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Integrating wireless ECG monitoring in textiles

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Abstract

This paper reports on the full realization of a garment embedded patient monitoring system, including wireless communication and inductive powering. The developed system is primarily intended for the continuous monitoring of the electrocardiogram (ECG) of children with an increased risk of Sudden Infant Death Syndrome (SIDS). The sensors and the antenna are made out of textile materials. All electronics are mounted on a flexible circuit to facilitate integration in the baby's pajamas. A significant increase in the comfort of patient and nursing staff is achieved by this integration in textiles. A prototype baby suit was fabricated and successfully tested.

Keywords: ECG; Intelligent textiles; Inductive powering; Telemetry

1. Introduction

Previous work on so-called 'intelligent textiles' shows that biomedical monitoring systems can greatly benefit from the integration of electronics in textile materials [1–8]. This synergy between textiles and electronics mainly originates from the fact that clothing is our most natural interface to the outside world. The higher the level of integration of the sensors and circuits in the clothing, the more unnoticeable the monitoring system becomes to its user and thus the higher the comfort of the user.

This paper presents a garment embedded patient monitoring system, primarily intended for the monitoring of babies prone to Sudden Infant Death Syndrome (SIDS). As these babies have an increased risk of cardiac arrest, a continuous measurement of the electrocardiogram (ECG) is necessary. Up to now adhesive gel electrodes are used, wired to an alarm unit. These wires hinder both the baby and the nursing staff. Furthermore, the long-term use of conventional gel electrodes can cause skin irritation and allergic contact reactions. This paper presents a solution that overcomes these inconveniences by the integration of the sensors, interconnects and the processing and transmission circuitry in the baby's suit. A general system overview is depicted in Fig. 1.

2. ECG measurement

As an alternative to conventional ECG electrodes, both knitted and woven stainless steel electrodes (called Textrodes) were developed in collaboration with Bekintex [8]. These dry Textrodes (Fig. 2) are less irritating than gel electrodes. In addition, stainless steel has a low toxicity, the yarns can be manipulated as a textile material, and can be washed without losing their properties. Disadvantages of the Textrodes however, are the increased sensitivity to motion artifacts, as well as the poor skin–electrode contact. Therefore, the analog ECG front-end had to be redesigned.

Because the Textrodes are of bad quality, a three-electrode ECG configuration circuit is preferred over a two-electrode ECG circuit. In a first attempt, a driven right leg (DRL) topology (see Fig. 3) was implemented, because of its known high common mode rejection ratio (CMRR) [9]. Fig. 4 (left) shows the equivalent circuit, for the calculation of the common mode interference.

In this equivalence, it is supposed that all electrode impedances Z_e are equal and that the amplifier's poles do not lie in the aimed frequency range (0.5–150 Hz). The common

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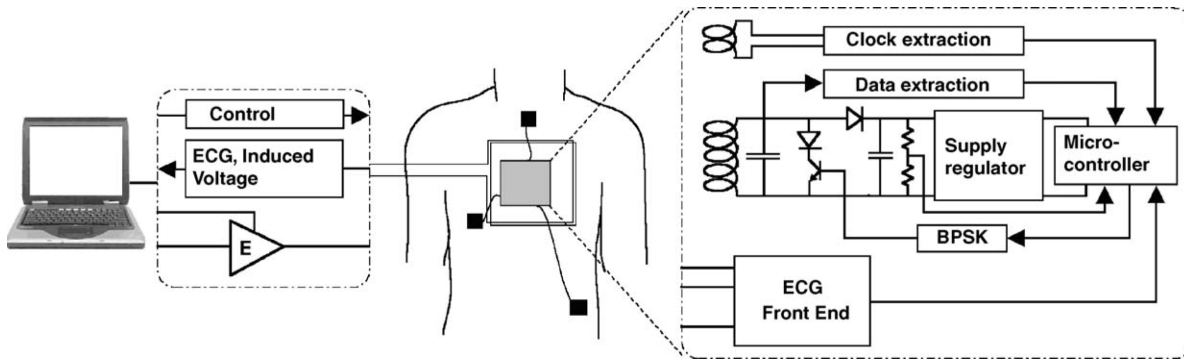

Fig. 1. General overview of the system.

mode voltage V_c is then given by:

$$V_c = \frac{s \cdot Z_g \cdot K \cdot (Z_e + Z_i)}{Z_i(1 + A) + Z_e + Z_g + s \cdot Z_g + Z_g \cdot C_x \cdot (Z_e + Z_i)} \cdot V_p \tag{1}$$

