A Low-Power Portable ECG Touch Sensor with Two Dry Metal Contact Electrodes

Long Yan and Hoi-Jun Yoo

Abstract—**This paper describes the development of a low-power electrocardiogram (ECG) touch sensor intended for the use with two dry metal electrodes. An equivalent ECG extraction circuit model encountered in a ground-free two-electrode configuration is investigated for an optimal sensor read-out circuit design criteria. From the equivalent circuit model, (1) maximum sensor resolution is derived based on the electrode's background thermal noise, which originates from high electrode-skin contact impedance, together with the input referred noise of instrumentation amplifier (IA), (2) 60 Hz electrostatic coupling from mains and motion artifact are also considered to determine minimum requirement of common mode rejection ratio (CMRR) and input impedance of IA. A dedicated ECG read-out front end incorporating chopping scheme is introduced to provide an input referred circuit noise of 1.3 μV_{rms} over 0.5 Hz ~ 200 Hz, CMRR of IA > 100 dB, sensor resolution of 7 bits, and dissipating only 36 μW. Together with 8 bits synchronous successive approximation register (SAR) ADC, the sensor IC chip is implemented in 0.18 μm CMOS technology and integrated on a 5 cm × 8 cm PCB with two copper patterned electrodes. With the help of proposed touch sensor, ECG signal containing QRS complex and P, T waves are successfully extracted by simply touching the electrodes with two thumbs.**

Index Terms—**ECG, touch sensor, dry electrode, portable, equivalent sensing circuit model, instrumentation amplifier**

Manuscript received Nov. 28, 2010; revised Dec. 15, 2010.
Dep. EE., KAIST, Korea
E-mail : yanlong@eeinfo.kaist.ac.kr

I. INTRODUCTION

Recently, there are growing demands for increasing personal quality of life. The degree to which a person enjoys the important possibilities of his or her life can be evaluated by measuring personal bio-potentials, such as ECG signal. It is essential nowadays to continuously or repeatedly monitor ECG signals both for the personal healthcare [1-4] and self-emotion management [5, 6]. For the healthcare application, portable medical diagnostics system will not disturbs the patient's daily routine and it also saves the cost of healthcare by preemptive medicine [8, 9]. A low-power Holter Monitor system in Fig. 1 is good example for such purpose to preventively measure the abnormal symptoms and warning the patient or doctor so as to avoid the risk of the irreparable damage to organs or even death. With this approach, a convenient measurement of personal heart rhythms without white coat effect [3] is possible in personal daily life. The measured physiological data is also useful for continuous self-emotion management. For example, the emotion recognition based on the heart rate variability (HRV) extracted from ECG by QRS detection method and the absolute value of the successive beat-to-beat interval difference

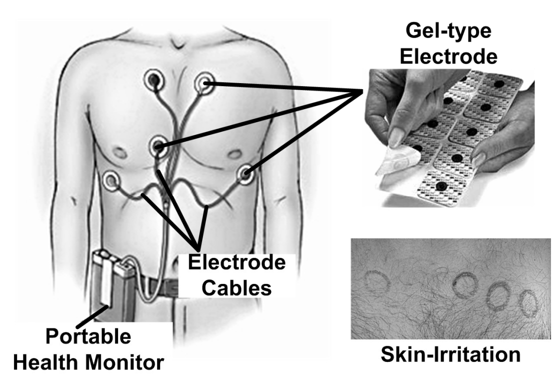


Fig. 1. Holter Monitor for continuously monitor heart rhythms.

can determine the personal mental state of relaxation by a low HRV, whereas a potential state of the mental stress or his/her frustration by an increased HRV [5-7].

The ECG signal, which is an electrical bio-potential originating from excitable cells of heart muscle, is commonly extracted from the patient's skin by attaching electrodes. Most of current approaches use wet or gel-type surface electrodes because it will increase signal quality by improving conductivity between skin-electrode interfaces. However, there is a toxic concern to use wet electrode for long-term application. Skin preparation with alcohol before electrode attachment is a mandatory process to prevent from the bacterial growth which will cause skin irritation. Using wet electrode for repeated ECG monitoring is therefore both expensive and inconvenient. In addition, the electrode cables in Fig. 1, which interconnects between electrodes and sensor monitor, are the most cumbersome part in use. Eliminating the cables by directly integrating the electrode with monitor device and even reducing the numbers of sensing electrodes not only increase patient's mobility but also reduce motion artifacts and interferences.

In this paper, we introduce an ECG touch sensor with two dry metal electrodes integrated on the PCB together with custom fabricated low-power ECG sensor Integrated Circuit (IC) for a low-cost repeated ECG signal acquisition. Based on the proposed ground-free ECG extraction equivalent circuit model, which encounters power-line 60 Hz electrostatic interference, dry contact impedance and its variation, motion artifact, skin-electrode-IA interface and circuit noise, the proposed ECG touch sensor provides optimal sensor resolution.

