Power-efficient wireless sensor for physiological signal acquisition

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Abstract. This work presents a power-efficient wireless sensor implemented using microelectromechanical system (MEMS)-based dry electrodes (MDE) and a ZigBee protocol chip for physiological signal acquisition. To improve signal quality with low electrode–skin interface impedance, a silicon-based MDE is fabricated via micromachining technology. The proposed wireless sensor can provide four different channels for up to 10 kHz bandwidth, 10-bit resolution biomedical signal transmissions. Different from other systems, the proposed wireless sensor employs a novel power management method for physiological signals to reduce power consumption. The proposed wireless sensor successfully transmits electrocardiogram (ECG) signals and four-channel electroencephalogram (EEG) signals with power consumptions of 92.7 and 56.8 mW respectively. It consumes 46% less power than the original sensor without power management (173 mW) in ECG acquisition and 67% less power in EEG acquisition. The circuit printed-circuit-band area in the proposed wireless sensor is 3.5×4.5 cm, suitable for various portable biomedical applications. [DOI: 10.1117/1.3124190]

Subject terms: MEMS sensor; biotelemetry; signal processing; power-efficient.

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1 Introduction

Physiological signals such as electrocardiogram (ECG), electroencephalogram (EEG) and electromyogram (EMG) signals are extensively used and analyzed in biomedical applications.[1–3] In these applications, biopotential electrodes transform the physiological signals from skin tissue to the processing circuit and generate electrode–skin interface impedance, which acts as a voltage divider with the amplifier input resistance to transport biosignals. High interface impedance contributes to the thermal noise and signal attenuation in the system. Accordingly, the most important characteristic of a biopotential electrode is low electrode–skin interface impedance, which is to propagate signals without attenuation or the production of noise.[4] Additionally, protocol-implemented systems are broadly applied and more convenient and comfortable for users and patients in biomedical applications.[5] The lifetime of these systems powered by batteries of limited size typically is too short for long-term applications, because the transceiver modules of these systems consume considerable power.[6] The channel bandwidth of these systems significantly exceeds the requirement of the physiological signal and therefore, power is wasted.

This work presents various microelectromechanical system (MEMS)-based dry electrodes (MDEs) and a novel power-management method associated with the ZigBee protocol for physiological signal acquisition. Silicon-based MDE fabricated via micromachining technology is proposed to improve signal quality with low electrode–skin interface impedance. Low electrode–skin interface impedance is usually required in surface-mounted sensing, because high interface impedance may cause serious signal attenuation while recording. In other words, the signal-to-noise ratio may be reduced. The MDE with microprobe arrays is placed on the skin, passing through the outer skin layer, stratum corneum (SC), into the electrically conducting tissue layer, stratum germinativum (SG), but not into the dermis layer, preventing pain or bleeding. Therefore, it avoids the high-impedance SC layer and provides superior signal transmission to standard wet electrodes that use electric gel.[7] Because MDE is expected to overcome the high-impedance characteristics of the SC, neither skin preparation nor applications of electrolytic gel are required.

The proposed wireless sensor includes MDE and a wireless sensing circuit, whose block diagram is displayed in Fig. 1. The wireless sensing circuit comprises an instrument amplifier, a filter, an analog-to-digital converter, memory, a microprocessor, a voltage regulator for power management, and a ZigBee transceiver. The total power consumption (173 mW) is measured using an Agilent U8001A dc power supply with the voltage set to 3.6 V. The average current in work mode and idle mode is carefully observed by an Agilent DAO5012A oscilloscope. The ZigBee transceiver (122 mW, 71% of power consumed by the whole wireless sensor) consumes most of the power consumed by the wireless sensing circuit. Hence, this work develops a novel power-management system with increased power efficiency, based on the characteristics of various physiological signal.[8] For example, the required bandwidth of the ECG is 40 kHz bps (4 k sample rate, 10-bit resolution), which is

