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Low Impedance Carbon Adhesive Electrodes with Long Shelf Life

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Abstract—A novel electrocardiogram (ECG) electrode film is developed by mixing carbon black powder and a quaternary salt with a visco-elastic polymeric adhesive. Unlike traditional wet gel-based electrodes, carbon/salt/adhesive (CSA) electrodes should theoretically have an infinite shelf life as they do not dehydrate even after a prolonged period of storage. The CSA electrodes are electrically activated for use through the process of electrophoresis. Specifically, the activation procedure involves sending a high voltage and current through the electrode, which results in significant reduction of impedance so that high fidelity ECG signals can be obtained. Using the activation procedure, the ideal concentration of carbon black powder in the mixture with the adhesive was examined. It was determined that the optimum concentration of carbon black which minimized post-activation impedance was 10%. Once the optimal carbon black powder concentration was determined, extensive signal analysis was performed to compare the performance of the CSA electrodes to the standard silver–silver chloride (Ag/AgCl) electrodes. As a part of data analysis, electrode–skin contact impedance of the CSA was measured and compared to the standard Ag/AgCl electrodes; we found consistently lower impedance for CSA electrodes. For quantitative data analysis, we simultaneously collected ECG data with CSA and Ag/AgCl electrodes from 17 healthy subjects. Heart rate variability (HRV) indices and ECG morphological waveforms were calculated to compare CSA and Ag/AgCl electrodes. Non-significant differences for most of the HRV indices between CSA and Ag/AgCl electrodes were found. Of the morphological waveform metrics consisting of R-wave peak amplitude, ST-segment elevation and QT interval, only the first index was found to be significantly different between the two media. The response of CSA electrodes to motion artifacts was also tested, and we found in general no difference in the quality of the ECG signal between the two media. Hence, given that CSA electrodes are expected to have a very long shelf-life, with potentially less cost associated with their fabrication, and have ECG signal dynamics nearly identical to those of Ag/AgCl, the new electrodes provide an attractive alternative for ECG measurements.

Keywords—Carbon, Visco-elastic polymeric adhesive, Electrodes, Silver chloride, ECG, Electrode–skin contact impedance, Heart rate variability.

INTRODUCTION

The electrocardiogram (ECG) provides a graphical representation of the electrical activity of the heart, and to properly monitor the cardiac health of a patient, it is crucial to obtain an ECG signal with a high signal-to-noise ratio (SNR). In a typical ECG setup, depending on the particular application, three to twelve signal receptive electrodes are attached to the patient's body. These electrodes are able to measure minute changes in potential that occur as a consequence of the propagation of electrical signal from atria to ventricles during each heartbeat cycle, thus rendering it possible to produce the characteristic ECG waveforms that can then be used for diagnostic purposes. The ECG can also be further quantified by computing the heart rate variability (HRV) indices, as these have been shown to provide a good estimate of the dynamics of the autonomic nervous system.[8] To ensure optimal signal strength of the electrode, it is imperative to minimize the impedance of the electrode–skin interface by lowering the impedance of the electrode. High impedance levels at the skin–electrode interface can result in significant loss in signal strength and low SNR, which will lead to poor quality ECG waveforms.

The current industry gold standard for electrodes is the Silver/Silver Chloride (Ag/AgCl) wet (hydrogel) electrode, which has been proven to provide accurate ECG waveforms.[17] These electrodes consist of a layer of silver chloride, often in the form of a paste-like

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hydrogel surrounding a silver disc. While the hydrogel layer significantly improves the signal quality by effectively lowering the impedance that exists at the electrode–skin interface, it is also the principal reason behind the relatively short shelf life of these electrodes. The hydrogel layer that exists in the skin–electrode interface degrades with time as it dehydrates, producing high impedance. This leads to a loss of signal quality[14] and an increased incidence of motion artifacts and noise in the ECG.[7] Moreover, the electrodes need to be carefully packaged to ensure prolonged retention of the hydrogel layer. Due to finite shelf life and electrode dehydration, these electrodes can only be used to record signals for a few days at maximum.[9] Also, Ag/AgCl electrodes are limited to short-term use because they are known to cause irritation to the skin especially after their removal. Moreover, Ag/AgCl electrodes are also relatively expensive since silver is an expensive commodity. For these reasons, dry electrodes have been developed in recent years to substitute traditional wet Ag/AgCl electrodes.[3,7,13]

A novel type of electrode that does not require a hydrogel layer has been developed to address the issue of dehydration with the current industry gold standard electrodes. The new dry electrodes are designed by combining a visco-elastic polymeric adhesive [4,5] with carbon black powder and a quaternary salt. This mixture is potentially much more economical than Ag/AgCl. The objective of this paper is to demonstrate and detail the fabrication process of our new carbon/salt/adhesive (CSA) electrodes and compare their ECG morphologies to the standard Ag/AgCl wet electrodes.