II. Ground-free ECG Extraction Equivalent Circuit Model

The portable ECG acquisition has to cope with various problems such as interferences and motion artifacts, while extracting weak ECG signals from the thumbs. In generally, portable ECG extraction requires three electrodes, two for the differential signal inputs of IA and one for the ground [10]. This electrode configuration is almost universally adopted because the instrumentation can give high CMRR to suppress power-line interference greatly. However, locating the ground electrode on the same sensor board is an impeding factor. It not only limits small sensor form factor but also dissipates huge power to effectively drive the human body. Several researchers has been reported the feasibility of ECG recording with using only two signal electrodes [11-14]. Despite of its advantages over convenience to use, the sensor development becomes more challengeable. *First*, ground-free electrode configuration introduces increased common-mode 60 Hz electrostatic interference, which couples from surrounding mains to the human body, to the sensor readout circuitry. *Second,* although the metal electrode can be repeatedly used for recording the ECG signals and the differential mismatch of electrode contacts can be minimized by using thumbs to touching the electrodes appropriately, it has larger impedance compared with wet electrode due to its dry contact interface. As a result, it requires solid understanding and careful analysis on several artifacts superimposed to ECG signals before designing the sensor circuitries.

1. Capacitive-coupling to the Human Body

An electrostatic coupling from the 60 Hz AC power line is in-band interference to ECG acquisition. The amount of displacement current coupling to the body determines common mode interference level for the sensor readout circuitry. As a result, a careful investigation on the amount of displacement current is important and it requires to taking into account several coupling capacitances. Among them, the capacitive coupling between human body and earth ground is most important because it determines dominant current path. To determine the value of coupling capacitance between earth ground and the sole of the foot, an experiment is conducted as shown in Fig. 2. A signal generator is used to apply a 1 $V_{p\text{-}p}$ constant voltage signal with a frequency sweep from 10 Hz to 5 kHz via a low impedance Ag/AgCl electrode connected to the foot. To neglects the variable electrode impedance dependency over the frequency, a 1.5 MΩ series resistor is inserted to Ag/AgCl electrode. From the measurement of the voltage on the electrode, 3 dB signal attenuation at 400 Hz is observed which indicates that the 3 dB roll-off bandwidth is basically determined by the coupling

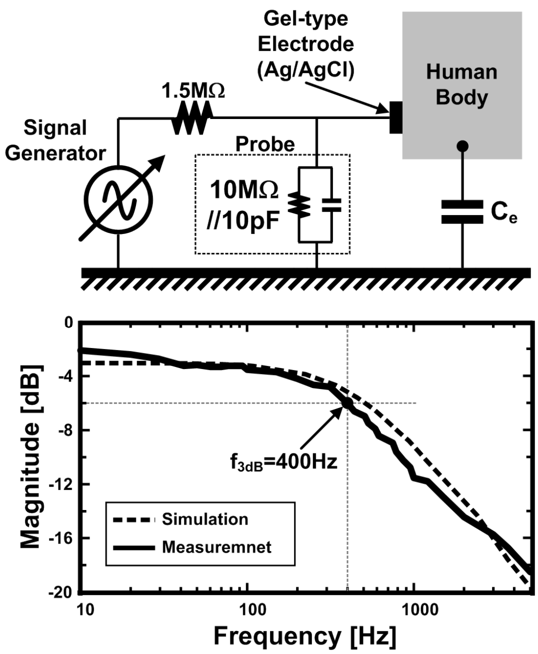


Fig. 2. The measurement of body-to-earth coupling.

capacitor (C_e) of 270 pF from the foot to earth ground.

Another important coupling capacitance (C_p) is from 220 V, 60 Hz power line to human body. To determine its value, the voltage of the 60 Hz interference in time domain is measured by an oscilloscope with its input impedance of (10 MΩ//10 pF) as shown in Fig. 3. During the measurement, a same type Ag/AgCl electrode is

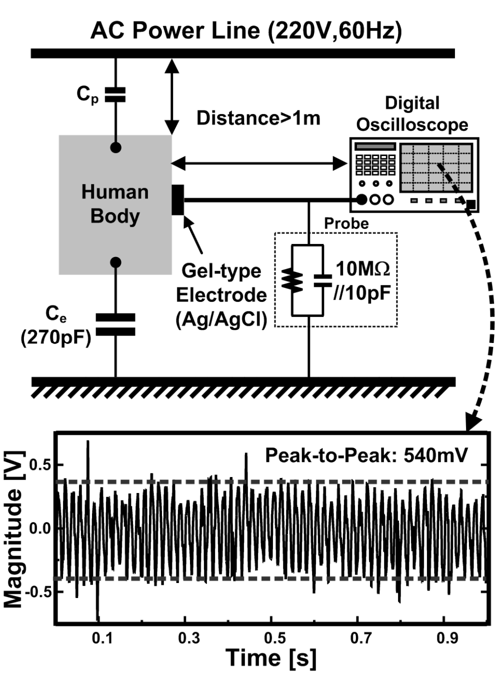


Fig. 3. The measurement of AC power-line to body coupling capacitance.

