Design of a Gel-less Two-Electrode ECG Monitor

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Abstract — **In this paper, an ECG amplifier design, specifically to interface two gel-less electrodes for low-power portable applications, is presented. The goal is to develop a circuit with performance sufficient to extract heart rate information reliably using digital signal processing techniques, when measuring a subject engaging in moderate physical activity (such as walking). This application, having no reference electrode, requires biasing of the input to avoid amplifier saturation, all the while maintaining a sufficiently high common-mode noise rejection. A baseline correction circuit and an isolated circuit ground are included in the proposed design to handle changes in DC voltage drift due to electrode movement as well as to minimize common-mode noise from mains.**

Keywords - heart rate monitoring;ECG amplifier; two-electrode ECG; portable medical device

I. INTRODUCTION

Heart rate and its derivative, the heart rate variability, provide valuable information about a person's physiological state and health [1][2]. Since heart rate (HR) can be extracted from a simple electrocardiogram (ECG) signal [3], it is a good candidate for embedding into commonly used health appliances for the aging population. To this end, heart rate monitoring is presently being included in the design of the Smart Rollator [4].

A heart rate monitor typically begins with acquisition of the ECG signal through a set of skin electrodes. A specialized ECG amplifier is used to isolate the signal from most of the noise and further processing, either in hardware or software, extracts the heart beats.

Power-line interference is a major component of noise in the ECG signal. This results from capacitive coupling to the body, electrodes, leads and circuit, as well as magnetic coupling to the loop created by the electrode leads [5][6].

The common-mode (CM) signal from the body is typically the largest of these noise components. Its effects can be greatly mitigated by (a) using a differential amplifier with high input impedance and common-mode rejection ratio (CMRR), (b) reducing the source impedance of the input (i.e. reducing the skin-to-electrode interface impedance through skin preparation), and (c) effectively decoupling the circuit ground from earth ground at power line frequency (60 Hz). Methods of doing these have been extensively discussed, simulated and tested in the literature [7][8][9][10].

What remains of this 60 Hz interference can be filtered using analog circuitry (such as a twin-T notch filter) [11][12] or, as is more common now, by using digital processing techniques (e.g. a finite impulse response notch filter) [13]. The challenge, in both cases, is in aligning the high-Q notch to be and stay exactly at the line frequency and is solved by using tracking techniques. Digital algorithms have an added benefit of being able to deliver a linear phase response throughout the passband, an important characteristic needed to reduce ECG waveshape distortion [14][15].

Active front-ends, in which current sources at the amplifier inputs create distinct DM and CM input spectral impedances, have also been presented as a solution [16][17], though at the expense of added circuit complexity.

Dry electrodes, in addition to requiring no skin preparation prior to use, can now provide comparable or better biosignals and produce less motion artifact than standard gel electrodes [18][19]. However, these electrodes have different characteristics and require different amplifier designs than standard Ag/Ag-Cl electrodes. ECG amplifiers specifically designed for dry electrodes provide high fidelity ECG output, meeting American Heart Association (AHA) requirements for electrocardiographic equipment [7][10][20].

The most distinguishable part of the ECG waveform is the QRS-complex. This complex may be detected by specialized analog circuitry [21] or by using a detection algorithm on the digitized ECG signal [3][22]. Once tagged, the heart rate, which is the inverse of the RR-interval, can be easily calculated.

Two-electrode ECG amplifier designs have also been proposed, with much discussion and analysis on eliminating CM noise without having to drive the body with a reference signal [23][24][25].

In this paper we study the challenges when using grip-style dry electrodes for ECG measurement during physical activity and present a new design for a portable ECG amplifier which mitigates some of the identified problems.