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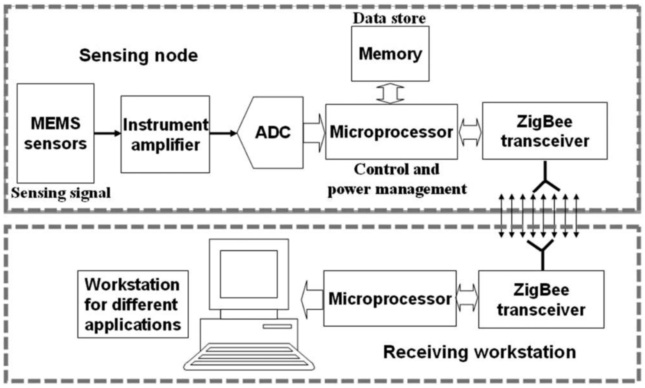

Fig. 1 Block diagram of proposed wireless sensor.

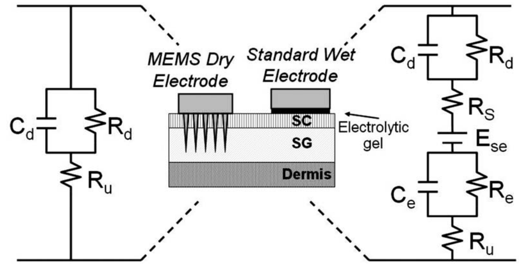

Fig. 2 Electrode–skin interface comparison between spiked dry and electrode standard wet electrode.

much lower than the single-channel bandwidth of the ZigBee standard (250 kbps) , so that the ECG signal could be stored six times in the memory (250 kbytes). Then, the six signals in the memory are sent at one time, saving 5/6 of the transceiver power because the transceiver is turned off the rest of the time. In this work, considering the ZigBee protocol data package characters and according to the ZigBee network specification,[10] the value on ZigBee network data broadcast is 120 kbytes. So, the microcontroller with 128 kbytes of in-system, self-programmable flash program memory is selected for this system. With the high-performance MDE and novel power-management approach in the ZigBee protocol, the proposed wireless sensor can be adopted power-efficiently for physiological signal acquisition.

2 Material and Methods

2.1 *Hardware Overview*

2.1.1 *MDE*

Electrode impedance is the most important issue in silicon MDE designs. To reduce the impedance of standard electrodes, the skin must always be prepared, involving the abrasion of SC and the use of electrolytic gel to improve electrical conductivity. The skin anatomy is a layered architecture, which can be simply divided into three layers from top to bottom: SC, SG, and dermis. The SC acts as a fluid barrier and therefore exhibits electrical isolation characteristics. This layer is constantly renewing itself and consists of dead cells. The SG is the region where the cells divide themselves, grow, and are displaced. Because the SG is composed of living cells, which consist almost completely of liquid, this layer of skin is an electrical conducting tissue that is comparable to an electrolyte. The dermis, which is below the SG, contains vascular and nervous components as well as sweat glands and hair follicles; it is also electrically conducting. Pain originates in the dermis. However, complete removal of SC is very painful and not recommended; most measurements using standard electrodes involve slight or no abrasion, and electrolytic gel is used, diffusing into the SC to improve its conductivity.

In this work, the MDE with microprobe array can penetrate into the SC layer and reach the SG layer to reduce the electrode–skin interface impedance. Figure 2 presents the electrical model of both standard wet electrodes and MDEs. Comparing these two models indicates that the overall impedance of the spiked electrode is less than that of the standard electrode. Additionally, an electrode with a microprobe array can be anticipated to have a lower electrochemical noise.[4]

The dry electrode is fabricated using both isotropic and anisotropic reactive ion etching-inductively coupled plasma processes. The isotropic etching process produces the sharp probe tip, and anisotropic etching produces a cylindrical probe body with a height of approximately 250 μm. In the final stage of the process, wet-etching is utilized to release the hard mask from the probe tip. To ensure electrical conductivity and biomedical compatibility, the probes are coated with Ti/Pt by dc sputtering. Figure 3 depicts the fabrication process.