attached on the arm, and the subject is kept away from the equipment by > 1 m to minimize undesired electrical coupling. In such circumstances, 540 mV_{p-p} interference at 60 Hz is measured. Considering C_e of 270 pF exhibits 9.83 MΩ at 60 Hz which is a similar value of the input impedance (10 MΩ) of oscilloscope, the measured voltage of 540 mV_{p-p} can be assumed to be the result of the voltage division between C_p of 1 pF and C_e//10 MΩ//10 pF. Since C_p dominantly determines the body induced displacement current level from 60 Hz power-line, an induced current of 80 nA is roughly determined.

2. Skin-electrode Interface

Surface metal electrodes can convert the ionic current originated from the contraction of the heart muscle within the body into electrical current. Previously, several types of metals were investigated as a dry contact electrode [15]. The impedance fluctuation of the different metal electrode is less than one decade in the frequency range below 1 kHz compared with Ag/AgCl electrode as the reference. In this work, a 1.2 cm diameter copper metal directly patterned on the PCB as the dry electrode. Since its dry contact, the high skin-electrode impedance must be characterized to interface with sensor readout circuitry. With the method reported in [16], the measurement was started after 5 minutes electrode attachment for stabilizing the contact. To maintain an equal contact during the measurement, a constant pressure of 20 mmHg is applied on electrode with the help of pressure cuff. According to the measurement results in Fig. 4, the dry copper electrode has the impedance about 3 times larger than the reference electrode (Ag/AgCl electrode with 0.9% saline as an electrolytic solution) in the desired bandwidth (0.5 Hz ~ 200 Hz). Equivalent passive circuit model parameters are extracted by spice simulation as shown in Fig. 4. The dominant impedance in the frequency range below 100 kHz is governed by a resistor (75 kΩ) parallel with a capacitor (40 nF). Such high series resistance is created by the absence of conductive paste, and the large capacitance is created because of the partially polarized skin-electrode interface. Another important parameter for the skin-electrode interface is DC half-cell potential (V_{hc}), which develops across the skin-metal interface due to an uneven distribution of anions and cations. It appears as a

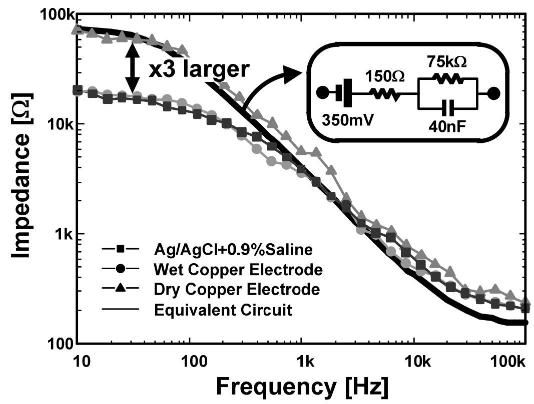


Fig. 4. Impedance characteristics of copper electrode.

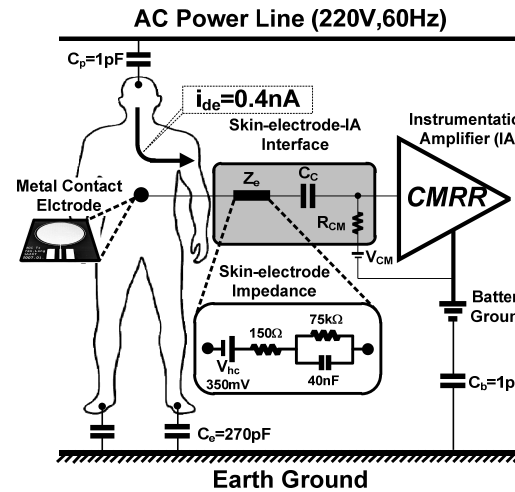


Fig. 5. Ground-free ECG extraction equivalent circuit model.

DC offset voltage of 350 mV. Different pressure applied on the electrode will yield the variation of ΔV_{hc} and the electrode contact impedance (ΔZ_e). As a result, the motion artifact potential (V_{ma}) can be represented as Eq. (1), where R_{CM} represents common input impedance of IA.

$$V_{ma} = V_{hc} + \frac{\Delta Z_e}{R_{CM}} \Delta V_{hc} \tag{1}$$

From Eq. (1), it shows that minimizing ΔZ_e and ΔV_{hc} while maximizing R_{CM} can minimize variation of V_{ma}. With such, AC coupling rather than DC coupling is preferred to filtering the V_{hc} prior to IA's amplification, and extracting ECG at thumbs can minimize both ΔZ_e and ΔV_{hc} by pressing the electrode appropriately to prevent from signal distortion.

