34th Annual International Conference of the IEEE EMBS
San Diego, California USA, 28 August - 1 September, 2012

Noncontact ECG System for Unobtrusive Long-Term Monitoring

Neil J. McDonald, Harini A. Anumula, Eric Duff, and Walid Soussou

Abstract— **This paper describes measurements made using an ECG system with QUASAR's capacitive bioelectrodes integrated into a pad system that is placed over a chair. QUASAR's capacitive bioelectrode has the property of measuring bioelectric potentials at a small separation from the body. This enables the measurement of ECG signals through fabric, without the removal of clothing or preparation of skin. The ECG was measured through the subject's clothing while the subject sat in the chair without any supporting action from the subject. The ECG pad system is an example of a high compliance system that places minimal requirements upon the subject and, consequently, can be used to generate a long-term record from ECG segments collected on a daily basis, providing valuable information on long-term trends in cardiac health.**

I. Introduction

Heart disease is a major cause of disability and is the principal cause of mortality in the US, responsible for 700,000 deaths in 2002 [1]. In the failing heart, a number of arrhythmias and changes in the electrocardiogram (ECG) can be observed due to underlying structural heart disease, electrical instability or the influence of increased sympathetic tone. However, due to the asymptomatic or transient nature of many arrhythmias, there are significant limitations to the diagnostic accuracy and quantification of arrhythmia burden in all of the available monitoring techniques employed.

Holter monitoring generally addresses the management of arrhythmias and testing the efficacy of antiarrhythmic therapies and is typically used for a period of 24 to 48 hours. However, if the arrhythmia is intermittent, the frequency and duration of episodes as recorded by one Holter may not be typical of other days or an average of many days or weeks. Electrocardiographic parameters derived from Holter recordings have been shown to be independent risk predictors of total mortality and the progression of heart failure [2].

Longer term monitoring using event recorders and implantable loop recorders, which use automated algorithms to detect atypical features in the ECG, have shown that asymptomatic clinically significant arrhythmias can be detected, where the ability to confirm the diagnosis of the arrhythmias is comparable to patient-activated looping event monitors [3], [4]. Furthermore, subclinical atrial tachyarrhythmias detected within the first 3 months after implantation of a pacemaker or defibrillator have been associated with a significantly increased risk of ischemic stroke or systemic embolism [5].

This paper describes an ECG pad system that can be positioned on a chair or placed on a bed. Ideally, a (non-implantable) ECG system for long-term monitoring would require no intervention or attention from the patient; the patient's ECG would simply be recorded without any significant change to the user's normal habits or environment. In a chair modality, the ECG is measured through the subject's clothing while the subject seated in the chair without any supporting action from the subject.

The ECG pad system described in this paper is based on QUASAR's noninvasive bioelectrode sensor technology, which is capable of through-clothing measurements of ECG [6], [7]. No modification of the skin's outer layer is required. This represents a considerable advantage with respect to conventional electrode technology, which requires the use of conductive pastes or gels, often with abrasive skin preparation of the electrode site. The biosensor technologies are comfortable, easy to use and can be incorporated into items worn by the subject [6], [8].

ECG chair systems using capacitive sensors have been developed independently by groups in Seoul [9] and Aachen [10]. However, the bandwidth of the sensing electronics in these systems has a low corner frequency limited to > 1 Hz. Therefore both systems are capable of determining rate from the QRS complex of the ECG waveform, but otherwise have limited clinical utility. In contrast, the sensors in QUASAR's ECG pad system are designed to possess a low corner frequency less than 50 mHz.

The ECG pad system is an example of a high compliance system that places minimal requirements upon the subject, and thus avoids the risk that the subject will not set up the system correctly, or forget to use it. Consequently, the pad system could be used to generate a long-term record from ECG segments that are collected on a daily basis, providing valuable information on long-term trends in cardiac health.

II. Hardware

A. QUASAR Capacitive Biosensors

Detection of human bioelectric signals using purely capacitive coupling was first developed in 1967 [11] and patented in 1970 [12]. To accommodate the current noise of available transistors, an electrode capacitance of order 5 nF was utilized. This was achieved by pressing an electrode with a thin (~20 μm) insulating layer of high dielectric constant against the skin. Nevertheless, these electrodes were

*Research supported by National Heart, Lung, and Blood Institute.

