Ask, 2007; Searle & Kirkup, 2000) and measure the skin-electrode impedances of different electrode pairs at the same time. Although this method could help to exclude time induced impedance variation, it could also introduce the electrode position induced impedance difference as the electrode pairs are placed on different skin position. If we simply neglect the electrode position induced impedance difference, errors might occur when comparing the impedance property of different electrodes. Therefore, the influence of electrode position on the skin-electrode impedance should be studied before measuring it.

Besides the study of electrode position, the electrode holding pressure and the electrode size have also been discussed in this paper due to their influences on the skin-electrode impedance. According to the widely accepted electrical equivalent circuit (figure 1), the skin morphology is highly related to impedance. The application of electrode holding pressure can deform the skin structure, thereby changing the skin-electrode impedance. In addition to this, the electrode holding pressure can also cause deformation of the electrode contact surface because of the surface structure and the mechanical properties of the textile electrode (figure 2). The size of the electrode has also been reported as having a significant influence on skin-electrode impedance and on the ECG signal's quality by some researchers. (Puurtinen, Komulainen, Kauppinen, Malmivuo, & Hyttinen, 2006) studied different sizes of textile electrodes and found that the skin-electrode impedance increases when decreasing the electrode size. (Marozas, Petrenas, Daukantas, & Lukosevicius, 2011) also found that textile electrodes with a contact area smaller than 4 cm^2 might cause distortions to signal's low frequency spectrum. However, the size of the electrodes cannot be infinitely large, not only because of the limitation of skin area but also because its influence on the spectrum of the ECG signals. Thus, a study on the dry textile electrode structural properties, holding pressure and size is essential for understanding their performance during signal monitoring.

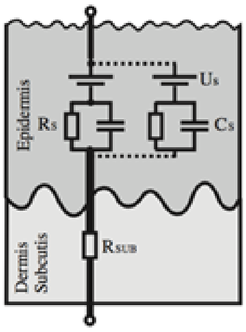


Figure 1. Electrical equivalent circuit of skin (Webster, 1978)

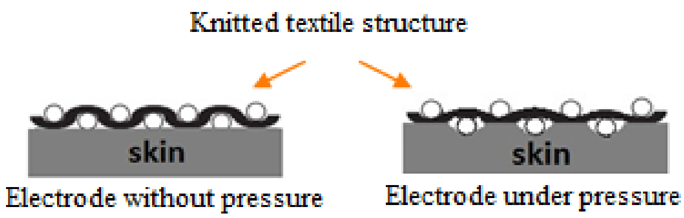


Figure 2. Side view of skin-electrode contact surface

Textile Electrode Fabrication

In smart textile research, various materials have been used to produce conductive textiles which are either embedded into fabrics as conductive yarns or plated with electrically conductive components, such as carbon, copper, nickel, or silver. However, when choosing materials that will come into contact with the human skin, as in the case of ECG electrodes, the biocompatibility becomes very important as the electrode is directly applied onto the human body. Different from most other materials, silver is not only innocuous to human skin but also antibacterial and stable (Tiller, Liao, Lewis, & Klibanov, 2001). Therefore, a 99% silver (Ag) plated knitted conductive fabric which has high electrical conductivity and good skin sensorial wear comfort was fabricated as an ECG electrode for this investigation, table 1.

Table 1. The properties of conductive fabric

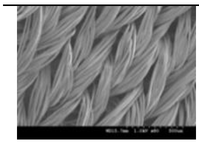
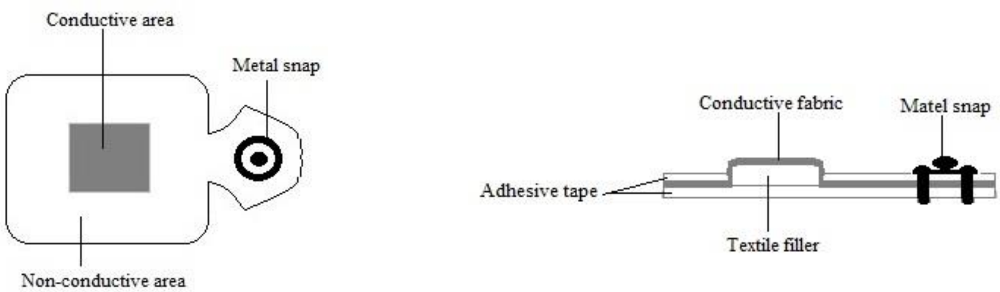
SEM image	Components	Structure	Weight (g/m^2)	Thickness (mm)	Wales per cm	Courses per cm	Surface resistance Ohms/sq
	78% Nylon 22% Elastomer	Plain knitted	130	0.45 ±10%	24	27	Average <100.