3. Crucial Requirements on Ground-free ECG Sensor IC Design

Fig. 5 shows ground-free ECG extraction equivalent circuit model. A DC blocking capacitor of C_c with input common mode bias resistor of R_{CM} is added prior to IA. In the frequency range less than 200 Hz, the transfer function of G_{HPF} (skin-electrode-IA's input) can be simplified by Eq. (2)

$$G_{HPF} = \frac{s(1 + sR_eC_e)}{R_eC_es^2 + s + 1/R_{CM}C_c} \tag{2}$$

where $C_c(1uF) >> C_e$ and $R_{CM}(10M\Omega) >> R_e$ to form a high pass corner below 0.5 Hz and to minimize large electrode's impedance loading effect.

Fig. 6 shows the simulated transfer curve (G_{HPF}) of the skin-electrode-IA interface. The G_{HPF} creates two zeros at DC and $1/R_eC_e$, respectively, and a dominant pole at low frequency below 0.1 Hz with second pole close to second zero to mitigate phase distortion. A maximum 2.5° phase shift with 0.6 dB gain variance is introduced by the change of the electrode's capacitance in the range from 1 nF to 100 nF, which encounters the poor electrode's contact with skin.

ECG signal extracted at thumbs is very weak which is in the range of 0.1 mV ~ 1 mV. After HPF, it is then amplified by IA. Two design considerations for the IA must be taken into account. *First,* high CMRR is required to suppress high in-band common mode interference at 0.5 Hz ~ 200 Hz. To determine minimum required CMRR in ground-free electrode configuration, a

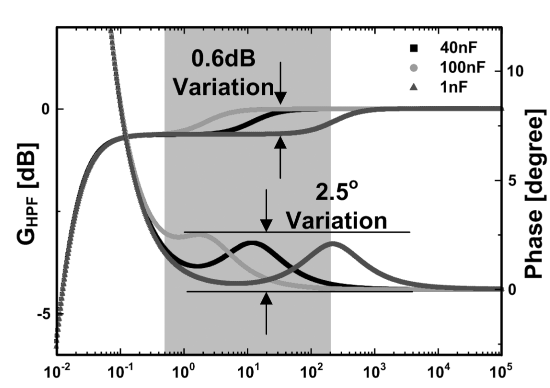


Fig. 6. Simulated transfer function of the skin-electrode-IA interface.

careful analysis on Fig. 5 is needed. By assuming equal current (i_{de}) induced into a pair of electrodes and common input impedance of the IA, the electrical interference (V_{ei}) introduced by displacement current can be simplified to Eq. (3)

$$V_{ei} = \frac{i_{de}}{2}\Delta Z_e + \frac{i_{de} \cdot R_{CM}}{2\,CMRR} \tag{3}$$

where i_{de} is around 0.4 nA and $\Delta Z_{e,max} < 40k\Omega$. The first term indicates an electrical interference created by skin electrode impedance mismatch for assuming the worst case which is in the order of 10 μV. Hence, the electrical interference represented in the second term of Eq. (3) must be suppressed down to μV level so as to V_{ei} as 10% of minimum ECG signal amplitude. With this requirement, The IA is desirable to have a CMRR > 60 dB with R_{CM} of 10 MΩ.

To define the overall sensor resolution, noise from both passive elements and IA its self must be encountered. The first noise source is thermal noise generated by R_{CM} 10 MΩ. However, its contribution to ECG signal band is negligible due to the noise filtering by C_C of 1 μF. The second noise source is the thermal noise generated by the electrode of $Z_e \approx 75$ kΩ. Its amount can be estimated by Nyquist's formula of Eq. (4)

$$V_{en} = \sqrt{4k_B \cdot T \cdot Z_e \cdot BW} = 0.48\,\mu V_{rms} \tag{4}$$

where k_B(1.38 × 10^{-23} J/K) is Boltzmann's constant, T(300K) is the temperature, BW (200 Hz) is the IA bandwidth. Finally, IA noise will be added during the ECG signal amplification. The amount of noise contribution is dominated by flicker noise and offset drift in the low frequency which calls for a proper cancellation technique to prevent from saturating the high gain IA.

With reference voltage of 800 mV and minimum gain of 46 dB in this design, 8 bits resolution for the sensor readout circuitry is chosen to resolve minimum input signal down to 15.6 μV which is not only much lower than ECG signal (minimum 0.1 mV) but also comparable with introduced total artifact signals (60 Hz interference and noise) as well as easy for external post digital signal processing.