W. Soussou is Vice President of Product Engineering of QUASAR, San Diego, CA 92121 USA (e-mail: wsoussou@quasarusa.com).

N.J. McDonald is with QUASAR, San Diego, CA 92121 USA (e-mail: neil@quasarusa.com).

H.A. Anumula is with QUASAR, San Diego, CA 92121 USA (e-mail: harini@quasarusa.com).

E. Duff is with QUASAR, San Diego, CA 92121 USA (e-mail: eduff@quasarusa.com).

subject to lift-off artifact that caused changes in the electrode-body coupling capacitance.

Transistor technology has improved dramatically since the late 1960s and state-of-the-art (SoA) devices now possess the combination of high input impedance, low current noise and low input bias current that enables the use of capacitive electrodes mounted off the body, such that their capacitance is of order 1 pF. Measurements of ECG at 5 cm stand-off from the body were published in 2002 [13].

QUASAR's capacitive biosensors (Fig. 1) can tolerate very small capacitances to the source, which enables the new electrodes to be operated at a standoff from the skin (up to several millimeters in practice). This is due to the exceptionally high input impedance, which is achieved using a combination of novel stabilization and feedback circuits developed by QUASAR [14]. These biosensors possess sufficient signal fidelity for operational recordings of physiologic signals, and are in fact capable of measuring bioelectric signals through several layers of fabric [15]. The low input capacitance (< 0.1 pF) provides a high degree of matching between sensors, which typically have a coupling capacitance of order 10pF, even through a layer of cotton. (Note that the coupling through clothing tends to be resistive [16]. The resistance is typically smaller than the impedance of the dielectric insulation on the capacitive sensors.)

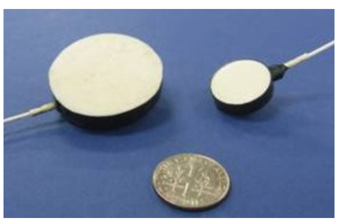

Figure 1. QUASAR capacitive sensors (with US 10c coin for scale). (left) Sensor used in ECG pad system. (right) Sensor from ECG belt systems.

B. *Common-Mode Follower (CMF) Technology*

QUASAR's high impedance sensors are used in combination with QUASAR's proprietary CMF technology. The CMF is a separate biosensor that is used to reduce the sensitivity of the biosensors to common mode signals on the body [8]. The output of the CMF is used as a reference for bioelectric measurements by QUASAR biosensors to dynamically remove the common-mode signal appearing on the body from the measurement. This typically achieves a common-mode rejection ratio (CMRR) of 50 to 70 dB.

C. *ECG Pad*

The ECG pad (Fig. 2) was originally built by QUASAR to explore cardiac monitoring in automobiles. The black regions in the Figure are Velcro™ patches to hold the sensors in place, and are intended to approximate the Right Arm (RA), and Left Leg (LL) electrode locations. Not shown is a third patch covering a region in the upper right to approximate the Left Arm (LA) electrode location.

The gold-colored fabric is a medical grade conductive fabric that acts as ground for the system. Not shown in the photograph is the cotton material that is placed over the pad to hide the electrodes and conductive fabric. The system can operate without direct contact to the subject's skin.

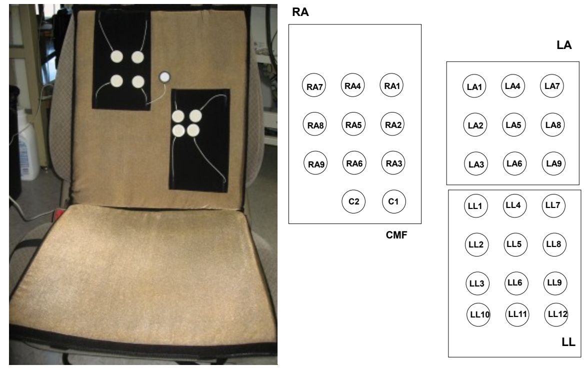

Figure 2. (left) ECG pad system for long-term monitoring of ECG. (right) Array of electrode positions on the RA, LA and LL Velcro^{TM} patches.

