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Sensors on Textile Substrates for Home-Based Healthcare Monitoring

Tae-Ho Kang, Carey Merritt, Burcak Karaguzel, John Wilson, *Member IEEE*, Paul. Franzon, *Senior Member IEEE*, Behnam Pourdeyhimi, Edward Grant, *Senior Member IEEE*, Troy Nagle, *Fellow IEEE*

***Abstract*—In this paper we describe progress in developing textile-based sensors for wearable physiological monitoring systems. Active electrodes on nonwoven textile substrates are described for capturing ECG and EOG data. A capacitive sensor for monitoring breathing is presented. Data transmission by coplanar waveguides is also a topic introduced. The future of these devices for home-based healthcare monitoring is considered.**

I. Introduction

As the focus of healthcare migrates from centralized hospital-based treatment to distributed home-based monitoring and health maintenance, reliable and unobtrusive technologies must be found to incorporate sensors and sensing systems into the everyday lives of medical patients. Remotely located healthcare professionals must be able to rapidly detect changes in a patient's health status and thus be able to initiate appropriate remediation, to allow to the maximum possible extent, these patients to continue their normal daily activities. In our project we are studying ways to incorporate sensors and electronic components into clothing and garments that patients can wear in the home environment. These electronic garments must be comfortable, durable, washable, and reliable as they measure physiological parameters of patients undergoing home-based healthcare.

II. Wearable Physiological Data Acquisition Systems

Fig. 1 illustrates our concept for a wearable physiological data acquisition system. Small sensors are incorporated into the structure of the textile garment. These sensors can be bioelectrodes, biosensors, microphones, and/or mechanical sensors such as pressure and strain gages. Specific sets of sensors will be selected for each application. The sensors have analog outputs representing physiological parameters from which the health status of a particular patent will be determined. Wiring and/or wireless interconnections provide power and control signals as appropriate to the sensors and returns information to an embedded data-gathering unit. Here the analog data is converted to digital form and transmitted or transferred to a local information logger, such as a personal data assistant (PDA). From this point, the data can be transferred back to a master database in a physician's office or a remote hospital data network.

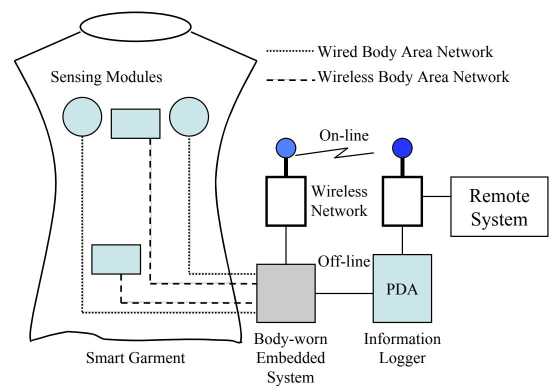

Fig. 1. Concept for a wearable physiological monitoring system.

In the following sections, we give two examples of textile-based sensors and one example of textile-based interconnects for use in the system depicted in Fig. 1.

III. Capacitive Breathing Sensor

A respiration sensor based on a variable parallel-plate capacitive design has been developed using nonwoven fabrics [1]. Fig. 2 shows the structure of the fabric sensor. Stretchable and non-stretchable segments of nonwoven fabrics are laterally attached. The stretchable fabrics are employed to respond to breathing effort resulting in a changing sensor length in the y direction. Conducting areas that form the capacitor's plates are deposited on the non-stretchable fabrics by screen printing silver ink. Their relative positions change when the stretchable portion changes length. Each plate is initially placed so that the conducting areas minimally overlap. As the stretchable

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T.-H. Kang, J. Wilson, and P. Franzon are with the Electrical and Computer Engineering Department, North Carolina State University, Raleigh, NC 27695 USA (e-mail: tkang@unity.ncsu.edu; jmwilson@eos.ncsu.edu; paulf@ncsu.edu).

C. R. Merritt and E. Grant are with the Electrical and Computer Engineering Department, North Carolina State University, and the Biomedical Engineering Department, North Carolina State University and University of North Carolina at Chapel Hill, Raleigh, NC 27695 USA (e-mail: crmerrit@ncsu.edu; egrant@ncsu.edu).