III. ARCHITECTURE OF SENSOR IC

Both enhancement of the CMRR and noise performance is important in ground-free ECG acquisition while consuming only microwatts level power for maximizing the battery life time. Fig. 7 shows the architecture of ECG sensor IC [17]. It mainly consists of (1) electrode interface for motion artifact filtering discussed in Section II-2 and II-3, (2) analog read-out front end including chopping amplifier as IA and programmable gain stage for enhanced dynamic range, (3) 8 bits SAR ADC for the continuous digitization of the ECG signal with low energy consumption, and (4) some auxiliary functional blocks including an reference generator [18] for ADC,

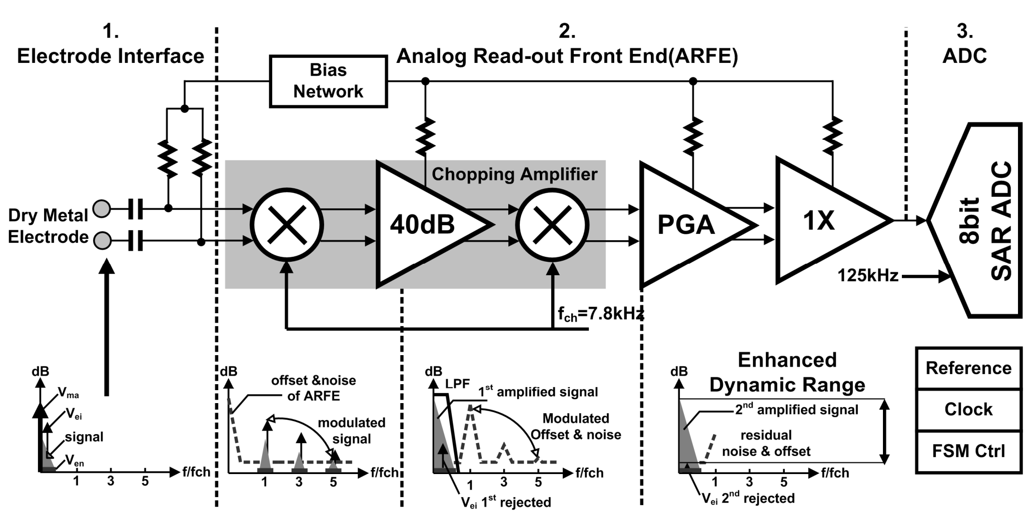


Fig. 7. Architecture of sensor IC and signal flow.

on-chip current starving oscillator, and FSM controller for the various clock generation and sensor parameter control.

A 0.5 Hz passive high-pass-filter (HPF) is implemented with off-chip capacitors of 1 μF and on-chip input common mode resistors of 10 MΩ. To dynamically mitigate input referred noise and offset caused by any mismatch introduced by readout front end, a chopper modulation technique [19, 20] is adopted for dynamic offset cancellation. A 7.8 kHz modulation and demodulation chopper switch is enclosed to 2-stage amplifier. It provides 40 dB mid-band gain and input referred noise power spectral density of 40 nV/√Hz while consuming only 2 μA from 1.8 V supply voltage.

A 1st-order low pass filter (LPF) is cascaded after IA stage to limits the bandwidth by 200 Hz for filtering out the modulated in-band 1/f noise, thermal noise and any other interferences. To further enhance ECG signal dynamic range before digitization, 6 dB ~ 30 dB mid-band gain is provided by programmable gain amplifier [17]. Cascaded by an 8 bits SAR ADC with unified sequencer and code resister [17], it can provides 11 kS/s ECG data throughputs with overall energy consumption below 3 nJ/conversion.

IV. Implementation and Measurement Results

The sensor IC was verified by successively recording ECG signal with two copper contact pads (each 4.5 cm^2 and separated by 6 cm). Fig. 8 shows the proposed ECG touch sensor with its form factor of 5 cm × 8 cm. A BGA packaged sensor IC chip, which is fabricated by 0.18 μm CMOS technology and occupies 1.2 mm × 2 mm including pads, is mounted on the PCB and it can continuously operate for a week with single 1000 mAh 3.3 V battery. Table 1 summarizes the performance of the sensor IC. The analog read-out front end provides 46 dB ~ 70 dB variable gain with 2 bits gain control, and its input referred noise performance is 1.3 μV_{rms} over the integrating bandwidth from 0.5 Hz ~ 200 Hz. To verify CMRR performance, a 60 Hz common mode signal with 100 m$V_{p\text{-}p}$ magnitude is applied to analog read-out front end, and it performs -37 dB rejection with the differential

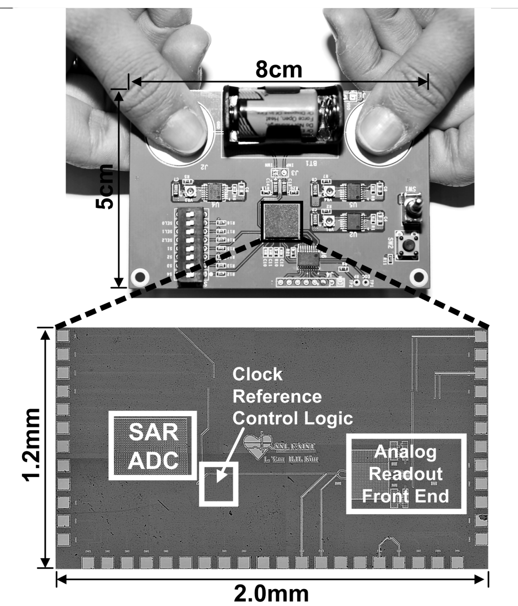


Fig. 8. Implemented ECG touch sensor.