An example of an ECG measurement recorded during a routine demonstration of the ECG pad system is presented in Fig. 3. It revealed that the person sitting in the chair experienced frequent asymptomatic episodes of premature ventricular contractions (PVCs). The onset of one such episode is shown in Fig. 3. The subject had no previously diagnosed heart condition; no serious condition was diagnosed when the subject subsequently consulted a cardiologist.

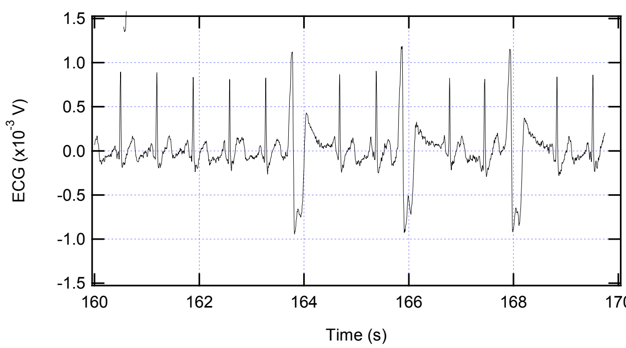

Figure 3. Asymptomatic PVCs detected using ECG pad system. Lead-II data sampled at 240 sps, filtered 0.5-50 Hz (-3dB).

It should be noted that throughout this paper the standard terminology for limb leads (Lead-I, Lead-II, Lead-III, Lead-aVL, Lead-aVR, Lead-aVF) will be used to refer to the ECG limb lead signals derived from the ECG pad electrodes. The capacitive sensors in the pad, however, are not in standard limb electrode locations. Therefore the signal morphology in each limb lead for the ECG pad can be expected to vary from that observed using electrodes in the standard electrode locations.

C. *Data Acquisition*

Data acquisition was performed using QUASAR's BioNode H-12 data acquisition hardware [8]. It uses 16-bit Σ-Δ analog-to digital converters to simultaneously acquire up to 12 channels of physiologic data. The analog electronics were designed to have a high CMRR between channels for physiologic signals between 0.05 Hz and 100 Hz. The bandwidth for the measurement is the Nyquist frequency.

The BioNode H-12 possesses a short range wireless transceiver that forms a Personal Area Network (PAN) with a custom Base Station. Communication is via a Gaussian Frequency Shift Keying (GFSK) protocol that has the ability to use 125 channels of the RF range between 2.400 GHz and 2.525 GHz of the Worldwide ISM band.

The data presented in the Results section was acquired using a sample rate of 960 sps.

III. METHODS

The inputs of the BioNode H-12 were configured to measure the signals from 8 capacitive sensors (CMF-referred) and 3 conventional ECG pre-gelled electrodes (referred to a fourth electrode). The last input was reserved for the CMF, referred to the ground of the ECG pad.

The electrode locations indicated in Fig. 2 were evaluated using the RMS noise of data after removing the averaged ECG waveform recorded at each electrode site. The sensor locations with lowest residual RMS noise, averaged across all subjects, were: LA (1, 4, 5), RA (1, 3, 5) and LL (1, 4, 5). For the data presented in this paper, LA 1 was omitted to reduce the number of pad sensors to 8.

Pre-gelled ECG electrodes (Softy Trode #5106) were positioned just below the clavicle (LA, RA) and on the lower torso just above the belt line (LL). The reference electrode was placed just above the belt on the right side of the torso. Wet electrode sites were prepared via abrasion with NuPrep, followed by cleaning with an ethanol wipe. No preparation of the capacitive electrode sites was performed.

ECG measurements were conducted in an electromagnetically shielded room. The subject sat down, leaning comfortably into the chair. Data were acquired with the subject sitting still, and with the subject performing the following movements: alternating lifting left leg/right leg from hip, alternating lifting left leg/right leg from knee, moving right arm to pick up and put down a glass of water, and moving left arm to pick up and put down a bottle of water. The tempos of the arm and leg motions were dictated by a metronome set to 0.06 Hz and 0.1 Hz, respectively.

Triboelectric charging of the electrode's sensing surface can be a major source of artifact with the small capacitances present in capacitive sensors. The influence of triboelectric charging was evaluated by measuring the ECG while the subject wore T-shirts made from different materials (100% cotton, cotton/polyester blend, 100% polyester). Each task was performed at room temperature and ambient relative humidity (76-78 F, 43-50% RH), and repeated at elevated temperature/relative humidity (79-90 F, 76-83% RH).