2.1.2 *Front-end circuit*

The transmitter node consists of six parts: front-end amplifiers, filters, flash programmable memories, a microcontroller, wireless transceivers, and a power supply. An instrumentation amplifier (INA327, Texas Instruments Incorporated, USA) is used as the preamplifier for sensing the biophysiological signals in the biotelemetry system. The characteristics of INA327 are high common-mode rejection, high performance, low cost, rail-to-rail input and output range, single-supply voltage (+2.7–+5.5 V), and small size (MSOP-8). INA 327 can be enabled by applying logic control signal from the microcontroller. A logic low-voltage level turns off the amplifier and reduces its supply current from 2.4 mA to typically 2 μA. These features make it suitable for the system application.

The gain of this instrumentation amplifier can be tuned for various biophysical measurements by trimming the resistor value, and the bandwidth of the filters can be designed to suit the system.

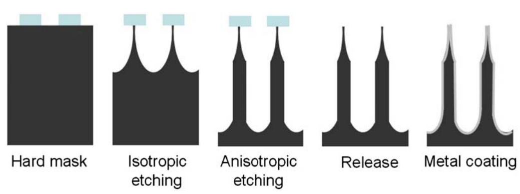

Fig. 3 Fabrication process flow.

2.1.3 *Microcontroller*

A high-performance, low-power AVR 8-bit Microcontroller (128 kbytes of in-system, self-programmable flash program memory, ATMEL) is selected. The microcontroller includes a successive approximation register (SAR) analog-to-digital converter (ADC). It can process multisensor analog signals using the SAR ADC and a multiplexer and store the digital data in flash program memory until the memory is full. MCU links to the radio frequency (rf) transceiver module and provides communicating data. In this system, the microcontroller performs the functions of power-management control, enabling the front-end amplifier circuits and filters, switching the sensing channels, processing the signals, and storing data in the memory.

2.1.4 *Wireless transceiver*

The wireless module is IEEE 802.15.4/ZigBee-compliant. It contains a powerful 8-bit 8051 microprocessor and an rf single-chip (CC2420, from Texas Instruments Incorporated) 2.4 GHz IEEE 802.15.4-compliant rf transceiver. The rf transceiver module is employed as a communication transceiver in bidirectional telemetric devices. A wireless sensor network can be built from the transceiver modules for various applications. The transmission range of the transceiver is 100 m (LOS).

2.1.5 *Power supply*

The power-supply device consists of a +3.3 V voltage regulator integrated circuit (IC), RT9167-33, and bulk decoupling capacitors. The RT9167 IC is a low-dropout-voltage and low-noise regulator and therefore, it is suitable for portable applications. The shutdown mode of nearly zero operation current makes the IC appropriate for use in battery-powered devices. A rechargeable 3.6 V Li-polymer battery pack (700 mA/h) provided the power to the unit.

2.1.6 *Connection to PC*

In the receiving node, the device can be connected to a PC via a universal serial bus (USB) port or a universal asynchronous receiver/transmitter (UART) port. The logic level converter (MAX232, from Maxim) is used to convert the transistor–transistor logic-level voltages to ± 12 V levels to implement the RS-232 serial protocol that governs communication with the PC. The custom-developed software, written in Basic and Lab-View, allows the PC to record, plot, and use the received data.