First, a brief description of the fabrication of the CSA electrodes is provided. After the CSA electrodes are fabricated, they need to be activated by a procedure known as electrophoresis.[10] Once activated, the electrodes must meet the Association for the Advancement of Medical Instrumentation (AAMI) requirement for defibrillation overload recovery and impedance. The AAMI states that for electrodes such as hydrogel, the average post-activation impedance must remain equal to or below 2 kΩ, and no single electrode pair can have impedance greater than 3 kΩ after activation.

To ensure that dry CSA electrodes meet the AAMI standard for electrodes, we designed and implemented an activation circuit which is capable of providing various levels of activation voltage and amperage with varying discharge times. This allowed us to obtain the best parameters required for the activation. This procedure was also used to discern the optimal concentration of carbon by measuring pre- and post-activation impedance values.

Once the activation parameters as well as the carbon concentration were determined, several comparisons between CSA and Ag/AgCl electrodes were carried out. First, electrode–skin contact impedance was measured for each type of electrode. Subsequently, simultaneous ECG data acquisition was performed using the two electrode types on human subjects in order to ensure similarities in the ECG waveforms. Finally, HRV analyses were performed with the hypothesis that the CSA will provide statistically similar indices as that of the Ag/AgCl electrodes.

MATERIALS AND METHODS

Carbon/Salt/Adhesive Electrode Fabrication

The base material used is a 0.001″ thick polyester film with a conductive carbon ink coating on one surface which gives approximately 100 Ω per square of measured surface resistivity. Silver/silver chloride coatings (which are commonly used in hydrogel based electrodes to provide electro-chemical stability) are not needed due to the characteristics of the waterless adhesive. Using a polar organo salt is a way to introduce into the adhesive composition a very polar molecule (electro-physically aligns with the applied electric filed) that orientates with a time varying signal, which can be alternating current, ramped, pulsed DC, mixed waveform or ECG. The salt in this case is a quaternary ammonium salt [hydrogenated tallowalkyl (2-ethylhexyl) dimethyl ammonium methylsulfate] that exhibits excellent compatibility with the chosen waterless adhesive by virtue of its organic functionality.

Since the adhesive is not in a very polar solvent (like water) to stabilize its disassociated ions (the solvent here instead being an acrylic polymer), there is little ionic formation and thus it does not contribute much (if any) ionic conductivity. The alignment response of the polar salt molecule to an applied electric field facilitates a displacement current which is a different mechanism than the ionic current arising from the movement of dissociated salt ions. The greater that the concentration of the polar salt is, the greater the number of available parallel current pathways there are, thus leading to lower impedance. By contrast, salts (such as NaCl) in a hydrogel composition break down to $Na+$ and $Cl-$ (hydrolytic stabilized disassociated ions). It is the formation of these ions that is the basis for ionic conductivity in hydrogels but not in the waterless adhesive used here. Thus, there is no similarity in the mechanism of signal detection between the salt in a hydrogel and the salt used in the CSA electrode.

To create electrodes, 30 g total of the dopant and the visco-elastic polymeric adhesive are added to a beaker. The amount of carbon and salt added is dependent on the percent to be used out of 30 g. A magnetic mixer is used for 40 min to stir the adhesive

and dopant. The mixture is spread onto a liner using a 12 mil spreading tool, and then placed into a 160 °F oven for 10 min. The dried adhesive is covered with a 2nd liner and is cut into 4.45 cm wide pieces, the Carbon electrode film is cut into 5.08 cm wide pieces and the adhesive is placed onto the film, leaving a small amount of carbon film exposed. Electrodes were cut to a size of 4.45 cm × 3.81 cm.

