27.3 A 3.9mW 25-Electrode Reconfigured Thoracic Impedance/ECG SoC with Body-Channel Transponder

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Recently, wearable heart monitoring systems have been developed for cardiovascular-related disease [1] with wearable body sensor network (WBSN) [2-3]. The WBSN introduced in [3] monitored ECG at maximum 48 points, and transferred data using arrayed inductive link for cm-range wireless inter-connectivity. However, most of the previous attempts were limited to sense only ECG signals at limited points [2] on the body with limited network coverage [3]. Thoracic impedance variance (TIV) from the change of aortic blood volume and velocity at each cardiac cycle provides important hemodynamic information (stroke volume, cardiac output). Combined with ECG signals from more than 6 points, it enables the early detection of abnormal symptoms of pandemic diseases like hypertension and heart failure so that the patients can take prophylactic measures [6]. In spite of its importance, the TIV detection was not realized in WBSN due to its requirement of high impedance (<0.2Ω) detection sensitivity which needs to detect AM signal with modulation depth as low as less than 3%. A pure single tone sinusoidal current signal at 1kHz-100kHz [6] is required to realize such a high sensitivity, and only a bulky implementation was reported so far [7].

In this paper, we report a 3.9mW low power SoC with body-channel-transceiver (BCT), which can detect TIV (0.12Ω) and ECG (up to 8 points) concurrently. The chip is integrated on a 4-layer fabric circuit board with thin flexible battery as a poultice-like plaster. In addition, it can reconfigure the 25-electrode array and optimize them in-situ to automatically consider the user dependency of the TIV/ECG signals. The recorded data is transmitted at 1Mbps through body-channel-communication (BCC) [8] with duty cycle modification to extend battery life time and enlarge the network coverage.

Figure 27.3.1 shows the proposed wearable TIV/ECG monitoring system. The 15cm x 15cm 4-layer patch fabricated by P-FCB [3] consists of Layer-1, 25-electrode array for reconfigurable TIV/ECG sensing, Layer-2, fabric inductor (2.2µH, Q=9.6) for system start-up, Layer-3, thin flexible battery (1.5V, 30mAh), and Layer-4, fabric circuit board on which the SoC is directly wire bonded. Layer-1 has 1) 16 voltage sensing electrodes (each 1.8cm²) divided into 4 sections, 2) 2 anodes and 2 cathodes electrodes (each 3.2cm²) for current injection and 3) 4 ground electrodes (each 1.8cm²) and 1 reference electrode (1.8cm²). A user puts the patch on the chest to monitor the TIV/ECG signals, and he/she can start and stop the system by inductively coupled power switch with ID verification function. BCC is used to upload recorded data to a central base station when on chip storage is full, and download system command when new configuration is required.

Figure 27.3.2 shows the overall block diagram of the proposed SoC architecture. It consists of 1) a system start-up module for battery turn on and initial BCC frequency allocation, 2) reconfigurable 4 x 4-electrode sensor front ends (RE-FE) for voltage sensing and digitization, 3) a differential sinusoidal current generator (DSCG) for TIV measurement, 4) a digital module containing FSM controller with special purpose registers (SPR), a 20kB data storage, a FIR filter, an 8:1 compression block [4], and a packet encoder/decoder, and 5) a duty-cycled BCT for external communication.

The system starts with 8b ID check which is ASK modulated at 13.56MHz through fabric inductor coupling [3]. A CMOS rectifier generates power-on-reset (POR) trigger signal to demodulate incoming ID packet, and an ID checker (8b-XOR) authenticates the ID with on-chip preprogrammed 8bit code for the flexible battery to turn on the SoC. The start-up module downloads the BCC frequency allocation packet at 0.8kb/s through the same link. ECG is measured (Mode 0) through 8 electrodes (impedance less than 120kΩ in 0.4Hz-1.1kHz) in section 0 and 2 in time multiplexing while TIV is monitored (Mode 1) through 8 electrodes (impedance of 150kΩ at 90kHz) in section 1 and section 3. In order to find the optimum sensing points automatically, it scans 8 different electrodes in 2 sections for 8s using the same injection electrodes. Then the current injection electrodes are changed from section 0-2 (Anode0-Cathode0) to section 1-3 (Anode1-Cathode1) with mode swapping between Mode 0 and Mode 1 as shown in Fig. 27.3.3. The BCT is activated every 4s to listen from base station for 0.1s, and it updates channel assessment with a new system command from the base station. When the 20kB storage is full, the stored data is transmitted through BCT to the base station, and then system resumes the sensing operation.