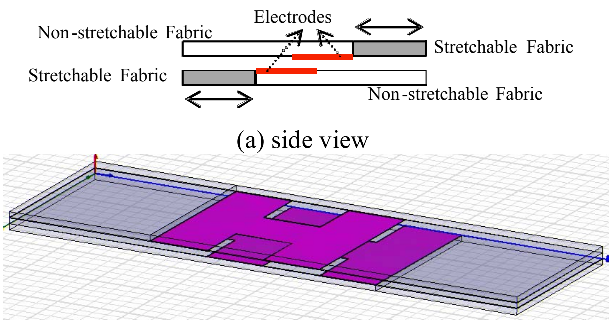
B. Karaguzel and B. Pourdeyhimi are with the Department of Textile & Apparel Technology & Management, North Carolina State University, Raleigh, NC 27695 USA (e-mail: bkaragu@unity.ncsu.edu; bpourdey@tx.ncsu.edu).

T. Nagle is with the Biomedical Engineering Department, North Carolina State University and University of North Carolina at Chapel Hill, Raleigh, NC 27695 USA (919-515-3578; fax: 919-513-3518; e-mail: nagle@unc.edu).

portions of the device are exercised, the two plates slide in opposite directions, changing the effective area and hence the capacitance value. Hence, as a patient breathes, his/her chest expands and contracts. This movement is sensed by the structure of Fig. 2 and is reported to the embedded data-gathering unit of Fig. 1.

IV. Active Electrodes

Electrodes in contact with the patient's skin are important for detecting physiological parameters such as the ECG and



(b) 3D bird's eye view

Fig. 2. Structure of a non-woven textile capacitive breathing sensor.

EOG. For electrodes to be practical for wearable health monitoring units, they should be robust to artifacts introduced by body movement. Body movement deforms the contact shape between electrodes and body skin, and thus changes the interface impedance in the process. Significant changes in contact impedance can introduce large offset voltages and saturate the downstream sensing amplifier. Even though good skin preparation with glued electrodes may reduce the contact impedance and hence the movement effects, this approach is not practical for daily use of electrodes incorporated into a patient's garments. Employing a high pass filter to remove DC offsets in the front-end circuit degrades the noise rejection performance of the input differential amplifier [2, 3]. Recommended solutions are a groundless differential high pass filter [3] and a fully differential amplifier for DC suppression [4]. Even with these improvements, the unshielded signal lines between the high impedance signal source and the measurement circuitry are susceptible to 60 HZ power line interference. However, by employing a buffer amplifier (a voltage follower) in the electrode structure, an "active" electrode is created that reduces the output impedance, dramatically suppressing 60-Hz and other sources of electromagnetic inference.

Our textile-based active electrode design [5] has two non-woven substrates to accommodate, on the first substrate, a transducer layer and a ground layer, and on the second, a signal-path layer with the electrical components, as shown in Fig. 3. Since the skin could interfere with the circuitry, the electrical components were isolated on a different substrate from the transducer. A ground layer and two insulation layers were introduced to reduce high frequency noise. The transducer layer and the ground plane form a capacitor. This

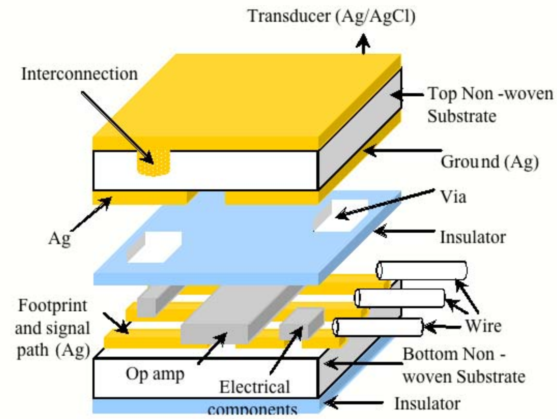


Fig. 3. 3D view of the schematic diagram for the active electrodes

capacitor and the transducer-skin contact impedance act as a low pass filter to remove high frequency noise.

V. Interconnects

To pass signals and power from one portion of a garment to another, we have adopted polymer thick film (PTF) technology [6]. Instead of weaving or knitting conductive yarns with fabrics, we screen print conductive inks onto nonwoven textile substrates. In order to determine the electrical conducting properties of these screen-printed nonwoven substrates, we have adopted time-domain reflectrometry (TDR) on coplanar waveguides (CPWs) as

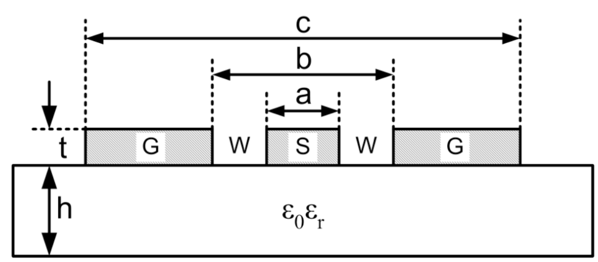


Fig. 4. Cross-section of ground-signal-ground coplanar waveguide (a, the signal width; W, the signal-ground gap; h, the substrate height; and t, the line thickness).