2.2 *Novel Power-Management Method*

Different physiological signals have different characteristics, which require different things of the implemented system. ECG requires 10-bit resolution and a 4 kHz sample rate, EEG requires 10-bit resolution and a 400 Hz sample rate, and EMG requires 7-bit resolution and a 4 kHz sample rate. In the proposed system, the fixed-specification ADC (10-bit, 10 kHz sample rate) is applied to acquire various physiological signals, which are then processed to meet the requirements described above. The bandwidth is reduced and stored in the memory until it is full. Before the memory is full, the ZigBee transceiver is in standby mode and will be turned on by the microprocessor when the memory is full. Theoretically, the ZigBee transceiver saves 66.67, 96.67, and 75% of the power in ECG, EEG, EMG applications. Figure 4 presents the flow chart of the proposed power-management method.

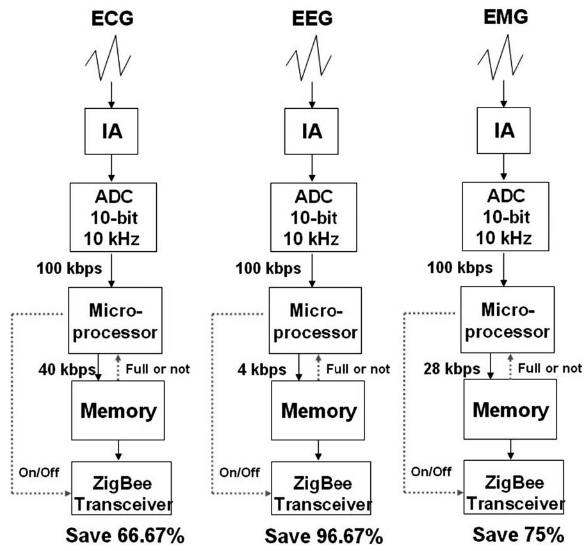

Fig. 4 Flow chart of power management.

2.3 *Firmware Overview*

The firmware that drives the microcontroller is developed in assembly language. Figure 5 shows the flow charts of the main program, the system signal process, and power control. The program, which runs on the microcontroller, comprises a main routine that is divided into three portions of code. The first part initializes the system circuits (communications module, ADC, timer) and sets the parameters of the system signal processing. The second part is the signal-processing function and the program loop that performs the ADC conversions, stores digital data in memory, and sends data through the rf transceiver module. The third part performs the power-management function. The microcontroller is programmed to enable/disable the front-end signal processing circuits, choose different numbers of sensor

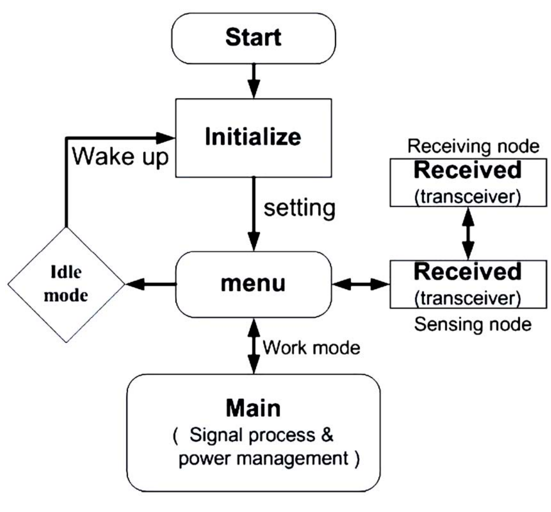

Fig. 5 Flow diagram of the firmware code.

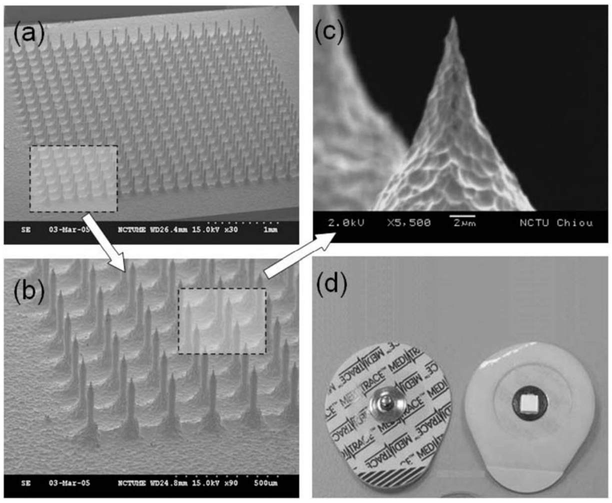

Fig. 6 Fabrication and packaged result of MDE.