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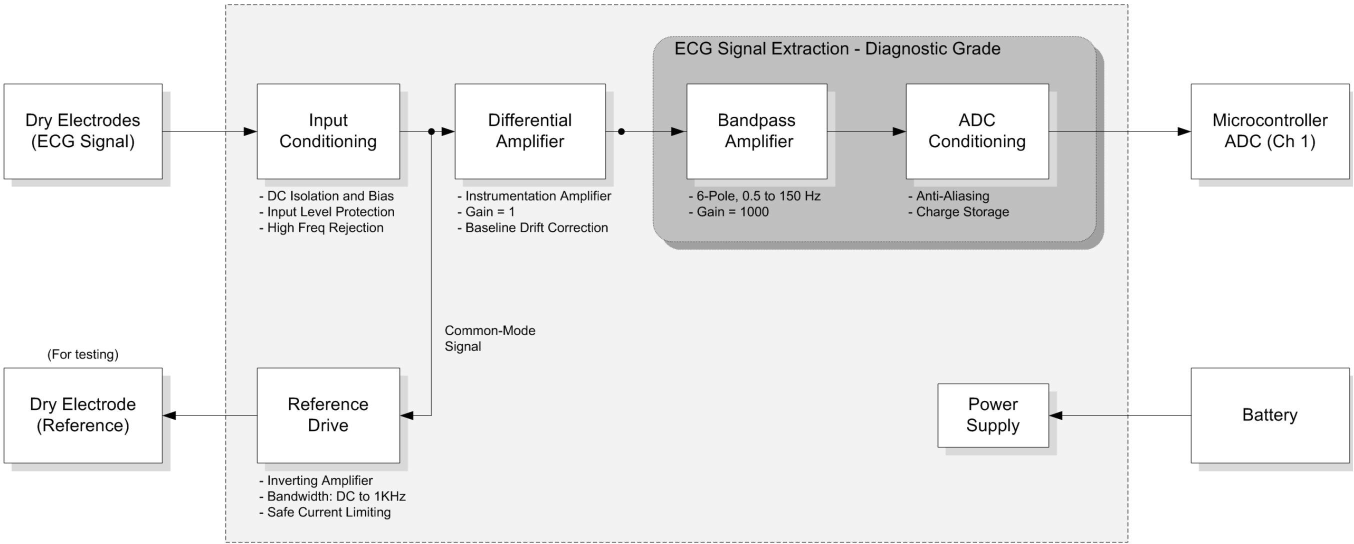


Figure 1: Block Diagram of 2-Electrode ECG Amplifier with Optional Driven Reference Electrode

II. SYSTEM DESCRIPTION

The main design challenges in this application are that:

1. Dry electrodes, compared to conventional Ag/AgCl gelled electrodes, typically have a wider variation of skin-to-electrode impedances due to contact quality. As a result, amplifier input impedances and bias currents will play a larger role in creating DM artifacts from the CM signal than would otherwise be the case.

2. During normal use, it is expected that the user will be changing the tightness of their grip and the weight placed on the handle grips. This will have an effect of creating large DC changes at the electrode inputs. Such large baseline drifts could easily saturate the signals within the amplifier circuit and even if not, if these are of a high enough frequency, may confound the ECG and HR processing algorithms.

3. The user, frequently being in motion, will have a dynamically changing coupling to both ground and the power lines. Not having a third (reference) electrode to help control the CM voltage of the subject's body directly, another strategy for managing the effects of this large and changing low-frequency noise source is needed.

As such, the following aspects of the design were identified as needing particular attention.

A. *Common-mode noise rejection*

CM rejection is primarily achieved by using an instrumentation amplifier (IA) which has a high common-mode rejection ratio (CMRR) and by isolating, as much as possible, circuit ground from earth ground. Doing the latter in effect "floats" the circuit, the amplifier inputs following the CM signal, with the IA, as a result, seeing little or no CM signal. What CM signal remains at the IA input will be reflected in the IA's output as per its CMRR specification.