Table 1. Performance Summary of ECG Sensor IC

Measurement Factors	Value
Electronic Gain Selection [dB]	46 ~ 70
Electronic Bandwidth [Hz]	0.5 ~ 300
Input Refered Noise Voltage [Vrms]	1.3 Vrms (@ 0.5~200Hz)
Total Harmonic Distortion(THD)	< 1%(@ minimum Gain)
CMRR & PSRR(@60Hz)	> 100dB
Power Consumption	40.2 uW (125 KHz)
DNL /INL	< ±0.6 LSB Typical
Effective Resolution	7-bit
Energy Per Sample	2.88 nJ/Conversion

signal gain of 64 dB as shown in Fig. 9, which is sufficiently high to mitigate V_{ei} of Eq. (3).

The resolution of the proposed sensor is verified by applying a 4 m$V_{p\text{-}p}$ sinusoidal voltage signal with frequency of 100 Hz to analog read-out front end. After 46 dB signal amplification, a spurious free dynamic range (SFDR) of 48.2 dB is measured with ADC sampling rate of 1 kHz as shown in Fig. 10(a). It is equivalent to 6.98 effective number of bit (ENOB) with its signal-to-noise and distortion-range (SNDR) of 43.8 dB. In addition, the measured linearity in Fig. 10(b) shows the sensor performs ±0.6 LSB of DNL/INL at minimum gain configuration of 46 dB.

ECG signals in Fig. 11(a) are recorded at the thumbs with the gain of ×200 and ×400. The proposed ECG touch sensor with gain of ×400 clearly tracks the QRS

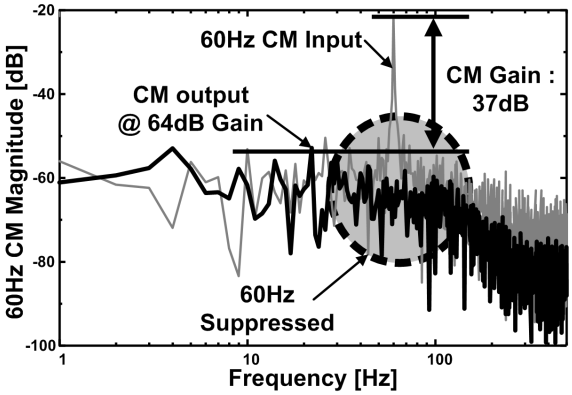
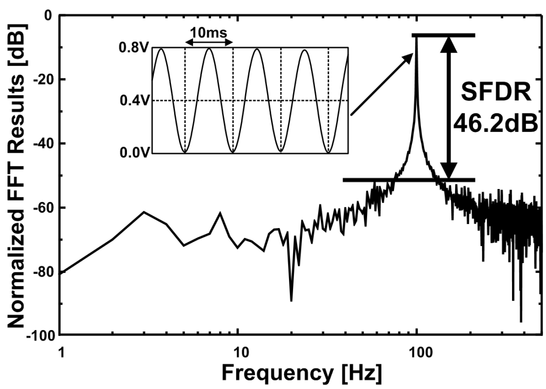
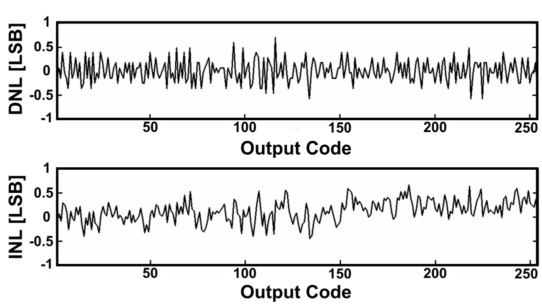


Fig. 9. Measurement of common mode rejection in regards to 60 Hz interference.