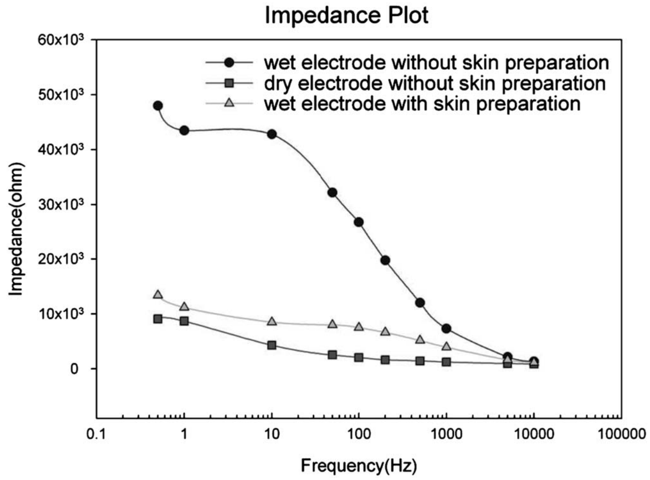

Fig. 7 Measurement results of the proposed MDE.

channels according to user requirements, and switch between the system's work/idle modes to reduce power consumptions.

The main state involves a program loop in which the system switches to the idle mode and periodically wakes up, acquires sensor data, and stores them in memory. As soon as the buffer memory is full, the system sends all of the data simultaneously through the rf transceiver device and starts to wait for new data. In the receiving node, the system can be connected to a PC through an RS-232 port or a USB port. Stored data in the internal electrically erasable programmable read-only memory are sent to the PC to be processed thoroughly using the graphical interface software.

The data broadcast of the ZigBee transceiver is 120 kHz bps, which exceeds most biomedical requirements, and can be set more efficiently with the proposed technique. Figure 4 displays the flow chart of the proposed power-management method. Different physiological signals have different characteristics and different requirements of the system. ECG requires 10-bit resolution and a 4 kHz sample rate, EEG needs 10-bit resolution and a 400 Hz sample rate, and EMG requires 7-bit resolution and a 4 kHz sample. In the proposed system, the fixed-specification ADC (10-bit, 10 kHz sample rate) is used to acquire different physiological signals, and these acquired signals are processed to meet their requirements. The bandwidth is reduced and stored in memory until the memory is full. Before the memory becomes full, the ZigBee transceiver is in idle mode and turned on by the microprocessor. Theoretically, the ZigBee transceiver consumes 66.67, 96.67, and 75% less power in the ECG, EEG, and EMG applications.

3 Experimental Details

Figures 6(a)–6(c) show the scanning electron microscopy (SEM) of the fabricated MDE. Figure 6(d) presents the packaged MDE. The diced dry electrodes were packaged on a flexible printed circuit board (PCB) for convenient usage. After the fabricated MDE was packaged, sterilization was performed with a 25 Kgy gamma-ray from Cobalt 60 (China BioTech Corp., Taiwan). Figure 6 depicts the SEM of the MEMS sensor. The etched dry electrode consists of a 20×20 microprobe array; the etch probe is approximately 250 μm high and 35 μm in diameter. A block bulge is observed around 50 μm at the base of the probe, caused by the isotropic etching shape in the second fabrication process. Accordingly, the effective penetration length of the probe is about 200 μm. The characterization of the electrode–skin interface impedance is of utmost importance in impedance-based biosensing.[10] To verify the performance of electrode–skin interface impedance, electrode impedance spectroscopy is adopted to evaluate the fabricated MDE. The experiment is held by two MDEs at-