IV. ANALYSIS

The digitized signals were filtered using a 0.1-100 Hz 8-pole Bessel band pass filter (-3dB). The ECG limb and augmented limb leads were calculated for both the capacitive and wet electrodes. The selection of LA, RA and LL electrodes for calculation of limb leads was determined using the minimum RMS voltage after the average ECG waveform had been subtracted.

The 6 ECG limb lead signals for the capacitive electrodes were printed in 10 second epochs for scoring by a cardiologist (horizontal scale 25 mm/second, vertical scale 20 mm/mV). A total of 1872 epochs were scored corresponding to 39 epochs per subject for each shirt at a given temperature and relative humidity (9 sitting still, 6 moving right arm, 6 moving left arm, 9 lifting left leg/right leg from knee, 9 lifting left leg/right leg from hip). Data were scored by a cardiologist using the scale in Table I.

TABLE I. SCALE FOR SCORING ECG DATA

	ECG Features
6	Noise and motion artifact levels are low enough to realize **ST-segment elevation**, in addition to (5), with a high level of confidence.
5	Noise and motion artifact levels are low enough to realize **identification of QT interval**, in addition to (4), with a high level of confidence.
4	Signal quality is such that it is possible to determine **detection of P-waves and their unambiguous association with QRS complex** and **determination of rhythm from QRS configuration and P-wave association with QRS complex**, in addition to (3). Noise and motion artifact is such that it is no longer possible to determine QT intervals.
3	Signal quality is such that it is possible to determine **identification of QRS configuration and QRS duration**, in addition to (2). It is no longer possible to determine rhythm because noise and motion artifact are large enough to obscure the presence of P-waves.
2	Signal quality is such that it is possible to determine **identification of T-waves and their association with QRS complex**, in addition to (1). Noise and motion artifact obscure features necessary for determination of QRS configuration and duration.
1	Signal quality is such that it is possible to determine only **representation of rate from QRS complex**. This is the minimum level of signal quality that provides useful information.
0	Noise and/or artifact prevent clear identification of QRS complex. Unable to identify rate.

The accuracy of heart rate determination was calculated using the timing of the peaks of QRS complexes in the ECG pad data, as compared to the timing for wet electrode data. The wet Lead-II signal was used as the reference channel. The time coordinates of the QRS complex in each lead were compared with the coordinate of the QRS complex in the reference channel. The timing accuracy was calculated as the standard deviation in the difference $t_{QRS,\text{limb lead}} - t_{QRS,\text{Lead-II(wet)}}$, denoted as $\sigma_{\Delta t}$.

V. RESULTS

A total of 9 subjects were enrolled for these tests. Signed informed consent was obtained. One subject was unavailable for the ECG measurements. Two subjects only completed the measurements at room temperature and ambient humidity.

The data in Fig. 4 is typical of sitting still data for a subject wearing a cotton shirt. In this example, the signal quality was scored as 6. Artifacts for subjects sitting still and wearing a cotton shirt were not significantly higher at lower humidity. Signals from the capacitive sensors show different ECG signal morphologies and lower amplitudes for the QRS complex relative to the wet electrodes due to the fact that the capacitive sensors are (a) not coincident with the wet electrodes, (b) sensing on the back, and (c) using a reduced vector across the heart than is normal for limb leads. This is most evident in the Lead-III and Lead-aVL signals, which have lower amplitude and different polarity of the QRS complex, respectively, relative to their wet electrode equivalents. Both can be attributed to the reduced vertical spacing between the chair's LA and LL electrodes relative to the wet electrodes.

