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Novel active electrodes for ECG monitoring on woven textiles fabricated by screen and stencil printing

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ABSTRACT

This paper describes a process for fabricating active electrodes and flexible conductive tracks on woven textiles for use in electrocardiogram (ECG) monitoring systems. The process involves the screen and stencil printing of dielectric and conductive polymer pastes on to a textile substrate. Unlike previous printed active electrodes which are printed on *non-woven* textiles, this paper reports active electrodes which are printed on to a significantly more challenging *woven* textile by making use of a polymer interface paste that reduces the surface roughness of the underlying textile. The conductive paths supplying power to the electrodes and carrying the buffered signal to the amplifier are implemented with a screen printable silver polymer paste. The electrode material is a stencil printed carbon loaded rubber. The buffer amplifier, which converts high impedance signals into low impedance signals, is integrated into the electrode structure, reducing the area taken up by the components and improving comfort during wearing. These electrodes are compared to passive electrodes fabricated with the same process and also to commercially available Ag/AgCl electrodes in an ECG monitoring application when the monitored subject is stationary, walking and jogging. It is shown that the textile active electrodes provide significantly improved performance compared to textile passive electrodes and similar performance to the Ag/AgCl electrodes.

1. Introduction

The electrodes used in the monitoring of human biopotentials are usually Ag/AgCl gel electrodes. The weakly polarising Ag/AgCl electrode layer has a stable electrode potential, which prevents DC voltage drift being observed at the amplifier output. The gel, which couples between the Ag/AgCl double layer and the skin, improves ion conductivity and reduces the effective impedance of the skin-electrode interface. Conductive paste can also be applied to prepare the skin, minimising the impedance of the electrode-skin interface and thereby reducing impedance mismatch. Reducing the impedance mismatch between electrodes minimises the effect of common mode signals at the electrodes, such as power line noise, on the amplified signal.

There is currently increasing research on shifting from treatment to prevention in healthcare organisations, resulting in increasing use of wearable monitoring systems, which allow the condition of a patient to be monitored without taking up hospital resources. Ag/AgCl electrodes are typically constructed with an adhesive layer and a hydrogel, both of which degrade and gather hair and dirt with use so that the electrode must be frequently replaced. Repeated use of these electrodes is also known to cause skin irritation [1]. The hydrogel will dry out over time and is not appropriate for longer-term monitoring [2]. Conventional electrodes are therefore an expensive solution for long term monitoring.

Researchers have therefore developed various alternative electrode types for monitoring human biopotentials. Many of these are textile-based, aimed towards providing a wearable solution that can be worn, removed, washed and worn again like a regular garment. Examples include knitted or embroidered electrodes fabricated with conductive yarns, as used in the MagIC project [3]. Researchers have also examined flexible polymer electrodes composed of conductive rubbers [4].

Although these electrodes function adequately in certain situations, where lower signal quality is acceptable, their performance tends to be poor in comparison to Ag/AgCl electrodes. Their weaknesses include a susceptibility to baseline drift, due to the lack of a stable electrode potential, and a high level of 50 Hz noise due to high electrode-skin impedance mismatch since the skin has a high impedance. These can be filtered out in certain situations but, for diagnostic applications, including ECG interpretation, a high level of filtering is unacceptable as it reduces the accuracy of the resulting diagnosis. Researchers have used various methods to improve

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performance, with Tao et al. applying a surface coating of silver chloride to knitted silver electrodes to create a stable potential at the electrode skin interface [5] and Merritt et al. producing active electrodes by screen printing on both sides of a non-woven textile [6].

This paper presents, for the first time, an active electrode structure on a woven textile. Using woven rather than non-woven textiles increases the breathability and wearability of the resulting garment. The textile interface paste [7] used in this process allows these electrodes to be fabricated on a range of different woven textiles with only minor changes to the fabrication process. The novel design of this electrode permits the full structure to be fabricated on a single side of a woven textile making it easier to print. It also allows the buffer amplifier circuit of the active electrode to be integrated within the electrode structure, reducing the obtrusiveness of the systems for the user and increasing comfort.

2. Active electrode design

Active electrodes reduce the signal impedance using a buffer amplifier at the site of the electrode. The active electrode described in this paper is based on the buffer amplifier circuit described by Merritt et al. [6]. However the design in this paper uses printing on only one side of the textile, which simplifies manufacturing. The buffer circuit is contained within the electrode. The design is shown in Fig. 1.