A feature for enhancing surface conductivity is then applied. The conductivity enhancing feature (the bridge) is a low impedance electrically conductive material that produces a lower electrode ohm value by connecting in parallel multiple isolated Z direction (out of plane) conductive pathways in the adhesive. This can be described by $1/R_{\text{total}} = (1/R_1) + (1/R_2) + (1/R_3) + \ldots (1/R_n)$ where R represents each isolated Z direction pathway. The bridge is specifically designed to balance electrical, mechanical, and electrode adhesion properties: its conductive loading level provides electrical conductivity, its polymeric content and thinness provide mechanical flexibility, and its large open area pattern minimizes reduction in adhesive bonding.

In general higher salt concentration leads to higher conductance, but it would also increase the risk of eliciting a reaction in subjects' skin. For the purposes of this study, salt concentration was fixed to 25%, given that such concentration produced very low to minimal skin irritation. A representative CSA electrode is shown on the right panel of Fig. 1 and the left panel shows a standard Ag/AgCl ECG electrode.

Carbon Black Concentration

The ideal concentration of carbon powder in CSA electrodes was determined by minimizing post activation impedance. The carbon concentration ranged from 2 to 12% loading levels with an increment of 2% and we measured pre-activation and post-activation impedance using an impedance meter. Initially, for each carbon load level, five different sample electrodes were used. Subsequently, for the three carbon levels with the lowest post-activation impedance, 30 additional samples for each level were further analyzed. Impedance measurements were taken at 10 Hz as reference because this is within the ECG frequency range of interest.

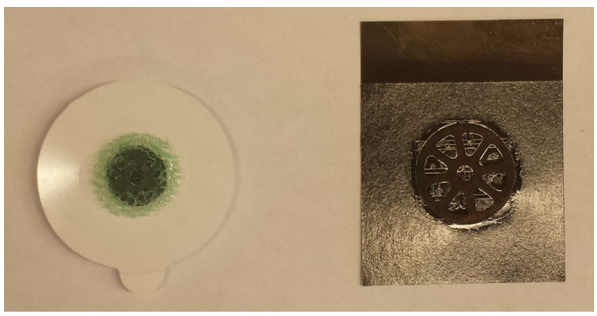

FIGURE 1. Hydrogel Ag/AgCl electrode (left) and carbon/salt/adhesive electrode (right). The circular piece of carbon in the center is the conductivity enhancing feature (bridge).

Activation Process

An activation device with variable parameters was designed and built in order to determine the best parameters to activate the electrodes. The test parameters are listed in Table 1. All permutations of parameter values were tested, in order to determine the optimum activation (i.e., lowest impedance). The design of the activation circuit was determined based on the AAMI standard defibrillation overload circuit. In order to vary parameters as listed in Table 1, dip switches were used to change values of resistors and capacitors. The power supply used provides an AC voltage from 0 to 240 V, but the circuit based on the AAMI standards is typically used with a DC voltage. In order to address this, a diode rectifier was designed at the input to the circuit to convert the voltage from AC to DC.

Activation was obtained by applying electrophoresis through the Z direction of each electrode. Carbon columns formed in the Z-direction were an indication that the activation was successful which was verified in many samples using the scanning electron micrograph (see Fig. 2). A voltage meter was applied across the activation capacitor to examine if the electrode pair was shorted during application of the two electrode adhesives. The electrodes were then removed and two impedance testing clips from a Hiroki IM 3570 were attached to the activation points of the material. The electrode's impedance was tested at 10 Hz by using a 1 V amplitude signal from the impedance testing device. The impedance of each electrode pair was measured before and after activation. In order to meet the Association for the Advancement of Medical Instrumentation (AAMI) standards, the average post-activation impedance is required to be under 2 kΩ with no single value over 3 kΩ.

Electrode–Skin Contact Impedance Measurements

The study protocol was approved by the Institutional Review Board of Worcester Polytechnic Institute and all volunteers consented to be subjects for the experiment.

Activated CSA electrodes were used to carry out over-skin impedance measurements and compare to Ag/AgCl electrodes. The skin of the test subject was cleaned before each measurement by wiping with a 2%-alcohol impregnated cotton pad, which was

TABLE 1. Variable parameters for electrode activation.

