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Low-Power System-on-Chip Implementation For Respiratory Rate Detection and Transmission

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Abstract—**Recent biosensors can measure respiratory rate non-invasively, but limits patient mobility or requires regular battery replacement. Respiratory effort, which can scavenge mW, may power the sensor, but requires minimal sensor power usage. This paper demonstrates feasibility of respiratory rate measurement by using a comparator instead of ADC. A low-power system-on-chip can implement respiratory rate detection and wireless data transmission with a total power consumption under 82µW. This approach produces significant power savings, and transmission uses under 30% of total power consumption.**

I. Introduction

Respiratory rate is a vital yet overlooked medical parameter necessary for monitoring a patient's medical condition [1]. Past techniques did not allow for reliable or simple measurement of respiratory rate. Recent advances have introduced non-invasive, comfortable, and accurate methods of rate measurement. However, these recent methods require decreased mobility of the patient; for example, they must stay within range of a large sensing system. Wireless systems require the regular replacement or recharging of batteries to power the sensors and/or radio frequency (RF) link. These limitations are uncomfortable and inconvenient for the user.

Utilization of human energy harvesting can power a wearable wireless unit. Human energy harvesting is commonly associated with kinetic energy, such as walking or other physical activities [2]. For this paper, we focus on the passive harvesting of energy from breathing. Respiratory effort itself will be used to power the system used to measure it.

Such a self-sustaining system requires the power consumption of the components to be less than the power harvested. For a system powered by a low-energy source, such as respiratory effort, all components must be utilized or implemented for high power efficiency. One principle component, system-on-chip (SoC), is used to calculate respiratory rate from the respiratory signal, as well as transmit it wirelessly to a remote receiver display. Power optimization for this component is highly dependent on the algorithm used to process data, and the wireless transmission protocol.

This paper proposes the utilization of the SoC comparator module rather than its ADC module for respiratory rate detection. Successful use of a comparator as a rate detector for a biological signal has been previously done for both heart rate and respiratory rate [3, 4], but not with the focus of low-energy requirements or the use of an on-chip comparator.

II. Zero-Net Energy Biosensor Design

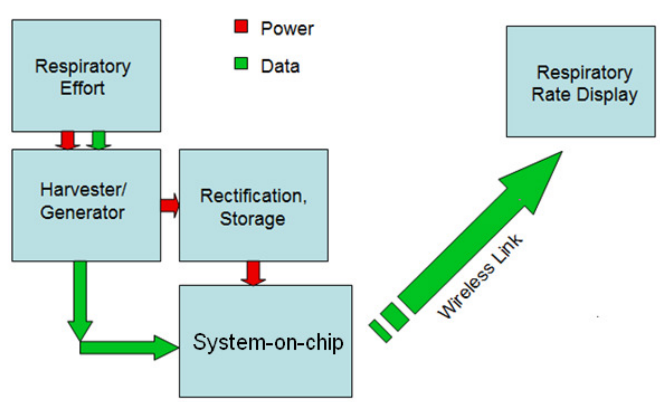

Fig. 1. Zero-Net Energy Biosensor Architecture.

Figure 1 shows the architecture of the proposed self-powered biosensor. Using electromagnetic generation, average power on the order of mW can be harvested, as shown in Figure 2. A wearable harvester was modified from a 9G EXI D213F servo motor to be worn around the chest. The movement of the chest walls during breathing turns the servo armature, thus generating power. This energy can be rectified, and stored in a capacitor or battery, which is used to provide power to the system on chip.

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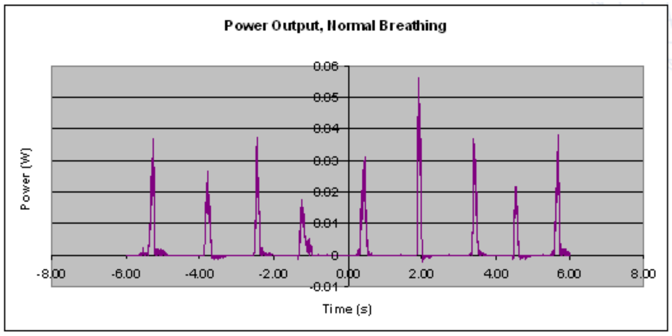

Fig. 2. Harvested Power, Electromagnetic Generation From Resp. Effort.

The same output from the generator is used by the processor to determine respiratory effort. The SoC must sustainably continue respiratory rate detection and wireless data transmission. In this paper, a low power system-on-chip (SoC) is used to process the respiratory signal from a sensor, determine respiratory rate, and transmit the rate to a remote receiver.

III. SoC Description

The SoC chosen for the biosensor design was Texas Instrument's (TI) CC430 chip, which combines the low-power MSP430 microcontroller with its low-power CC1101 915 MHz RF transceiver [5, 6].