To eliminate the mismatch between 2 pairs of current injection electrodes, a tetra-polar electrode configuration [6] is adopted for TIV measurement. 2 fully differential amplifiers and RC bridge are used to generate a 90kHz (determined by RC time constant) differential sinusoidal voltage as shown in Fig. 27.3.4. The 2nd harmonic is reduced by the differential signaling, and variable voltage swing (Vsw) is achieved by adjusting the gate voltage of M1, which is controlled by 2b DAC. The Vout is converted to differential current (Vout/R) in the range of 100-350μA. It accommodates the differences of the individual users. The measured spectrum of DSCG shows that the 2nd harmonic is suppressed down to -58dBc, and it equals to 0.81% THD at 250μA

The reconfigurable electrode instrumentation amplifier (REIA) of Fig. 27.3.5, shared by ECG and TIV, is proposed to reconfigure electrodes as shown in Fig. 27.3.3. The 4 negative inputs of REIA share 1 reference electrode for Mode=0 while it connects to 4 distinct electrodes for Mode=1. The current balanced IA [5] keeps Vout constant to obtain 96dB CMRR. To amplify both ECG and TIV signal with the same circuit, a dual band operation is realized by adjusting HFP and controlling the load capacitor at R1 in Fig. 27.3.5. 25.6dB gain in 0.4Hz-1.1kHz and 27.8dB gain in 20kHz-280kHz are realized for ECG and TIV, respectively. The 90kHz modulated TIV is additionally amplified by 20dB before down conversion by recovered in-phase carriers. As shown in Fig. 27.3.6, a 0.1Ω of TIV is measured at sensitivity of 3.17V/Ω.

The FSK modulated BCT of Fig. 27.3.2 operates in 20MHz-40MHz with 8b capacitor bank LC digitally controlled oscillator (LC-DCO). Its initial communication frequency is allocated at system start-up. The BCT wakes up within 1µs, and it is 10 faster than [8]. On the receiver side, an extra 20dB gain stage is cascaded to LNA to enhance noise figure by 3dB for robust operation in the unstable receiver FSK spectrum. The BCT shows -75dBm sensitivity with BER 10^4 at 1Mbps which is 10dB enhancement compared to [8] while consuming only 3.2mW.

The 5mm x 5mm chip is fabricated by 0.18μm 1P6M CMOS process. Figure 27.3.7 shows the chip micrograph and its performance summary. It dissipates only 3.9mW peak power when BCT operates at 1Mbps with sensitivity of -75dBm, and consumes 2.4mW static power when it operates in TIV detection mode with sensitivity of 3.17V/Ω. The user-friendly reconfigurable TIV/ECG monitoring patch enables wearable hemodynamic monitoring for possible healthcare applications.

References:
Figure 27.3.1: Proposed wearable thoracic impedance variance (TIV)/ECG monitoring system.

Figure 27.3.2: Overall block diagram of thoracic impedance variance (TIV)/ECG SoC architecture.

Figure 27.3.3: Reconfigurable operation in thoracic impedance variance (TIV)/ECG sensing.

Figure 27.3.4: Differential sinusoidal current generator (DSCG).

Figure 27.3.5: Reconfigurable electrode instrumentation amplifier (REIA).

Figure 27.3.6: Measured thoracic impedance variance (TIV) with ECG.
Figure 27.3.7: Chip micrograph and its performance summary.