B. *Low electrode current*

Any current passing through an electrode will result in voltage drops across all impedances between the signal source (e.g. the heart muscle) and the IA. The impedance which is greatest in this signal path is typically at the skin-to-electrode interface, and can typically be in the 10's to 100's of K Ohms, varying by electrode contact area and quality. When there is a difference between the voltage drop across each skin-electrode interface, and in particular, a changing differential, such as by movement, that noise will manifest itself in the DM signal. CM rejection techniques will not be effective in mitigating this DM noise.

C. *Signal fidelity*

To accurately reproduce the ECG for diagnostic purposes, according to standards being recommended by the AHA, the signal path requires a bandwidth of about 0.5 to 150 Hz and to be flat in this passband within +/- 3 dB. The AHA recommends a single pole HPF at 0.5 Hz and a double pole LPF at 150 Hz. It is to minimize noise content as well as for anti-aliasing prior to digital conversion that the signal should be attenuated outside of this range. Adhering to this standard also makes the signal suitable for processing by many signal processing algorithms and libraries. However, having as wide a bandwidth as this will pass significantly more noise to the digital processor, and so this noise will need to be managed using digital techniques.

D. *Low power*

Low power CMOS-based operational amplifiers (OAs) and IAs were used in the design, to keep supply current requirements down and maximize battery longevity. Passive components (resistors, in particular) were chosen such that currents in their circuit paths would not, as a rule, exceed 100 uA, on average, during operation.

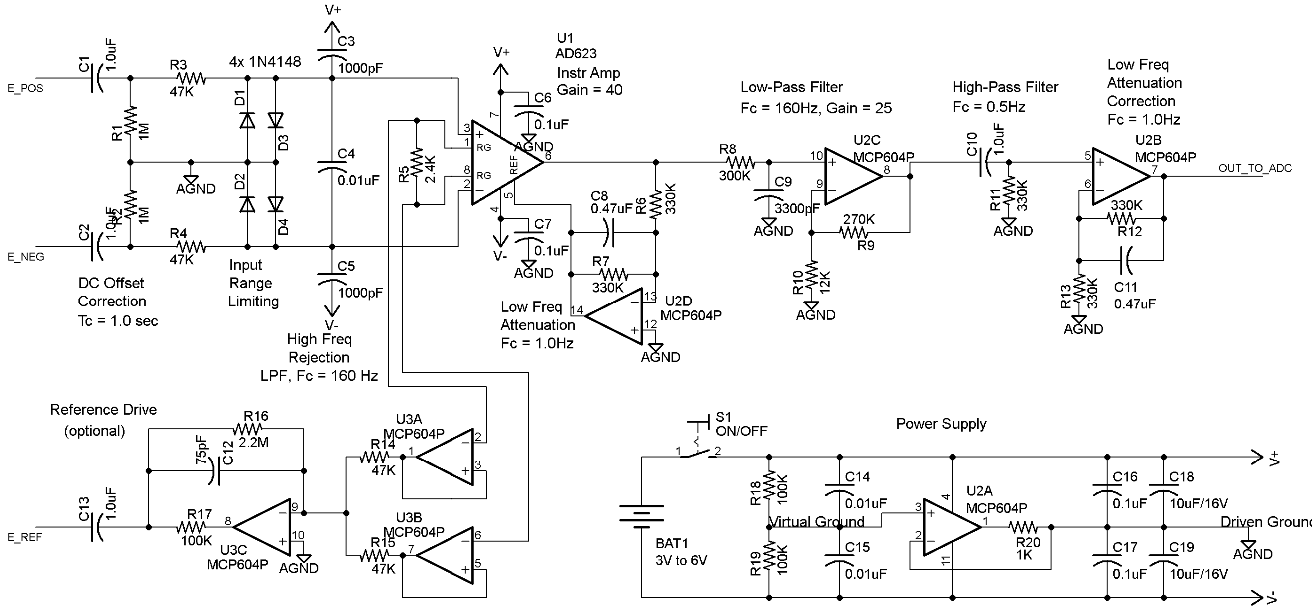


Figure 2: Schematic of 2-Electrode ECG Amplifier with Optional Driven Reference Electrode

III. Results

A. Circuit Design

The proposed ECG amplifier (Fig. 1) consists of an input conditioning stage, followed by a differential amplifier and a bandwidth-limited amplification stage prior to digital conversion. A reference drive section is also included, to compare the effect of biasing the body to reduce CM noise and artifacts. Finally, a power supply section is responsible for creating a clean bipolar supply from a single battery.