In addition to volatile (random-access memory, RAM) and non-volatile (flash) memory, the CPU, and several digital and analog I/O pins, the CC430F1637 on the evaluation board contains a unified clock system (UCS), a power-management module (PMM), a direct memory access (DMA) controller, a port-mapping controller, a cyclic-redundancy check (CRC) module, an AES 128-bit encryption accelerator, a watchdog timer (WDT), two general 16-bit timer modules, a real-time clock (RTC), a 32-bit hardware multiplier, a reference voltage generator, a 12-bit analog-to-digital converter (ADC), a comparator, two universal serial communication interfaces (USCI) that, combined, support UART/IrDA/SPI/I2C protocols, a liquid crystal display (LCD) controller, an embedded emulation module, and a sub-GHz radio.

The CC430 can operate at frequencies of 300-348 MHz, 387-464 MHz, and 779-928 MHz bands. The CC430 supports frequency, amplitude, and phase shift modulation formats. Frequency-shift keying (2-FSK), Gaussian frequency-shift keying (GFSK), and minimum shift keying (MSK) are all supported, as well as on-off keying (OOK) and flexible amplitude-shift keying (ASK). Not all of these modules are pertinent to the proposed implementation; the modules of particular interest are the RAM, flash memory, CPU, analog I/O pins, UCS, 16-bit timers, 12-bit ADC, comparator, and sub-GHz radio.

IV. Low-Power SoC Utilization

This paper proposes the utilization of the SoC comparator module for respiratory rate extraction, and compressed respiratory rate wireless transfer, to minimize power consumption. Since the comparator only does a single comparison with little computational effort, it will utilize less power, making it the more ideal option for analog-to-digital conversion in this particular application. Essentially, comparator operates like a single-bit analog-to-digital converter. To determine respiratory rate, there are only two states of the respiratory signal that must be tracked: (1) a breath has been detected, and (2) a breath has not been detected.

Figure 3 depicts the ADC implementation that most modern technology uses. If ADC runs continuously, its power consumption may dominate power requirements of the SoC. The proposed method minimizes power consumption by using the comparator instead of the ADC, and minimizing data content sent over the wireless link.

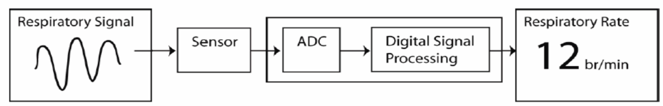

Fig. 3. Respiratory Rate Processing Using ADC.

The power-savings is a relative comparison between utilizing the ADC + TX and utilizing the comparator + TX. The choice of the base clock was made somewhat arbitrarily to be that of the clock available with the kit. It is assumed that lowering the base frequency will result in an equivalent power savings in both the ADC and comparator setups, so the power savings should be viewed in a normalized fashion.

Respiratory rate is obtained using breath counting, a simple technique used to measure respiratory rate during a regular physical exam. The general idea is to count the number of breaths that occur within a certain time interval. If the time interval used is one minute, then the number can be taken straight as the respiratory rate in breaths per minute (br/min). If the time interval is less than a minute, the number can be multiplied to create a minute interval. The comparator SoC implementation of this algorithm is pictured in Figure 4.

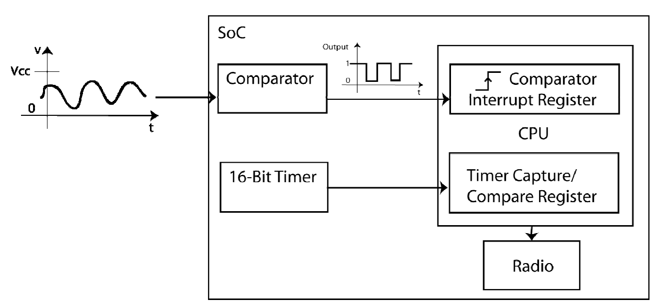

Fig. 4. Breath Counting Algorithm Using SoC.

On the rising edge of the comparator output, an interrupt occurs to indicate that the breath has been detected. In the background, a 16-timer constantly runs and creates an interrupt each time the time interval chosen has elapsed. The interrupt initiates a transmission sequence, which results in the transmission of the respiratory rate.

V. Experimental Setup and Results

The algorithm was coded using the Code Composer Studio Version 4 Integrated Development Environment and programmed onto the EM430F6137RF900 CC430 evaluation board with an MSP-FET430UIF via JTAG connection [7, 8]. The system was tested using simulated respiratory waveforms produced using a signal generator, with a rate of 0.2333Hz, to simulate 14 breaths/min. Validation of the measured respiratory rate was done by transmitting the rate from one evaluation board to another, and then viewing the received rate via an RS232 connection to the Hyperterminal program on a PC. Figure 6 depicts a block diagram of the basic setup, and Figures 7a and 7b are photographs of the transmitter and receiver setups, respectively.