The red sections of Fig. 1(left) show the layout of the conductive layer with a 3-line connection supplying power to and a signal path from the electrode. The conductive layer also provides the circuit connections for the buffer amplifier and a large electrode contact surrounding the circuit in a *U* shape. The black section of Fig. 1(left) shows the layout of the interface layer which lies underneath the red sections and provides a smooth flat base on which the upper layers of the electrode can be printed. After the components are attached, the circuit is then encapsulated with a stencil printed dielectric to prevent unwanted electrical contact between the buffer amplifier circuit and the conductive encapsulation, as shown in Fig. 1(middle). Finally, the electrode is covered with a stencil printed conductive encapsulation composed of carbon loaded rubber, which creates an electrical connection between the skin surface and the electrode surface, as shown in Fig. 1(right).

In this way, the circuit is contained within the electrode structure, which makes the device more compact and the circuit better protected from the environment. The increased thickness of the electrode improves the stability of the electrode contact, which is important for dry electrodes as they have no self-adhesive properties. Instead, they are usually held to the skin by an elastic garment in completed wearable systems [8,9].

For the experiment described in this paper, a bipolar ECG printed electrode layout was designed. This incorporates a pair of differential active electrodes, with the design previously described in this paper, and a passive electrode constructed from the same carbon loaded rubber formulation used in the active electrode. Ideally these electrodes would be printed in a single design but the maximum printing area (150 mm × 150 mm) of the DEK248 screen printing equipment used in this work necessitated two separate printed designs. The conductive tracks from the electrode lead to via connections using 9 mm diameter fabric snap connectors, which can be connected to wires leading to the external amplifier and power supply. The full design is shown in Fig. 2. The spacing between the two differential electrodes is 30 cm.

3. Fabrication

The fabrication process for these electrodes is designed to be homogeneous with printing used for all the fabrication stages except the placement of passive components. A textile called Lagonda is used as a substrate. This textile is composed of yarns containing cotton, polyester and Lycra fibres and was supplied by Klopman International [10]. It has a thickness of 290 μm and offers increased elasticity compared to standard 65/35 polyester cotton whilst maintaining good comfort and printability. Woven, rather than non-woven, textiles are used to improve the breathability and comfort of the resulting wearable monitoring systems. The effect of the anisotropic mechanical properties of the textile on printed structures such as those in this paper have been analysed previously in [13]. First, the textile has three layers screen printed to fabricate a structure with a thickness of approximately 150 μm above the textile surface. The first layer is an interface printed with three deposits of FabInks-IF-UV-1004, making a total of approximately 90 μm thick. This layer is UV cured, requiring a curing energy of 1240 mJ/cm^2 for each deposit. Then, a 5 μm layer

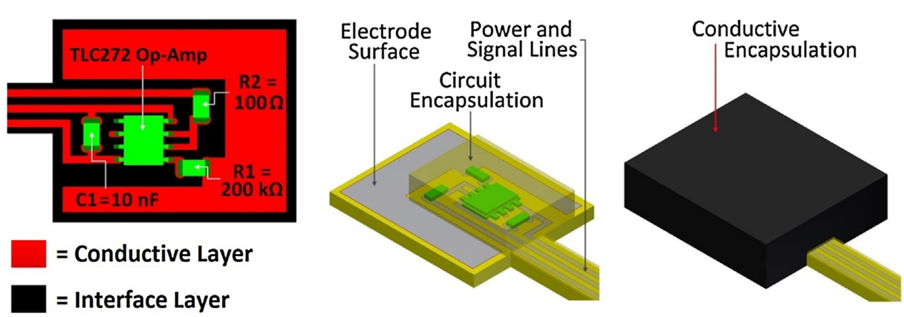

Fig. 1. Active electrode circuit (left) and structure (middle and right).

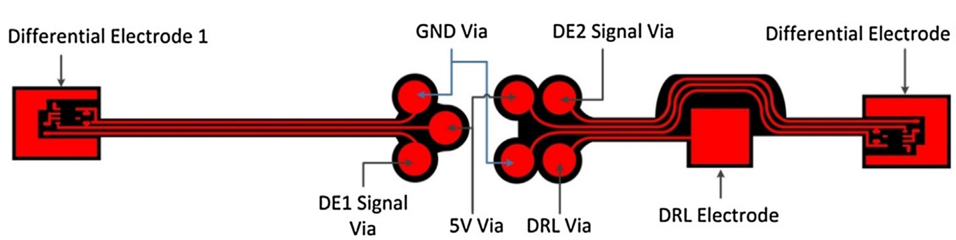

Fig. 2. A three electrode layout for a one-lead bipolar ECG chest band using active electrodes.