The input conditioning stage provides safety to the user through DC-isolation (blocking capacitors (C1 and C2, in Fig. 2) and current limiting (resistors R3 and R4). Resistors R1 and R2 DC-bias the instrumentation amplifier (IA) inputs, whereas diodes D1 through D4 clamp the IA inputs to a safe voltage. These diodes also help charge or discharge the input capacitors (through the current limiting resistors) during large or fast changes in differential voltage at the electrode inputs (E_POS and E_NEG), in order to speed up recovery time following electrode movement. Capacitors C3, C4 and C5 quench high frequency signals at the IA inputs, as per manufacturer's recommendations.

The differential amplifier section consists of a high-performance single-IC IA (AD623). It can be programmed for high gain (RG = 2.4K Ohms, for a gain of 40) as DC offsets and excursions are managed at the input conditioning stage. Nevertheless, to maximize signal headroom and avoid signal saturation, a 3dB low-frequency attenuator (U2D, R6, R7 and C8) drives the output stage of the IA such that any low frequency or DC-offset at the input of the IA spans the positive and negative range equally. Another way of looking at this is that C8 at high frequencies makes U2D a follower, driving the IA reference to ground, whereas at low frequencies makes U2D an inverter, subtracting half of the IA's output by biasing its reference negatively. This effectively reduces the periods of saturation, internally in the IA as well as in the following stages (U2C and U2B).

The bandpass amplification stage is designed to meet the 1990 AHA standards and maintain integrity of the ECG signal, with corner frequencies of 0.5 and 160 Hz. The AHA recommends a first order high-pass (provided here by R11 and C10), as well as a second order low-pass filter (produced by R8, C9, in addition to R3, R4, C3, C4 and C5 at the input stage). The system's cutoff frequency of 160 Hz also provides effective anti-aliasing when converting at speeds above 300 SPS. An overall gain of 1000 brings the sensed 2 mVp-p signal

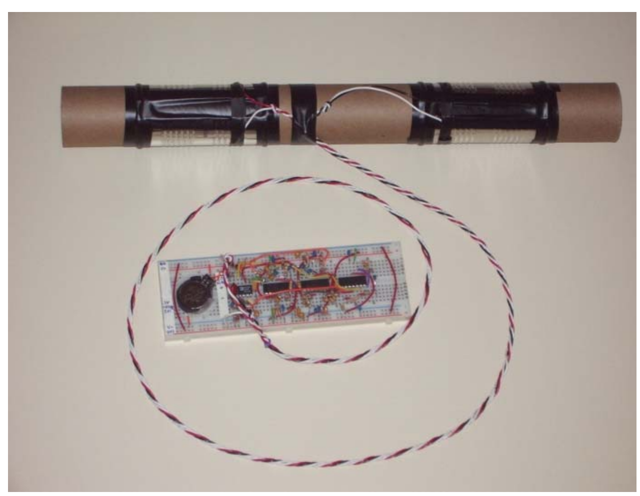
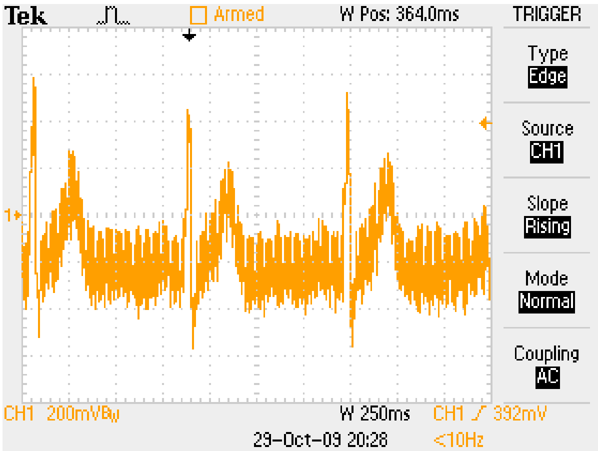
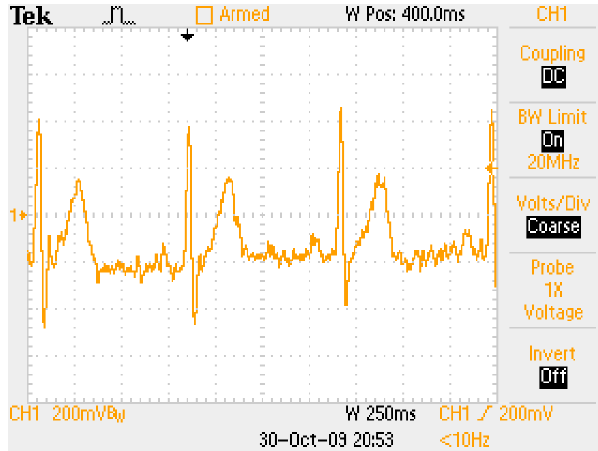