Parameter	Values to test	Units
Voltage (DC)	5, 20, 45, 75, 120, 200	V
Current	1, 2, 5, 10, 20, 50, 100	mA
Frequency	DC, 60, 1000	Hz
Time	1, 10, 100, 1000	ms

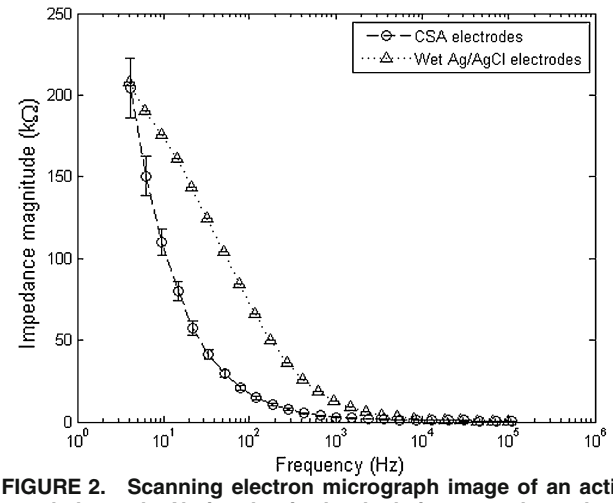

FIGURE 2. Scanning electron micrograph image of an activated electrode. Notice that in the depletion zone the carbon particles moved in from the surrounding area and form vertical columns in the *Z*-direction.

allowed to evaporate before applying the electrodes. Two electrodes were mounted on the right forearm, one on the palm side of the wrist, and the second 5 cm apart from the first but situated towards the elbow. These electrodes were connected to the Hiroki IM3570 impedance analyzer, and each measurement is the result of averaging 20 measurements. The signal voltage amplitude was set to 1 V and the frequency range from 4 to 100 kHz, as used in previous studies.[7,12] $N = 7$ pairs of electrodes were used. To keep skin properties as constant as possible, all measurements were performed in a single day.

Collection of Electrode Comparison Data

A simultaneous set of 3 CSA (post-activation and optimal-mixture) and wet Ag/AgCl electrodes was used to collect ECG signals from 17 healthy subjects (12 males, 5 females, 28 ± 5 years old). An electrode was placed on xiphoid process (ground), a second one in the center of the manubrium (1−), and the third under the left pectoral near V6 electrode position (1+). These placements mimics lead II ECG configuration.[6] Subjects were seated for 7 min followed by standing for the same duration (Rozinn® Holter @ 180 Hz). The ECG signals were band-pass filtered (0.05–40 Hz) offline to reduce noise and motion artifacts. Throughout all the experiments for this study, the skin area was cleaned with 70% isopropyl alcohol prior to each measurement.

HRV Indices

Heart rate variability (HRV) series were obtained from the ECG signals. The R-waveform peaks were detected using a widely-used peak detection algorithm.[15] Time and frequency domain HRV indices were then calculated from the HRV series.

Time Domain Indices

Temporal measures of the HRV considered in this study were: mean RR interval (millisecond units), SDNN (standard deviation of all RR intervals; millisecond units), RMSSD (square root of the mean of the sum of the squares of differences between adjacent RR intervals; millisecond units), NN50 count (number of pairs of adjacent RR intervals differing by more than 50 ms in the entire recording; unitless), and pNN50 (NN50 count divided by the total number of all RR intervals; percentage).

Frequency Domain Indices

The power spectra of HRV were calculated using Welch's periodogram method with 50% data overlap. The RR interval series were converted to an evenly time sampled signal by cubic spline interpolation. A Blackman window was applied to each segment and the Fast Fourier Transform was calculated for each windowed segment. Finally, the power spectra of the segments were averaged. From spectral analysis, two frequency bands were considered: low frequency (LF) band (0.045–0.15 Hz) and high frequency (HF) band (0.15–0.4 Hz).[8] The very low frequency (VLF) band was not taken into account because the physiological correlates are still unknown.[8]

Spectral powers were designated as the Total, LF, HF, LF/HF ratio and are presented as the absolute (square milliseconds, ms^2) units. The spectral contents of the LF and HF are known to reflect dynamics of the autonomic nervous system. Parasympathetic nervous system activity is the major contributor to the HF component.[1,11] The interpretation of the LF component is more difficult, but previous works have demonstrated that it reflects both sympathetic and parasympathetic dynamics.[1,2] Consequently, the LF/HF ratio is considered as a reasonable estimate of the sympatho-vagal balance.[8] The paired *t* test was used to compare all of the obtained parameters.