Figure 3 illustrates the top and the cross-sectional views of the electrode, in which the square area in the middle shows the conductive fabric surrounded with the outside non-conductive fabric. A metal snap was used to connect the conductive fabric with insulated connecting leads. A textile filler is placed between the conductive fabric and the non-conductive electrode base to provide support to the electrode forming it as a pad. All textile electrodes used in this investigation were prepared with the same materials and structure.



(a) (b)

Figure 3. The structure of experimental textile electrode. (a) Top View. (b) Cross-sectional View

Experimental

To eliminate errors caused by individual differences, our experiments only use one female volunteer as a test subject, and all measurements were carried out on the inner surface of the subject's left forearm without any skin preparation. To establish an "identical" environmental condition for our experiments, all measurements were performed in a conditioned laboratory where the room temperature was controlled at 20±2°C and the relative humidity at 65±2%. In addition, to allow conditioning of the skin to adapt to the measurement environment, the subject was asked to stay in the conditioning lab one hour before starting the experiments. A high-precision LCR-Bridge meter HM8118 (HAMEG instruments, Germany) has been used to measure the skin-electrode impedance, having set the test sinusoidal signal to 100Hz.

The effect of electrode position to the impedance of skin-electrode

In order to study the influence of electrode position on the skin-electrode impedance, six electrode positions on the inner side of the subject's left forearm have been investigated, as seen in figure 4. Each position has one pair of electrodes located on it, and the distance between each electrode pair was 10 mm. All textile electrodes were prepared with the same conductive fabric and have the same skin contact area (15mm x 15mm). In order to place the six electrode pairs in the right position and under the same electrode holding pressure, a whole piece of self-adhesive fabric was used to assemble all the electrodes together in a precisely positioned array. Furthermore, to allow the skin-electrode interface to settle, any measurement was carried out after 30 minutes. This measurement was repeated ten times on ten different days, and only the impedances at 100Hz were recorded. In order to compare the influence of position of the textile electrode, disposable electrodes (type 2228) purchased from 3M were also measured under the same method and conditions. The commercial disposable Ag/AgCl electrodes are referred to as wet electrodes in the following papers due to the presence of an electrolyte gel layer between the electrodes and the skin.

(a) (b)

Figure 4. Electrode positions (a) Electrode positions on subject's forearm. (b) Assembled electrodes

The effect of electrode holding pressure to the impedance of skin-electrode

To ensure the stability of ECG signals and reduce the disruption in the skin-electrode interface, electrodes must be held tight onto the skin. Clamps, suction bulbs, and adhesive foams are normally being used to fix the electrode in position. But the development of textile electrodes offers us the option of using pressure from garments to hold these electrodes in position. Weaving, knitting, and embroidery are the most common textile techniques to integrate textile electrodes into a garment. Ideally, a high electrode holding pressure can ensure a tight and steady contact between electrode and skin, but high pressure is not always following "the bigger the better" principle, as the body sensory comfort also needs to be considered.

Therefore, an investigation was set up for finding the optimum electrode holding pressure by a garment or a chest belt, which was also comfortable for the user. Some research suggests that the garment pressure over 6 kPa (Zhang, Yeung, & Li, 2002) could make the wearer feel uncomfortable and influence the blood flow of the skin. So in this experiment, the electrode holding pressure was chosen from 5mmHg (0.7 kPa) to 45 mmHg (6 kPa), in steps of 5 mmHg. A pair of small textile electrodes of size 15mmx15mm was held on the subject's inner forearm together with a compression sensor by an elastic band. The specified electrode holding pressure is achieved by adjusting the length of the elastic band. The measurement of the pressure is carried out using a compression measurement system from PicoPress. Figure 5 shows the measurement setup of the electrode holding pressure. The skin-electrode impedances were measured at 100 Hz by the LCR Bridge after 30 minutes electrode stabilization period. A pair of wet electrodes (type 2228) purchased from 3M was also been measured in this experiment for relating their impedance variation with the textile electrodes under investigation.