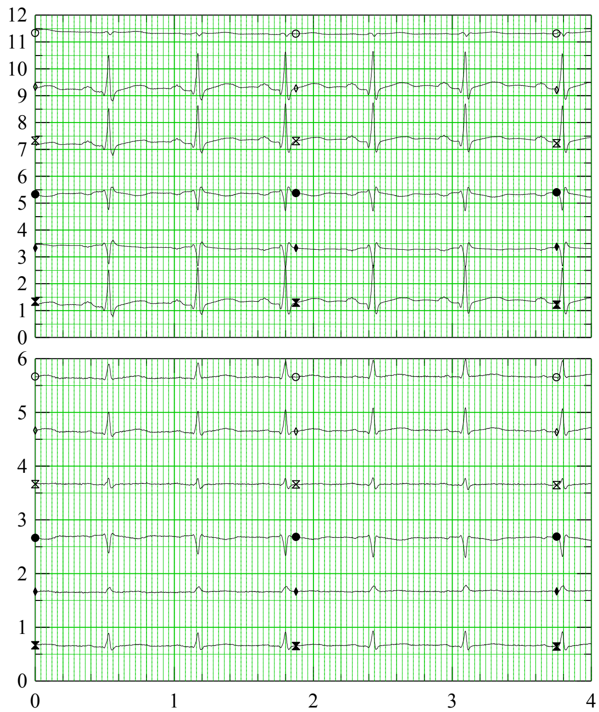

Figure 4. Subject 7 – Subject sitting still, 100% cotton shirt, 86 F, 82% RH. Wet electrode data (upper) & capacitive electrode data (lower) filtered 0.1-100 Hz (-3dB). Lead-I (—○—), Lead-II (—♦—), Lead-III (—⧗—), Lead-aVR (—●—), Lead-aVL (—♦—), Lead-aVF (—⧗—).

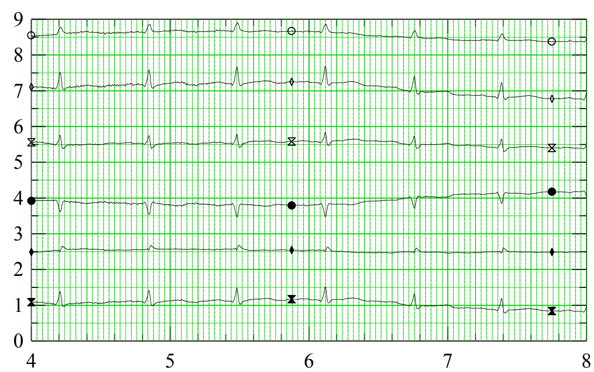

Figure 5. Subject 7 – Subject lifting leg from hip, 100% cotton shirt, 88 F, 81% RH. Capacitive electrode data filtered 0.1-100 Hz (-3dB). Lead-I (—○—), Lead-II (—♦—), Lead-III (—⧗—), Lead-aVR (—●—), Lead-aVL (—♦—), Lead-aVF (—⧗—).

In the case of lifting leg (from hip), the data in Fig. 5 demonstrate the quality of signal that can be realized during subject motion; in this example the signal quality was scored as 6. The lifting motion mimics the motion of the subject if they were to cross their legs, or adjust their position in the chair. It is possible for this motion to cause artifact, partly due to skin-stretch artifact resulting from changes in pressure as the subject presses back into the chair [17], [18]; this is likely cause of the baseline drift evident in Fig. 5. A second contribution can be triboelectric charging due to shifting in the chair, caused by either a momentary loss of contact with the sensors, or rubbing the shirt material against dielectric on the surface of the sensors. Triboelectric charging typically results in considerably larger artifacts that can render the data useless (corresponding to a score of 0 in Table I).

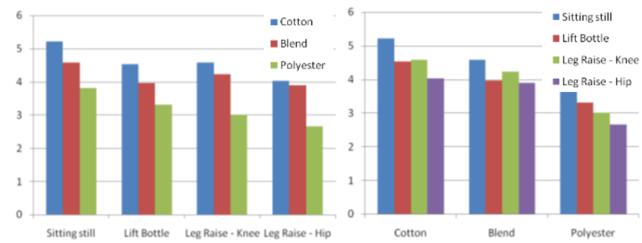

Figure 6. ECG scores for low relative humidity averaged across all subjects as a function of: (left) activity, and (right) shirt material.

Results from ECG scoring of low humidity data by a cardiologist, according to the scale in Table I, show a trend of reduced data quality during subject motion, with leg raises from the hip possessing the worst ECG scores (Fig. 6). The data also indicate that the 100% cotton material possesses the best ECG scores, with 100% polyester possessing the worst scores. This is in agreement with our expectation that synthetic materials would be subject to greater triboelectric charging. A Friedman non-parametric two-way analysis of variance across the different shirt materials revealed a statistically significant difference between the groups ($p<0.008$). However, post-hoc multiple comparisons indicated that no two groups were significantly different from one another, suggesting a strong trend in impact of the fabrics, but none that are particularly better or worse.