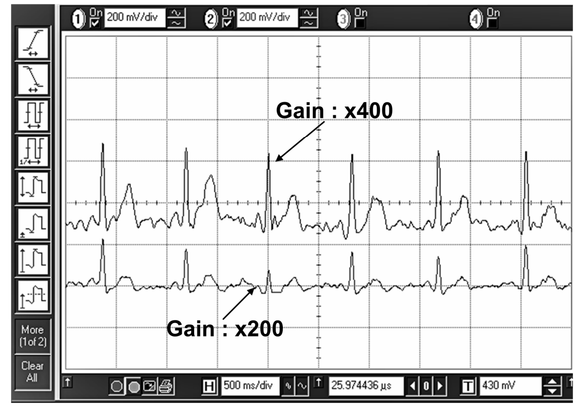
(a)



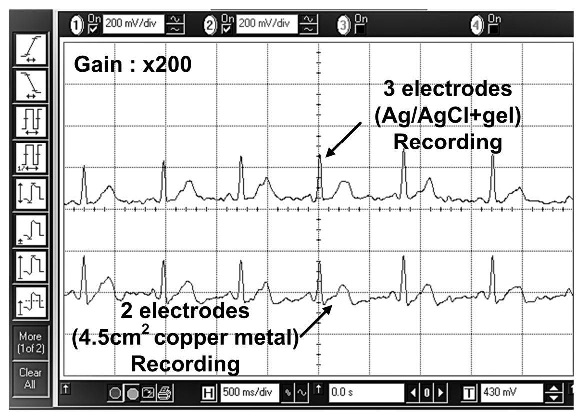
(b)

Fig. 10. Sensor resolution and linearity with respect to 100 Hz 4 mV_{p-p} sine input (a) FFT results of ADC output and (b) DNL/INL.

complex and P, T waves without 60 Hz interference. Fig. 11(b) shows the waveform comparison between ECG signal recorded by 3 electrodes (Ag/AgCl+gel) simplified ECG sensor introduced in [21] and proposed ground-free (4.5 cm^2 copper pad each) ECG touch sensor. Both sensor provides amplification gain of ×200 and applied to forearms and thumbs, respectively. It shows the proposed ECG touch sensor provides comparable signal quality while the subject sitting at rest.



(a)



(b)

Fig. 11. Recorded ECG waveforms. (a) ECG recorded by proposed ECG touch sensor with ×200 and ×400, (b) Compared ECG waveforms with 2 and 3 electrodes.

V. CONCLUSIONS

We propose a low-power ground-free ECG touch sensor which is suited for the energy constrained portable healthcare application. Its design considered 60 Hz power-line interference without ground electrode and the effect of dry metal electrode interface with skin. The ECG sensor board includes 2 copper contact pads and a low-power sensor IC chip with single battery. With the proposed ECG touch sensor, continuously and repeatedly monitoring personal heart rhythms is possible in a convenient way by simply touching the electrode pads with thumbs.

REFERENCES

[1] Fact sheet for obesity and cardiovascular disease: A report from the American Heart Association, 2009.

[2] D. Jabaudon, et.al, "Usefulness of ambulatory 7-day ECG monitoring for the detection of atrial fibrillation and flutter after acute stroke and transient ischemic attack," *Stroke, J.Amer. Heart Assoc.*, Vol.35, pp.1647-1651, May, 2004.

[3] E.Rodriguez-Cruz, Hypertension : Overview-eMedicine. Available : http://emedicine.medscape.com/article/889877-overview

[4] Task Force of the Eourpean Society of Cardiology and the North American Society of Pacing and Electrophysiology, "Heart Rate Variability: Standards of Measurement," Physiological Interpretation, and Clinical Use, 93 1996.

[5] C.Jing, et al, "The Research on Emotion Recognition from ECG Signal," *International Conference on Information Technology and Computer Science*, pp. 497-500, 2009.

[6] A.Haag, et al, "Emotion Recognition Using Bio-Sensors: First Steps Towards an Automatic System," *Proceedings of the Kloster Irsee Tutorial and Research workshop on Affective Dialogue Systems,* pp.36-48, 2004.

[7] Rigas, et.al, "A User Independent, Biosignal Based, Emotion Recognition Method," *Lecture notes in Computer Science,* No.4511, pp.314-318, 2007

[8] J. Yoo, et.al, "A Wearable ECG Acquisition System With Compact Planar-Fashionable Circuit Board-Based Shirt," *IEEE Transactions on Information Technology in Biomedicine*, Vol.13, pp.892-902, 2009.

[9] A. C-W. Wong, et al, "A 1V Micropower System-on-Chip for Vital-Sign Monitoring in Wireless Body Sensor Networks," *ISSCC Dig. Tech. Papers,* pp.138-139, Feb., 2008.

[10] E.M. Spinelli, et.al, "A Transconductance Driven-Right-Leg Circuit," *IEEE Transactions on Biomedical Engineering,* Vol.47, pp.1466-1470, 1999.

[11] S.N. Pocock, "Earth-free patient monitoring, Part 1: principles, practice and hazards," *Biomed. Eng.,* Vol.7, pp.21-25, 1972.

[12] E.M. Spinelli, et.al, "Two-Electrode biopotential measurements: Power line interfere Analysis," *IEEE Transaction on biomedical engineering*, Vol. 52, pp.1436-442, Aug. 2005.