Figure 3: Prototype 2-Electrode ECG circuit and grip-style electrodes.



(a) Bandpass: 0.5 to 100 Hz



(b) Bandpass: 5 to 30 Hz

Figure 4: Capture of 2-wire ECG signal (a) without and (b) with additional bandwidth limiting.

to a scale suitable for digital conversion.

The frequency response and (more importantly) the phase change introduced by the IA biasing circuit (U2D, C8, R6 and R7) are reversed by a corresponding correction circuit (U2B, C11, R12 and R13).

A reference drive section was added to the design in order to assess whether forcing the body's CM voltage would improve the response of the system. This forcing is done by taking the CM level of the input (the midpoint between the voltages at the RG inputs), inverting and amplifying, and then driving this voltage into the body. The result is that the CM input (i.e. body) voltage is kept at (or close to) circuit ground. A current limiting resistor (R17) limits the maximum possible drive current to a safe level (less than 100 uA p-p) and a blocking capacitor provides DC-isolation, to add an extra layer of user safety.

B. *Prototype System*

A prototype system was built, consisting of a breadboarded version of this circuit connected to custom built grip-style electrodes (Fig. 3). The circuit is powered by a 220 mAh 3V pill battery (CR2032), also mounted on the breadboard.

An electrode system was assembled in order to easily test the circuit on the bench. It consisted of two 10 by 10 cm tin plates wrapped on a sturdy insulated tube. Tin was chosen due to its availability and its low corrosion properties. A more appropriate set of conductive rubber electrodes will be mounted on a test Rollator for further study.

C. *Power Usage*

The proposed circuit was measured as consuming 447 uA during operation. Given that the circuit can be expected to operate normally down to about 2.7 V (per the AD623 data sheet) we could expect, from one 3V lithium pill battery, based on its 0.5 mA discharge curve [26], about 400 hours of continuous operation.

In practice, this circuit would be part of a microcontroller–based acquisition system which, if designed with an ability to turn this ECG amplifier on as needed, could further reduce the overall power usage.

D. *Preliminary Measurements*

As an initial verification of the circuit's operation, an ECG was acquired from an individual at rest, using the circuit with two gelled electrodes and no reference electrode. Fig. 4a illustrates its performance when built as per the AHA recommended filtering (0.5 to 150 Hz) and Fig. 4b shows the output when the circuit was modified for a narrower bandwidth (5 to 30 Hz). This change was achieved by modifying the filter components (C4, R8, C9, C10 and R11).

This test was repeated using a three-electrode arrangement, where the third electrode was the driven reference (E_REF), connected to the central anterior area of a subject's abdomen (near the "belly button"). The ECG waveforms observed (not shown here) were similar to those with the two-electrode configuration.