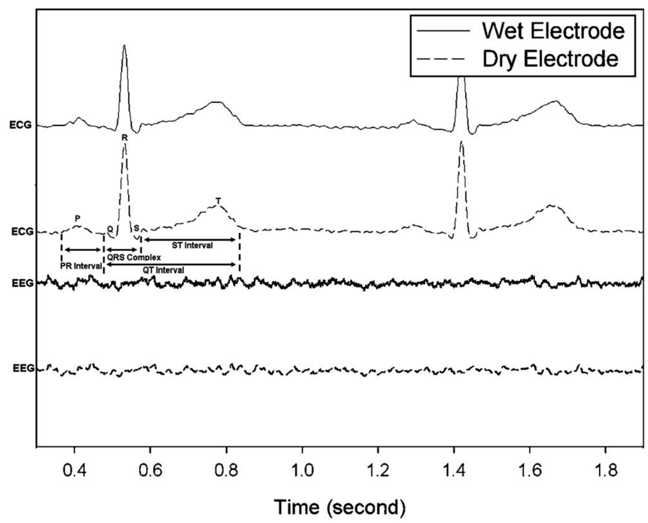

Fig. 8 Comparison of EEG and ECG signals measured by proposed MDE and standard wet electrode.

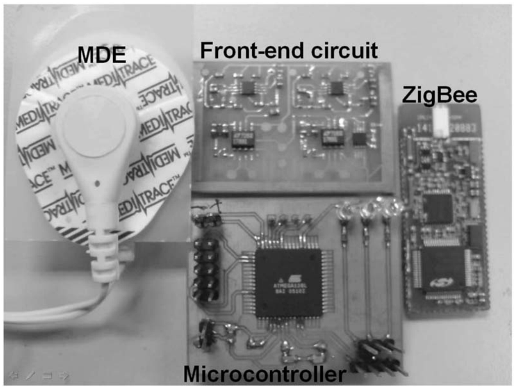

Fig. 9 Picture of the proposed wireless sensor.

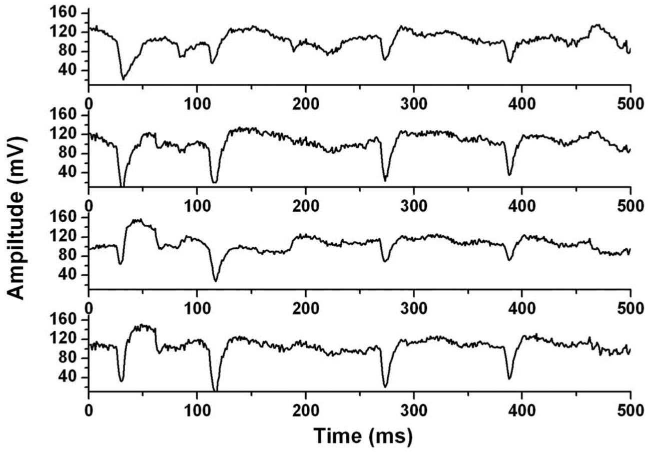

Fig. 10 Four EEG signals sensed and transmitted by proposed wireless sensor.

tached to the skin at a distance of 4 cm from each other, and an electrode–skin–electrode impedance architecture is thus constructed. The test signal is applied to the first MDE, flowed through the skin tissue, and finally passes the second MDE. Therefore, the test results can be regarded as a summary of the impedance of a two-electrode–skin interface.

The measurements plotted in Fig. 7, without skin preparation, demonstrate that the MDE is superior to standard wet electrode in reducing the electrode–skin interface impedance. Nineteen test series of samples are adopted carefully in the impedance experiment, involving five subjects and three electrodes (wet electrode with skin preparation, wet electrode without skin preparation, and dry electrode without skin preparation). Figure 7 plots the measured results. In the frequency range of interest for EEG and ECG (0.5–100 Hz), the impedance of MDE is at least 13 times lower than that of the standard wet electrode.