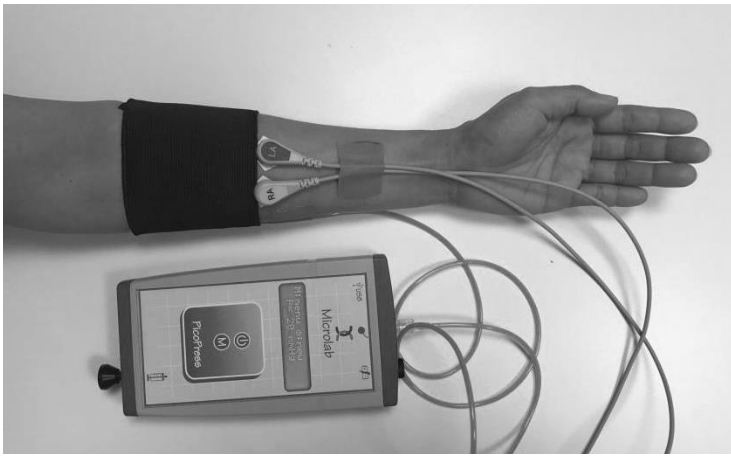


Figure 5. The setup of the electrode holding pressure measurement

The effect of electrode size to the impedance of skin-electrode

It is already known that the skin-electrode impedance increases when decreasing the electrode size, so a better signal can be achieved by using a larger electrode. However, the electrode size cannot be increased infinitely, restricted by available skin area. In order to choose the optimum size for our textile electrode, three different electrode sizes have been made, as seen in table 2. All of them were made using the same silver plated conductive fabric (see table 1), and two identical electrode samples were made for every size, six electrodes in total.

Table 2. Textile electrode in different size

	Large Textile Electrode	Medium Textile	Small Textile Electrode

		Electrode	
Shape of Electrodes			
Electrode Area (cm^2)	8	4.5	2.25
Electrode Dimension (WxL,mm)	20x40	15x30	15x15

To exclude time induced impedance differences, all three pairs of textile electrodes were attached to the skin at the same time and measured immediately. However, we must bear in mind that the electrode position still induces impedance differences, which makes the measurement results not only reflecting the difference of electrode properties but also reflecting the skin position difference. In order to understand and analyse the influence of electrode position on the results of the skin-electrode impedance, the three pairs of electrodes were placed on skin positions that had been measured in the first experiment. Due to the size of these electrodes, only the skin positions 2, 4, and 6 have been used in this experiment. The experiment was performed twice in two different electrode position groups, as seen in figure 6. In electrode position group A, the large, medium and small electrodes were placed on the skin position 2, 4, and 6 separately, while in the electrode position group B, the large and small electrodes exchanged position, while the middle electrodes remained placed in position 4. The three pairs of electrodes were initially assembled in a whole piece of self-adhesive fabric and precisely positioned so that they can be attached on to skin at the same time and under the same electrode holding pressure. The skin-electrode impedances were measured every 20 seconds for 30 minutes in succession.

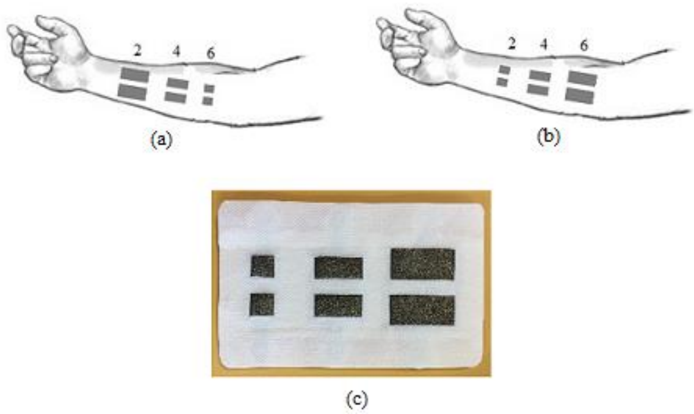
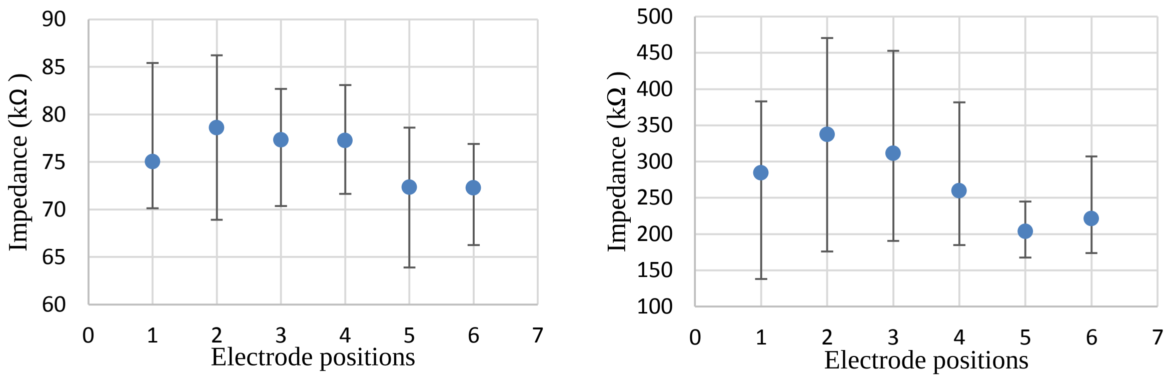