[13] D.E.Wood, et.al, "Comparative analysis of powr line interference between two-or three-electrode biopotential amplifiers," *Medical & Biological Engineering & Computing,* Vol.33, pp.63-68, 1995.

[14] N.V. Thakor, et.al, "Ground-free ECG Recording with Two Electrodes," *IEEE Transaction on biomedical engineering*, Vol.BME-27, pp.699-704, Dec. 1980.

[15] K. Hachisuka, et al, "Development and performance analysis of an intra-body communication device," in *Int. Solid-State Sens., Actuators, Microsyst. Conf.*, pp.1722-1725, 2003.

[16] George E. Bergey, et al, "Electrocardiogram recording with pasteless electrodes," *IEEE Trans. on Bio-medical Engineering,* Vol.18, pp.206-211, 1971.

[17] Long Yan, et al, "A Two-Electrode 2.88nJ/Conv. Biopotential Acquisition System for Portable Healthcare Device," *Proc. Of the IEEE Asian Solide-State Circuit Conference (A-SSCC),* pp.329-332, Nov., 2008.

[18] K.N.Leung and P.K.T.Mok, "A CMOS Voltage Reference based pm weighted ΔV_{GS} for CMOS Low-Dropout Linear Regulators," *IEEE Journal of Solid-State Circuits*, Vol.38, pp.146-150, Jan., 2003.

[19] Refet Firat Yazicioglu, Chris Van Hoff, et.al, "A 60 uW 60 nV/srtHz Redout Front-End for Portable Biopotential Acquisition Systems," *IEEE Journal of Solid-State Circuits*, Vol.42, pp.1100-1109, May., 2007.

[20] Christian C. ENZ and Gabor C. Temes, "Circuit Techniques for Reducing the Effects of Op-Amp Imperfections: Autozeroing, Correlated Double Sampling, and Chopper Stablization," *Proc. IEEE*, Vol.84, pp.1584-1614, Nov., 1996.

[21] Datasheet of INA321, Simplified ECG Circuit for Medical Applications. Available : http://focus.ti.com/lit/ds/sbos168d/sbos168d.pdf

Long Yan (S'07) received the B.S. and M.S. degrees in Electrical Engineering from Korea Advanced Institute of Science and Technology (KAIST), Daejeon, Korea, in 2007 and 2009, respectively.

He is currently a Ph.D. candidate in Electrical Engineering at KAIST and also with Microsystems Technology Laboratory, Massachusetts Institute of Technology (MIT), Cambridge, MA, USA, as a visiting student. As a chief researcher at the Semiconductor System Laboratory in KAIST, He has worked on developing low energy FSK transceivers for Wireless Body Area Network and low noise, wirelessly powered patch sensors for Wearable Body Sensor Network. His current research focuses on low power low noise sensor readout circuit technology for wearable healthcare sensors, low energy near field communication around body, SoC design to system realization for wearable healthcare applications, and power management in biomedical micro-system.

Hoi-Jun Yoo (M'95 – SM'04 – F'08) graduated from Electronic Department of Seoul National University and received MS and Ph.D. degrees from Electrical Engineering, KAIST. He was the VCSEL pioneer in Bell Communications Research at Red Bank, NJ and Manager of DRAM design group at Hyundai Electronics designing from 1M DRAM to 256M SDRAM.

Currently, he is the full professor of Department of Electrical Engineering at KAIST and the director of SDIA (System Design Innovation and Application Research Center). From 2003 to 2005, he was the full time Advisor to the Minister of Korean Ministry of Information and Communication for SoC and Next Generation Computing. His current research interests are Bio Inspired IC Design, Network on a Chip, Multimedia SoC design, Wearable Computing, Robot SoC, and high speed and low power memory. He published more than 200 papers, and wrote 5 books, "DRAM Design" (1997, Hongneung), "High Performance DRAM" (1999 Hongneung), "Low Power NoC for High Performance SoC Design" (2008, CRC), "Mobile 3D Graphics SoC" (2010, Wiley), and "BioMedical CMOS ICs" (2010, Springer), and many chapters of books.

Dr. Yoo received the Electronic Industrial Association of Korea Award for his contribution to DRAM technology the 1994, Hynix Development Award in 1995, the Korea Semiconductor Industry Association Award in 2002, Best Research of KAIST Award in 2007, Design Award of 2001 ASP-DAC, and Outstanding Design Awards 2005, 2006, 2007 A-SSCC. He is a member of the executive committee of ISSCC, Symposium on VLSI, and A-SSCC. He was the TPC chair of the A-SSCC 2008, and is the TPC Chair of ISWC 2010, IEEE Fellow, IEEE Distinguished Lecturer ('10-'11), and Far East Chair of ISSCC ('10-'11).