our testing scheme. The advantages of CPWs include: simple fabrication, straight-forward surface mount component assembly, no via holes, and reduced radiation loss [7, 8]. Also, the ground planes of CPWs isolate adjacent signals, which effectively reduce the crosstalk commonly encountered when conductive lines are in close proximity to each other [8]. The design of a coplanar waveguide transmission line consists of a center conductor, which acts as the signal (S), surrounded by two ground planes (G) as shown in Fig. 4.

The CPW circuit layout was screen printed onto Tyvek® and Evolon® as shown in Fig. 5. Results indicated that these structures support data rates acceptable for physiological monitoring.

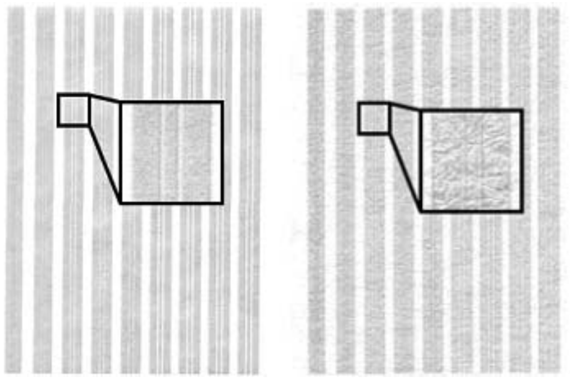


Fig. 5. Screen-printed sample of CPWs on nonwoven substrates Tyvek® (left) and Evolon® (right).

VI. Experimental Results

We fabricated prototype active electrodes (see Fig. 6). The transducer layer was hand-printed with conductive silver ink on the nonwoven fabric. On the back side, a circuit layer, op amp and electrical components were attached to provide a very short path between the electrodes and the op amp inputs.

We collected ECG data using the prototype non-woven passive electrodes and the prototype non-woven basic active electrodes. Fig. 7 displays ECG signals captured using non-woven passive electrodes and non-woven active electrodes, respectively. There is little difference between the two types of electrodes when ECG signals are captured during quite activities such as sitting. However, the active electrodes show better performance than the passive electrodes when the ECG signals are captured during vigorous activities such as jogging, the case presented in Fig. 7. ECG signals captured with the non-woven passive electrodes are not saturated due to the multi-stage scheme to remove DC offset in the front-end circuit. However, we notice repeated

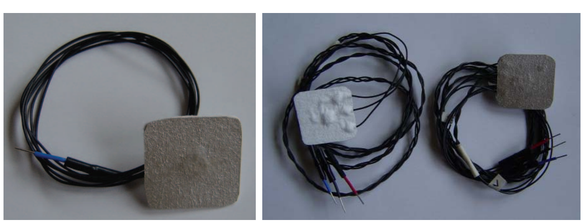


Fig. 6. Prototype fabric electrodes. Passive (left) and active (right) electrodes

negative peaks in the ECG signals captured with the non-woven passive electrodes. The noise is greatly reduced in the ECG signals captured with non-woven active electrodes.

VII. Conclusion

To date, this project has demonstrated the practicability of incorporating sensors for measuring health-related parameters directly into a patient's garments. Textile-based sensors for electrode skin contacts and breathing monitoring are being developed. We expect that these sensors and

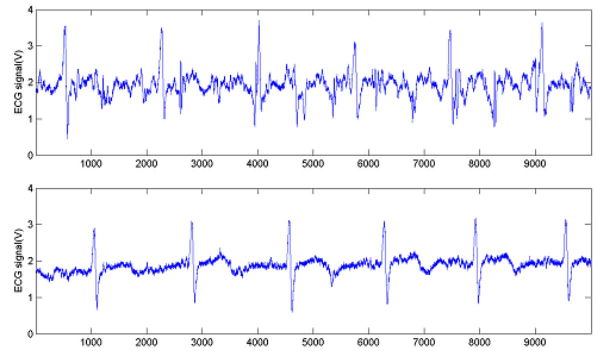


Fig. 7. ECG signals captured with passive (top) and active (bottom) fabric electrodes during jogging.

interconnects will be inexpensive to fabricate and very durable (survive more than 50 garment washing cycles). The sensor-skin interface must be comfortable for the patient. Minimizing skin sensitivity to our sensor designs is an important challenge for the project team over the next year.

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