Figure 6. Electrode size with electrode positions order. (a) Electrode position group A. (b) Electrode position group B. (c) Assembled electrodes

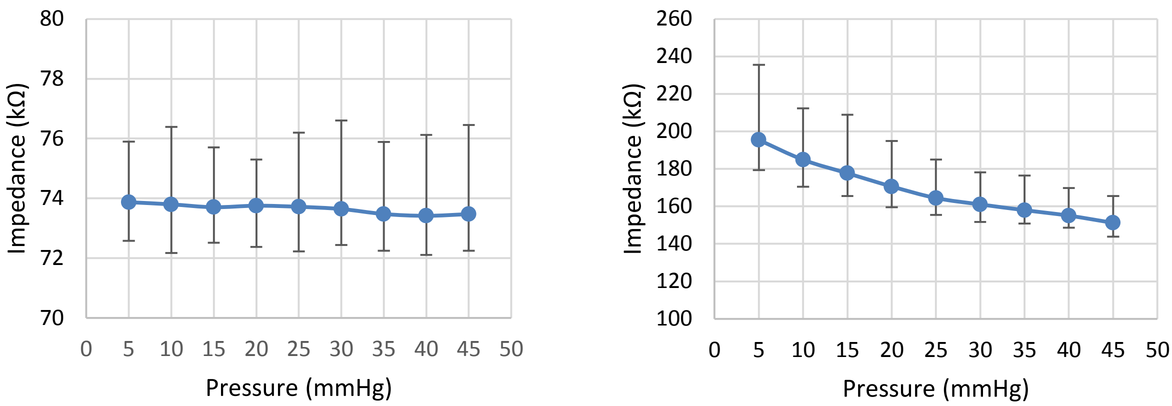
Results



(a) (b)

Figure 7. The impedance of skin-electrode at 6 different electrode positions on the subject's forearm. (a) Wet Electrodes. (b) Dry Textile Electrodes

Figure 7 shows the variation range and the average value of skin-electrode impedance that is measured over 10 different days and at 6 different electrode positions. For wet electrodes, the average impedance has a small difference less than 7 kΩ at these six different electrode positions, the highest average impedance of 78.6kΩ appears in position 2, while positions 5 and 6 have the smallest value of 72.3 kΩ. The wet electrodes also have a relatively small variation range less than 18 kΩ and a standard deviation (SD) less than 7 kΩ at the same electrode position according to the results. In contrast to wet electrodes, huge impedance variations can be observed from the results of the textile electrodes. As shown in figure 7 (b), the electrode position 2 has both the largest impedance variation range nearly 300 kΩ over ten days and the biggest average impedance value of 337.4 kΩ, while the electrode position 5 has the smallest impedance variation range of 77.1 kΩ and the smallest average impedance value of 203.5 kΩ. Furthermore, different electrode positions also cause large differences on the average impedance value of textile electrodes, which could be as great as 133.9 kΩ, as shown in positions 2 and 5. Even at adjacent positions impedance differences occur, which could be as high as 56 kΩ, as seen in positions 4 and 5. The results in figure 7 highlight that the skin-electrode impedance changes with time and is electrode position specific and in comparison with the wet electrodes, the dry textile electrodes are more sensitive to position and time change. For textile electrodes, these impedance variations caused by electrode positions cannot be simply ignored because they have significant influence on the skin-electrode impedance no matter how close to the skin these electrodes are. Therefore, when comparing the skin-electrode impedance of different textile electrodes, the design of the experiment should be very careful, otherwise, the impedance difference caused by positioning of the electrode could be mistakenly attributed to electrode properties.