TABLE II. TIMING ACCURACY AVERAGED ACROSS ALL SUBJECTS

Limb Lead (wet)	$\sigma_{\Delta t}$ (ms)	Limb Lead (pad)	$\sigma_{\Delta t}$ (ms)
Lead-I	0.52	Lead-I	0.67
Lead-II	N/A	Lead-II	0.35
Lead-III	0.28	Lead-III	1.86
Lead-aVR	0.08	Lead-aVR	0.41
Lead-aVL	0.44	Lead-aVL	0.42
Lead-aVF	0.14	Lead-aVF	3.92

Results for the timing of the QRS peaks demonstrate that the determination of heart rate using the pad system possesses timing accuracy < 1 ms, which is comparable to that obtained using wet electrodes. Table II presents the standard deviation, $\sigma_{\Delta t}$, of the timing difference for QRS peaks for each sitting still data set, averaged across all subjects, fabric and conditions. For the pad data, the two limb leads that have $\sigma_{\Delta t}$ greater than the sample interval (>1ms) are Lead-III and Lead-aVF, which are consistently observed to have the lowest QRS complex amplitudes. All other leads possess an average $\sigma_{\Delta t}$ that is less than the sample interval.

VI. Discussion

Due to the non-standard electrode locations in the chair ECG system described above, the signal morphology is noticeably different in nominally equivalent leads, and QRS amplitudes are substantially reduced. It is likely that the principal issue with implementing an ECG pad system lies in interpreting the change in ECG waveform morphology and signal amplitude that results from the non-standard electrode positions. Some cardiac conditions are indicated by changes on a particular lead; new interpretations will be required for correct identification of these conditions. Ultimately, this should not detract from the utility of the system, as some features of the ECG waveform (presence of P-waves, width of QRS complex) will not be affected, and so interpretation will be straightforward.

The ECG chair represents a new paradigm for long-term cardiac monitoring, in which the ECG can be monitored daily over months or years, and may lead to revolutionary at-home monitoring approaches. Patients may be monitored for the early detection of cardiac disease or tracking the progression of heart disease in sick patients. Alternatively, the evaluation of dynamic parameters, such as heart rate variability/turbulence (HRV/HRT) and the rate-corrected QT interval (QT_C), and their evolution over extended periods, may offer new insights and understanding into cardiac health and pharmaceutical effects.

Long-term monitoring is limited by current technology due to its invasiveness and lack of user compliance. The new possibilities afforded by prolonged ECG monitoring in home settings are yet to be quantified or even assessed at a preliminary level owing to the prior lack of suitable technology. In general, the ECG pad described herein supports the present trend towards in-home monitoring of patients, with the associated reduced burden upon health resources. A strategy of primary monitoring is more cost effective than conventional testing for cardiac patients [19], and can deliver timely and more effective preventative care in addition to clear cost benefits [20].

Acknowledgment

The project described was supported by Award Number R43HL097440 from the National Heart, Lung, and Blood Institute. The content is solely the responsibility of the authors and does not necessarily represent the views of the National Heart, Lung and Blood Institute or the National Institutes of Health.

References

[1] CDC, "Deaths: Leading Causes for 2002," *National Vital Statistics Reports*, 53(17), 2005.

[2] I. Cygankiewicz, W. Zareba, and A.B. de Luna, "Prognostic value of Holter monitoring in congestive heart failure," *Cardiology Journal*, vol. 15, pp. 313-323, 2008.

[3] J.A. Olson, A.M. Fouts, B.J. Padanilam, and E.N. Prystowsky, "Utility of Mobile Cardiac Outpatient Telemetry for the Diagnosis of Palpitations, Presyncope, Syncope, and the Assessment of Therapy Efficacy," *J. Cardiovasc. Electrophysiol.*, vol. 18, pp. 472-477, 2007.

[4] S.A. Rothman, J.C. Laughlin, J. Seltzer, J.S. Walia, R.I. Baman, S.Y. Sioffi, R.M. Sangrigoli, and P.R. Kowey, "The diagnosis of cardiac arrhythmias: a prospective multi-center randomized study comparing mobile cardiac outpatient telemetery versus standard loop event monitoring," *J. Cardiovasc. Electrophysiol.*, vol. 18, pp. 241-247, 2007.