IV. Discussion

A. *Noise*

The noise observed in the ECG of a test subject at rest was predominantly 60 Hz and, as expected, was greatly reduced when the cutoff frequency of the low-pass filter was changed to 30 Hz. Such bandwidth limiting may be desirable in order to reduce the dominant 60 Hz noise and aid in the extraction of QRS complexes. However, filtering within the bandwidth of the ECG signal will have an effect on its morphology and therefore need to be considered when deciding on an optimal filtering strategy. In practice such in-band filtering could be done through digital processing, with more flexibility in design and less phase distortion.

Nevertheless, the QRS complexes were clearly discernable even in the wide bandwidth signal. As a result, extraction of the R-wave should be easily achievable, at least when subjects are still. In cases where the subject is moving we can expect some baseline drift due to electrode movement and would likely require further processing to reliably identify the R-waves.

B. *Reference Electrode*

The designed circuit is battery powered and DC-isolated from the subject by blocking capacitors (C1, C2 and C13 in Fig. 2). So long as the circuit (and other parts of the acquisition system) has little or no capacitive coupling between circuit ground and either mains or earth ground, then circuit ground will be floating and follow the body's CM signal. This could offer great benefit, as there would then be little CM induced currents coming in through the inputs. The result is a reduced voltage drop across the electrode-to-skin interface (a potentially significant source of error when using dry electrodes) as well as across the series input elements (C1, C2, R3 and R4).

Since, as previously mentioned, the circuit ground follows the body's CM voltage, there is no (or little) 60 Hz CM signal at the IA inputs. This is the likely explanation as to why we observed little difference when using (versus when not using) the driven reference electrode (E_REF).

C. *Performance Evaluation*

Follow up research will consist of characterizing and testing the proposed ECG amplifier and the dry grip electrode system.

V. CONCLUSIONS

The complex issue of ECG signal acquisition and noise management was studied for an application adding transparent physiological monitoring to a portable health appliance. At the expense of acquiring a higher level of line-frequency interference, ECG integrity can be maintained over longer periods by blocking low frequency noise due to motion artifact. A system is proposed using this approach, with a passive fast-recovery front-end amplifier. Digital processing techniques are used to remove the added line noise.

REFERENCES

[1] W. B. Kannel, C. Kannel, R. S. Paffenbarger Jr and L. A. Cupples, "Heart rate and cardiovascular mortality: the Framingham Study," Am. Heart J., vol. 113, pp. 1489-1494, Jun. 1987.

[2] R. F. Gillum, D. M. Makuc and J. J. Feldman, "Pulse rate, coronary heart disease, and death: the NHANES I Epidemiologic Follow-up Study," Am. Heart J., vol. 121, pp. 172-177, 1991.

[3] Ruha, S. Sallinen and S. Nissila, "A real-time microprocessor QRS detector system with a 1-ms timing accuracy for the measurement of ambulatory HRV," IEEE Transactions on Biomedical Engineering, vol. 44, pp. 159-167, 1997.

[4] D. C. Chan and J. R. Green, "Smart rollator prototype," in Medical Measurements and Applications, 2008. MeMeA 2008. IEEE International Workshop on, 2008, pp. 97-100.

[5] J. C. Huhta and J. G. Webster, "60-Hz interference in electrocardiography," IEEE Transactions on Biomedical Engineering, pp. 91-101, 1973.

[6] M. F. Chimeno and R. Pallas-Areny, "A comprehensive model for power line interference in biopotential measurements," IEEE Trans. Instrum. Meas., vol. 49, no. 3, pp. 535–540, Jun. 2000.

[7] B. Winter and J. G. Webster, "Driven-right-leg circuit design," IEEE Transactions on Biomedical Engineering, pp. 62-66, 1983.

[8] M. J. Burke and D. T. Gleeson, "A micropower dry-electrode ECG preamplifier," IEEE Transactions on Biomedical Engineering, vol. 47, pp. 155-162, 2000.