(a) (b)

Figure 8. The skin-electrode impedance under different electrode holding pressure, (a) Wet Electrodes, (b) Dry Textile Electrodes

Figure 8 shows the relationship between electrode holding pressure and skin-electrode impedance. Two different electrodes, (a) wet electrodes, and (b) dry textile electrodes have been used in this experiment. The performance of wet electrodes and dry textile electrodes under different electrode holding pressure are very different. With an increase in electrode holding pressure, the skin-electrode impedance of textile electrodes has a very clear decline, the average impedance is reduced from 195.4 kΩ at 5 mmHg to 151.2 kΩ at 45 mmHg; the impedance variation caused by those pressure changes is almost 45 kΩ. In the meantime, the variation range of the textile electrodes also decreases from 56 kΩ at 5 mmHg to 21 kΩ at 45 mmHg. In contrast to textile electrodes, the wet electrodes were not sensitive to pressure changes, only a slight fluctuation of around 0.5 kΩ presented with changes of pressure. This might be due to the presence of conductive gel on the skin-electrode. Therefore, the electrode holding pressure is more important in the dry textile electrode than in the wet electrode. Increasing the textile electrode holding pressure can reduce skin-electrode impedance, which can also help to reduce the impedance imbalance of the two electrodes. In order to balance the requirements of body sensory comfort and low skin-electrode impedance, 30 mmHg (4 kPa) has been chosen as an optimum electrode holding pressure for further investigations because over 30 mmHg the subject starts feeling uncomfortable, while the decrease of impedance is not significant.

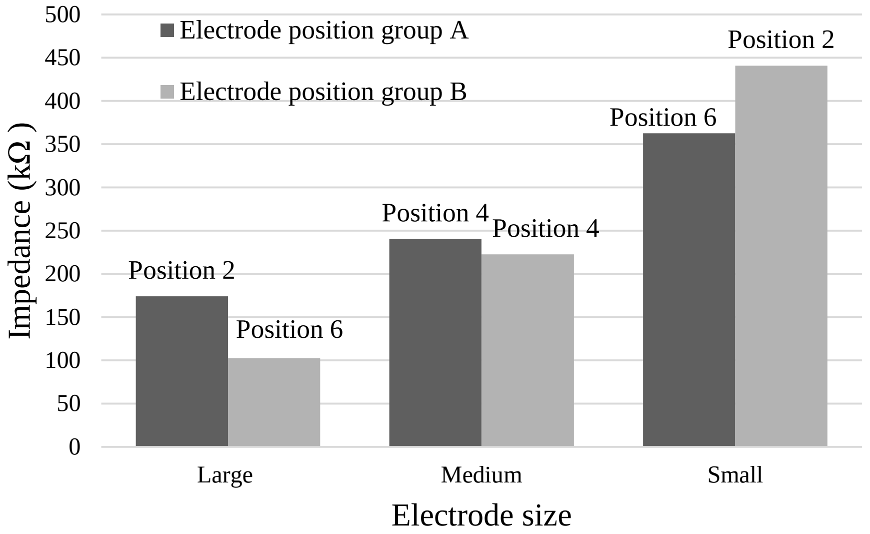


Figure 9. The skin-electrode impedance of different sizes of textile electrodes at different electrode positions

Figure 9 shows the data of skin-electrode impedances measured 30 minutes after the electrodes are applied on the skin. The results show consistency, the skin-electrode impedance decreases when increasing the electrode size. It can also be seen that the electrode positions have great influence on the impedance magnitude. For both large and small electrodes, the skin-electrode impedance in position 2 was always higher than in position 6, the impedance difference is nearly 72 kΩ for large electrodes and 78 kΩ for small electrodes. These impedance differences can be reasonably explained by the effect of electrode position, shown in figure 7(b), where the impedance in position 2 is much higher than position 6. As for the medium sized electrodes, the impedance difference in position 4 can be explained by the time induced impedance variations, as the data from group A and group B were measured at different times. But the time induced impedance difference is only 17.8 KΩ, which is not significant compared with the impedance difference caused by the electrode position and size. Therefore, when measuring the skin electrode impedance, it is very necessary to consider the influence of the electrode size and position on the measurement result.