[5] J.S. Healey, S.J. Connolly, M.R. Gold, C. W. Israel, I.C. Van Gelder, A. Capucci, C.P. Lau, E. Fain, S. Yang, C. Bailleul, C.A. Morillo, M. Carlson, E. Themeles, E.S. Kaufman, and S.H. Hohnloser, "Subclinical Atrial Fibrillation and the Risk of Stroke," *The New England Journal of Medicine*, vol. 18, pp. 120-129, 2012.

[6] R. Matthews, N. J. McDonald, I. Fridman, P. Hervieux, and T. Nielsen, "The invisible electrode – zero prep time, ultra low capacitive sensing," presented at the 11th International Conference on Human Computer Interaction (HCII), Las Vegas, NV, July 22-27, 2005.

[7] R. Mathews, N. J. McDonald, H. Anumula, L. J. Trejo, "Novel Hybrid Sensors for Unobtrusive Recording of Human Biopotentials," in *Foundations of Augmented Cognition,* 2nd ed., San Ramon, 2006, pp. 91-101, 2006.

[8] R. Matthews, N.J. McDonald, P. Hervieux, P.J. Turner, M.A. Steindorf, "A Wearable Physiological Sensor Suite for Unobtrusive Monitoring of Physiological and Cognitive State," in *Engineering in Medicine and Biology Society, 2007. EMBC 2007*, Proceedings of the 29th Annual International Conference of the IEEE, 22-26 August, 2007, pp. 5276-5281, 2007.

[9] Y.G. Lim, K.K. Kim, and K.S. Park, "ECG Measurement on a Chair Without Conductive Contact," *IEEE Trans. Biomed. Engin.,* vol. 53, pp. 956-959, 2006.

[10] M. Walter, B. Eilebrecht, T. Wartzek, and S. Leonhardt, "The smart car seat: personalized monitoring of vital signs in automotive applications," *Pers. Ubiquit. Comput.,* vol. 15, pp. 707-715, 2011.

[11] P. C. Richardson, "The insulated electrode: A pasteless electrocardiographic technique," *Proc. Annu. Conf. Eng. Med. Biol.,* vol. 7, pp. 9-15, 1968.

[12] P.C. Richardson, and A. Lopez, Jr., "Electrocardiographic and bioelectric capacitive electrode," U.S.Patent 3,500,823, March 17, 1970.

[13] C.J. Harland, T.D. Clark, and R.J. Prance, "Electric potential probes – new directions in the remote sensing of the human body", *Meas. Sci. Technol.*, vol. 13, pp. 163-169, 2002.

[14] M.A. Krupka, A. Hibbs, "Low Noise, Electric Field Sensor." US patent 6,686,800, February 3, 2004.

[15] R. Matthews, M. A. Krupka, C. Say, and A. D. Hibbs, "Capacitively coupled non-contact sensors for measurement of human bioelectric signals," presented at Advanced Technology Applications for Combat Casualty Care (ATACCC), Ft. Walton Beach, FL, September 9-14, 2001.

[16] Y.M. Chi, T-P. Jung, and G. Cauwenberghs, "Dry-Contract and Noncontact Biopotential Electrodes: Methodological Review," *IEEE Rev. Biomded. Eng.*, vol. 3, pp. 106-119, 2010.

[17] N.V. Thakor, and J.G. Webster, "The Origin of Skin Potential and its Variations", *Proc. Ann. Conf. Eng. Biol. Med.*, vol. 20, pp. 212, 1978

[18] H. de Talhouet, and J.G. Webster, "Origin of Skin-Stretch-Caused Motion Artifacts under Electrodes", *Physiol. Meas.*, vol. 17, pp. 81-93, 1996.

[19] A.D. Krahn, G.J. Klein, R. Yee, J.S. Hoch, and A.C. Skanes, "Cost implications of testing strategy in patients with syncope: Randomized assessment of syncope trial," *J. Am. Coll. Cardiol.*, vol. 42, pp. 495-501, 2003.

[20] M. Paynter, "Delivering expert cardiac support in the community," *Br. J. Community Nurs.*, vol. 12, pp. 460-462, 2007.