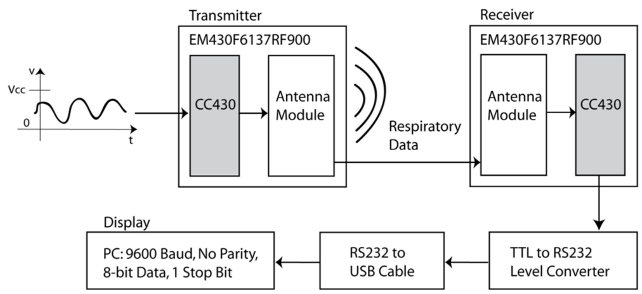

Fig. 6. Experimental Setup Block Diagram.

(a) (b)

Fig. 7. Voltage Measurement Setup on Transmitter (a), and Receiver Setup With PC (b).

Power consumption was measured for the breath counting algorithm using the comparator and compared against the power consumption of an ADC. Measurements were taken over a 2.2kΩ resistor in series with the transmitter evaluation board and its power supply. The power consumption for the ADC implementation was calculated from the SoC specifications.

The average power for transmission was measured and distributed over various transmission intervals to determine total power consumption over one minute. The voltage was measured across a 10Ω shunt resistor.

The calculated powers for the breath counting algorithm, ADC, and data transmission are shown in Table 1. All calculations are based on 3.0V power supply operation, with a generated respiratory input signal having a frequency of 14 breaths/min. Table 2 shows the calculated powers for the same processes at different generated respiratory input signals, with data transmission at 10 second intervals.

TABLE I. Power Consumption Comparison For Select Transmission Intervals, 14 Breaths/min

Algorithm/Conditions	Power Consumption (μW)			
	10-s TX	15-s TX	30-s TX	1-min TX
RF TX only	19.8	13.2	6.6	3.3
Breath counting w/**TX off** (1-bit)	61.4	61.4	61.4	61.4
Breath counting w/**TX on** (1-bit)	81.2	74.6	68.0	64.7
12-bit ADC operating power @ 3.0V w/200ksps	660	660	660	660

TABLE II. Power Consumption Comparison For Different Respiratory Rates, 10-s Transmission Interval

Algorithm/Conditions	Power Consumption (μW)			
	6 br/min	15 br/min	30 br/min	90 br/min
RF TX only	19.8	19.8	19.8	19.8
Breath counting w/**TX off** (1-bit)	61.4	61.4	61.4	61.4
Breath counting w/**TX on** (1-bit)	81.2	81.2	81.2	81.2
12-bit ADC operating power @ 3.0V w/200ksps	660	660	660	660

As shown in Table 1, the power consumed using the comparator and breath counting algorithm is about ten times less than that of the ADC power consumption, regardless of the rate of transmission. Table 2 also shows that using the comparator and breath counting algorithm consumes much less power than the ADC, for a wide range of respiratory frequencies.

Over one minute, the breath counting implementation uses around 67μW, including transmitting once a minute. The ADC implementation, operating at 200 kilo-samples per second, uses around 660μW, which also includes transmitting once a minute. The 200ksps ADC is the native ADC available on the MSP430. This is due to the binary aspect and simplicity of using the comparator just to detect if a breath is taken or not, as opposed to sampling the entire waveform. Using RF transmission at longer intervals saves power, obviously, which is beneficial and may be necessary for a self-sufficient respiratory biosensor.

Zero-net energy biosensors can provide accurate respiratory information, while still being comfortable and convenient for the user. However, since the energy harvested from respiratory effort is on the order of mW, all components of the system must have high power efficiency.

Here, a TI CC430 Ultra-Low Power SoC is used to process the respiratory signal to determine and transmit respiratory rate.

VI. CONCLUSION

In this paper, a comparator method and breath counting algorithm is used instead of traditional ADC implementations. The comparator is used because of lower power consumption as compared to the ADC. Also, the entire respiratory waveform is not needed; only the respiratory rate is transmitted. Thus, the comparator can be used to determine if a breath has been taken, and rate can be determined from the time interval chosen between transmissions.

It is shown that the comparator method uses nearly 10 times less power than the ADC method. Respiratory rate may be sent over the RF link every minute, which is feasible for remote monitoring as well as the power budget. This technique also demonstrated that power required for RF transmission does not necessarily dominate the power budget, due to minimum data content transferred over the radio link.

Future work will include power consumption measurements involving real signals from the wearable respiratory effort harvester, as well as using lower ADC sampling rates to make a better power comparison for the type of signal that is being monitored.

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