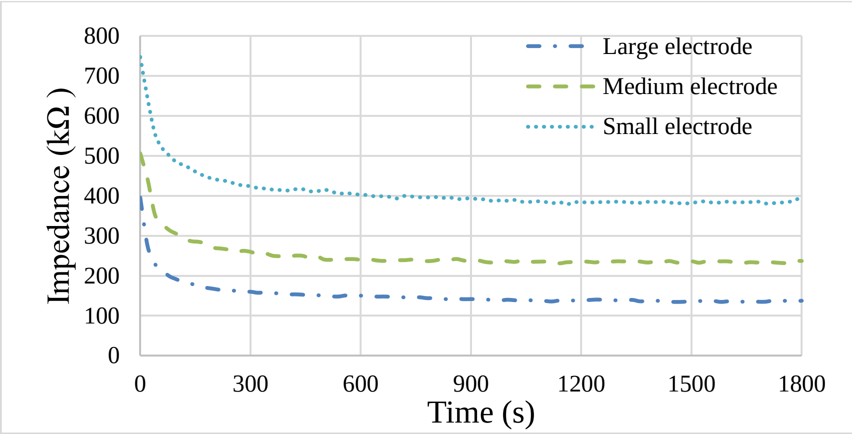


Figure 10. The average results of skin-electrode impedance at 100Hz against time

Figure 10 shows the average impedance results measured at two different electrode position groups. All impedance values are rapidly decreasing within the first three minutes and then gradually become stable. The impedance of the small electrodes reduces from 747 kΩ to 448 kΩ, the medium electrode reduces from 506 kΩ to 280 kΩ and the large electrode reduces from 396 kΩ to 170 kΩ. This is mainly due to the accumulation of perspiration between the electrode and skin, which overtime forms a conductive layer that increases the electrical conductivity of the human skin. In order to minimize impedance imbalance, a low impedance magnitude and short impedance stabilization time is desirable. Therefore, the large textile electrode pair is most suitable for signal recording. A three minutes electrode stabilization period before performing any measurement will allow the electrode to become stable.

Implementation in ECG wearable measurement

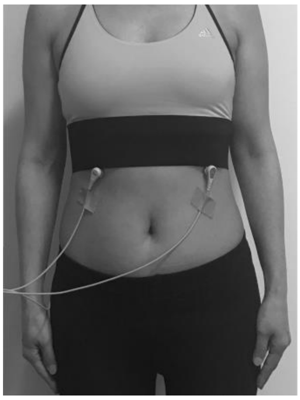
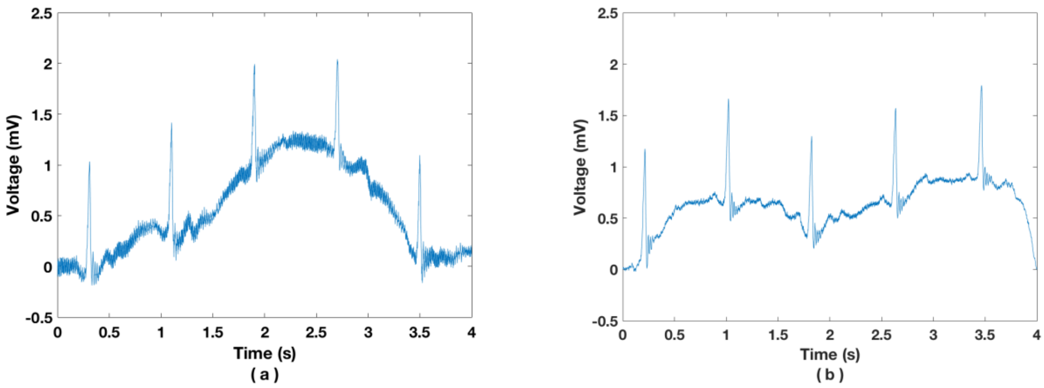


Figure 11. The ECG measurement

In order to test the performance of the textile electrodes, three pairs of textile electrodes of different sizes and one pair of wet electrode were used to measure the real ECG signal. All ECG signals were recorded using the Texas Instruments ADS1292ECG-FE demonstration kit, with only the 50 Hz notch filter operating and all other filters switched off. The sample rate of the signal is 500 Hz and a reference electrode is used to reduce the common-mode noise. All electrodes were secured to the skin with a 30 mmHg pressure applied by an elastic chest band, as shown in figure11. Three minutes skin-electrode interface stabilization period was given before starting the ECG recording.