[9] E. M. Spinelli, R. Pallàs-Areny and M. A. Mayosky, "AC-coupled front-end for biopotential measurements," IEEE Transactions on Biomedical Engineering, vol. 50, pp. 391-395, 2003.

[10] M. J. Burke and C. Assambo, "An improved micro-power pre-amplifier for dry-electrode ECG recording," in Proceedings of the 11th Conference on Proceedings of the 11th WSEAS International Conference on Circuits-Volume 11, 2007, pp. 234-239.

[11] Anonymous, "High Q Notch Filter," National Semiconductor Linear Brief 5, March 1969, at http://www.national.com/ms/LB/LB-5.pdf

[12] S. K. Yoo, N. H. Kim, J. S. Song, T. H. Lee and K. M. Kim, "Simple self-tuned notch filter in a bio-potential amplifier," Medical and Biological Engineering and Computing, vol. 35, pp. 151-154, 1997.

[13] R. Warlar and C. Eswaran, "Integer coefficient bandpass filter for the simultaneous removal of baseline wander, 50 and 100 Hz interference from the ECG," Medical and Biological Engineering and Computing, vol. 29, pp. 333-336, 1991.

[14] W. W. Dawson and H. W. Doddington, "Phase distortion of biological signals: extraction of signal from noise without phase error," Electroencephalogr. Clin. Neurophysiol., vol. 34, pp. 207-211, 1973.

[15] D. Tayler and R. Vincent, "Signal distortion in the electrocardiogram due to inadequate phase response," IEEE Transactions on Biomedical Engineering, pp. 352-356, 1983.

[16] Dobrev and I. Daskalov, "Two-electrode biopotential amplifier with current-driven inputs," Medical and Biological Engineering and Computing, vol. 40, pp. 122-127, 2002.

[17] Dobrev, "Two-electrode low supply voltage electrocardiogram signal amplifier," Medical and Biological Engineering and Computing, vol. 42, pp. 272-276, 2004.

[18] Searle and L. Kirkup, "A direct comparison of wet, dry and insulating bioelectric recording electrodes," Physiol. Meas., vol. 21, pp. 271-284, 2000.

[19] Gruetzmann, S. Hansen and J. Müller, "Novel dry electrodes for ECG monitoring," Physiol. Meas., vol. 28, pp. 1375-1390, 2007.

[20] J. J. Baily, A. S. Berson, A. J. Garson, L. G. Horan, P. W. Macfarlane, D. W. Mortara and C. Zywietz, "Recommendations for standardization and specifications in automated electrocardiography: bandwidth and digital signal processing," Circulation, vol. 81, pp. 730-739, Feb. 1990.

[21] M. J. Burke, "Low-power ECG amplifier/detector for dry-electrode heart rate monitoring," Medical and Biological Engineering and Computing, vol. 32, pp. 678-683, 1994.

[22] J. Pan and W. J. Tompkins, "A Real-Time QRS Detection Algorithm," Biomedical Engineering, IEEE Transactions on, vol. BME-32, pp. 230-236, 1985.

[23] N. V. Thakor and J. G. Webster, "Ground-free ECG recording with two electrodes," IEEE Transactions on Biomedical Engineering, pp. 699-704, 1980.

[24] B. Winter and J. G. Webster, "Reduction of Interference Due to Common Mode Voltage in Biopotential Amplifiers," Biomedical Engineering, IEEE Transactions on, vol. BME-30, pp. 58-62, 1983.

[25] R. Pallas-Areny, "On the Reduction of Interference Due to Common Mode Voltage in Two-Electrode Biopotential Amplifiers," Biomedical Engineering, IEEE Transactions on, vol. BME-33, pp. 1043-1046, 1986.

[26] Anonymous, "CR2032 3V Lithium Battery Technical Data Sheet," Renata Batteries, Rev. CR2032.06/12.06, at http://www.renata.com/pdf/3vlithium/CR2032_v06.pdf