Moreover, to validate the practical recording performance of the proposed MDE, EEG and ECG simultaneously recorded by the proposed MDE and standard wet electrode are shown in Fig. 8. The solid line and the dotted line represent the measurements of the ECG and EEG signals by the proposed MDE and the standard wet electrode, respectively. The ECG signal is an Einthoven triangle, produced by an ECG amplifier with a right-leg and shield drive circuit.[9] The signals are all sensed by MDE and processed by the front-end circuit. All of the signals are acquired by a 10-bit resolution analog-to-digital converter with a 4 kHz sample rate; four EEG signals used independent channels for wireless transmission. From Fig. 8, the comparison result indicates that the recordings of the MDE are the same as those made using the standard wet electrode. Because wet electrodes are commonly used in modern medical applications, these results demonstrate that the proposed MDE meets modern medical monitoring requirements.

4 Results and Discussion

Figure 9 shows the whole wireless sensor. To verify the effectiveness of the system, ECG and four-channel EEG signals are input to measure the performance of the system. Figure 10 plots the received EEG signals processed with the proposed power-management method. Figure 8 displays the received single ECG signal lead II; the significant features of the waveform are the P, Q, R, S, and T waves, the

Table 1 Power comparison table in different applications.

Signal	Normal	ECG*	EEG*	EMG*
Resolution	10-bit	10-bit	10-bit	7-bit
Sample rate	10 kHz	4 kHz	400 Hz	4 Hz
p(transceiver)	122 mW	40.2 mW	4.1 mW	30.6 mW
P(total)[a]	173 mW	92.7 mW	56.8 mW	82.8 mW
P(save)[b]		80.3 mW (46%)	116.2 mW (46%)	90.2 mW (46%)
Delay time	~0 s	3 s	30 s	4 s

[a]P(total): the power consumptions consist of front-end amplifiers, filters, flash programmable memories, microcontrollers, and wireless transceivers.

[b]P(save): the saving power value in different biomedical applications.

duration of each wave, and particular time intervals, such as the P-R, S-T, and Q-T intervals.[11] The P, Q, R, S, and T functions can be used for ECG monitoring applications. From Figs. 8 and 10, the ECG and EEG are successfully transmitted and the power consumption of the whole sensing node is 92.7 mW (ECG) or 56.8 mW (EEG), 46% (ECG) and 67% (EEG) less than the power consumed (173 mW) by the original system. Table 1 compares the power consumed by the proposed system in various applications.

Figure 9 presents the proposed wireless sensor, including MDE, front-end circuit, microprocessor, and ZigBee module. Three circuit PCBs are stacked, and the connection lines are hidden between the PCBs. The area of the circuit PCB is 3.5×4.5 cm, and the circuit is powered by a 3.6 V Li-ion battery. Although the proposed system sufficiently reduces the power consumed by the protocol-implemented biotelemetric system, it also increases the delay between the sensors and the receiving station. This delay can be adjusted by varying the memory and the data compression ratio. The power is theoretically reduced, but the delay allowed by the application must also be considered. However, in the on-site monitoring and recording, any time delay is not suggested; the portability used for homecare or off-site recording is usually not concerned with the delay. The equipment needing the power-saving methodology should mention the delay and limit the application in homecare or off-site recording.

5 Conclusion

The proposed wireless sensor, including MDE and a novel power-management method, was successively used to acquire physiological signals. Use of this system improves comforts of wearing sensors and increases device lifetime substantially. These results can be the starting point for the development of a novel and miniaturized biomedical sensing device that can be either worn or implanted by the patients and lead to more individualized health care services.

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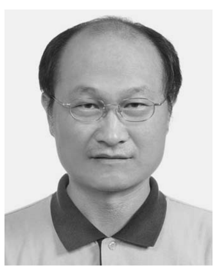

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