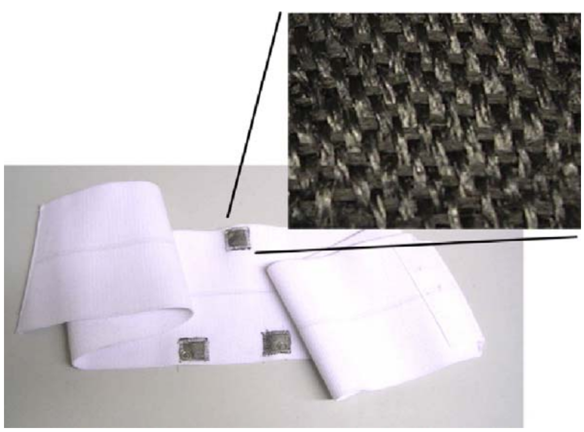

Fig. 2. Belt with Textrodes.

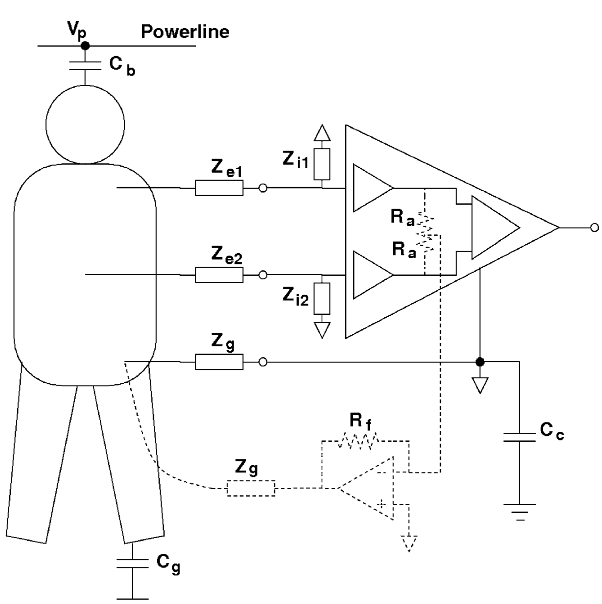

Fig. 3. Simplified model for thee-electrode ECG configuration. Dotted line: DRL; full line: grounded configuration.

with $Z_{e1} = Z_{e2} = 2Z_e$, the input impedances and $Z_{i1} = Z_{i2} = 2Z_i$, the DRL feedback gain A:

$$A = \frac{2 \cdot R_f}{R_a}, \quad C_x = \frac{C_c \cdot (C_g + C_b)}{C_g + C_b + C_c} \quad \text{and}$$

$$K = \frac{C_c \cdot C_b}{C_g + C_b + C_c}.$$

V_p is the common mode source, e.g. due to 50 Hz power-line interference.

Eq. (1) can be further simplified: the electrode impedances are much smaller than the input impedance of the amplifier (Z_e, $Z_g \ll Z_i$) and the pole in (1) lies out the frequency range of interest. Eq. (1) then becomes:

$$V_c = \frac{s \cdot Z_g \cdot K}{1 + A} \cdot V_p \tag{2}$$

We can make the same calculations for a grounded three-electrode configuration. In this case, the equivalent circuit for the calculation of the common mode is shown in Fig. 4 (right). The calculations then lead to:

$$V_c = \frac{s \cdot Z_g \cdot K \cdot (Z_e + Z_i)}{Z_i + Z_e + Z_g + s \cdot Z_g \cdot C_x \cdot (Z_e + Z_i)} \cdot V_p \tag{3}$$

Simplification yields:

$$V_c = s \cdot Z_g \cdot K \cdot V_p \tag{4}$$

From Eqs. (2) and (4), the advantage of the DRL configuration compared to the grounded configuration becomes clear. The common mode is reduced with a factor A, which is the gain of the feedback amplifier in the DRL configuration. To minimize the common mode interference, the gain A can be increased.

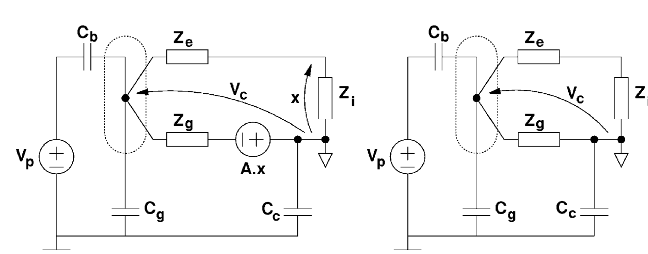

Fig. 4. Equivalent circuit for the common mode. Left: DRL; right: grounded configuration.