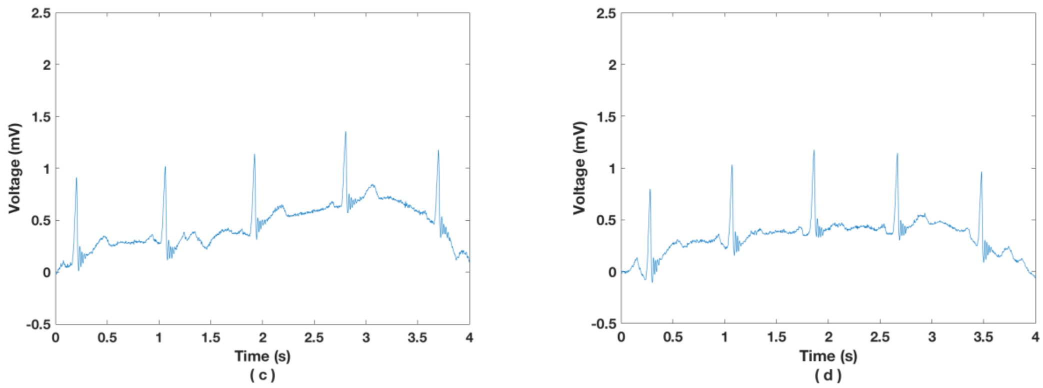


Figure 12. Resting ECG signals recorded by different electrodes. (a) Small textile electrode. (b) Medium textile electrode. (c) Large dry textile electrode. (d) Wet electrode

Figure 12 shows the real ECG signals recorded with three pairs of dry textile electrodes and one pair of wet electrode. As can be seen, baseline drifts exist in all ECG signals, it is mainly due to body respiration and its effect on body volume change causing skin potential imbalance. Besides the baseline drift, noise can also be observed in all graphs because of the presence of electromagnetic field in the vicinity of the patient and during the ECG recording system. As seen in figure 12 (a), the small dry textile electrodes introduced more noise than all others because of its high skin-electrode impedance. However, with increasing the electrode size, the ECG signal quality is getting better. The large dry textile electrode can achieve similar signal performance as the wet electrode, as seen in figure12 (c) and (d).

Discussion and Conclusion

The skin-electrode impedance is an important parameter when evaluating electrode performance, because any imbalance of skin-electrode impedance will transfer the common noise into differential noise, which will cause signal disruption. The ideal signal performance is when all electrodes have the same skin-electrode impedance. However, because of the impedance variation of the human skin, it is very difficult to ensure that the skin-electrode impedances are the same at different body positions. Therefore, excluding the differential noise is not realistic, but by reducing the skin-electrode impedance difference between the two or more electrodes we can minimize any noise. To that end, the most efficient way is to reduce the skin-electrode impedance, and that's why skin abrasion and conductive gels are being widely used in current ECG monitoring. However, textile electrodes for wearable continuous long-term monitoring without any skin preparation and conductive gel application is essential. Therefore, efforts should be directed towards reducing the skin-electrode impedance. Results of this research suggest that increasing the electrode size and the electrode holding pressure can effectively reduce the skin-electrode impedance. Moreover, a three minutes electrode stabilization period is important before performing any signal recording for reducing electrode imbalance.

Another important aspect which has been established in this paper is the unexpected influence of electrode position on skin-electrode impedance. Sometimes, researchers measure different types of electrodes on a subjects' leg or forearm and then compare the skin-electrode impedance difference. Our experiments on a subject's inner forearm suggest that the average impedance variation between two close electrode positions, such as at 1cm distance, could be as big as 56 kΩ. This means that great skin impedance variation still exists even when the electrode positions are very close to each other. If we can't determine whether the impedance difference is caused by the position of the electrode or by the properties of the electrode, errors will occur when analysing and comparing the properties of the electrode. Therefore, we recommend that before determining any electrode characteristics, we need to exclude the impedance difference introduced by the electrode position.

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