ECG Morphology Comparison

Out of the group of 17 subjects, 5 were enrolled to collect 5-lead ECG signals in order to perform ECG morphology (lead I, HP 78354A ECG) comparison between CSA and Ag/AgCl electrodes; data were sampled at 1000 Hz for 2 min. With the subjects standing still, either a set of Ag/AgCl or CSA electrodes were placed and ECG signals were collected. Upon immediately after data collection involving a set of electrodes, a second set of electrodes which were not previously used were placed in the same exact position and ECG signals were acquired. The skin area was cleaned with 70% isopropyl alcohol prior to each measurement.

Motion Artifacts

One of the disadvantages of dry electrodes compared to gel electrodes is the lack of flexible contact provided by the gel. This produces variations in contact during motion, and tend to introduce motion artifacts to the ECG signal. We presume that Z-direction formation of carbon columns after activation provides a better reaction to shifting against skin, as the electrodes should be less sensitive to X- and Y-direction potentials. Same 5 subjects enrolled for ECG morphology comparison were used to perform a motion artifacts evaluation (lead I, Chroma® Holter @512 Hz sampling rate). Subjects were requested to perform horizontal movements (HM; torso rotations), vertical movements (VM; squats-like up and down movements), and random walking (RW). Each subjects wore both types of electrodes (Ag/AgCl and CSA) and simultaneous ECG measurements were collected using two Holter monitors.

RESULTS

Optimum Carbon Concentration

Table 2 shows three different levels of carbon powder concentration from a sample size of 35. As shown in Table 2, the most consistent results (e.g., smallest standard deviation values) were obtained for carbon loads at 10 and 12%. According to AAMI standards, the mean post activation impedance is required to be under 2 kΩ and no single electrode pair can have impedance greater than 3 kΩ. Based on this criterion, only the 10 and 12% carbon concentration levels meet the AAMI criterion. Note that 2, 4, and 6% carbon concentration data are not shown because their post-activation impedance values were significantly higher than 2 kΩ.

TABLE 2. Electrode impedance test results.

Carbon load (%)	Pre-activation impedance (MΩ)	Post-activation impedance (Ω)
8	10.5 ± 6.01	1170 ± 1170
10	2.09 ± 2.79	256 ± 51.4
12	3.96 ± 2.39	286 ± 56.5

Values are expressed as mean ± standard deviation.

In order to verify that the data gathered for pre-activation and post-activation impedances are statistically significant, two-tailed t tests were performed using MATLAB. Only the 12 and 10% carbon concentrations were found to be comparable. Given that there was no statistical difference between 10 and 12% carbon powder concentration, we selected the former as the optimal value. Notice that a slightly higher carbon loading of 15% would be comparable to 10 and 12%.

Activation Parameters

Pre-activation and post-activation impedances were recorded for 840 electrodes. Each parameter value shown in Table 1 was permuted with the other parameters to determine the optimal activation of electrodes. Table 3 details the mean and standard deviation values of the pre- and post-activation impedances based on some of the chosen voltage, capacitance, and current parameter values. We only show those data with the lowest post-activation impedance values.

As shown in Table 3, the optimum activation parameters are 200 V, 1 μF, and 100 mA since this choice provides the lowest post-activation impedance. The post-activation values are well under the AAMI requirements of the mean value being no more than 2 kΩ and no single impedance exceeding 3 kΩ.

Electrode–Skin Contact Impedance

Figure 3 shows the comparison of skin impedance between Ag/AgCl and CSA electrodes as a function of varying frequencies. As shown, activated CSA electrodes exhibit much lower electrode–skin contact impedance than Ag/AgCl electrodes for the frequency band of 4–100 kHz.

Note that the measured Ag/AgCl impedance shown in Fig. 3 is in agreement with previously reported studies.[7]

HRV Indices

For quantitative HRV analysis, R-wave peak detection was applied[15] and the resultant RR segments

TABLE 3. Optimum parameters for activation.

Parameters				
Voltage (V)	Capacitance (μF)	Current (mA)	Pre-activation (MΩ)	Post-activation (kΩ)
200	0.1	20	18.83 ± 3.70	3.68 ± 1.87
200	0.1	100	17.17 ± 8.40	6.04 ± 6.67
200	1	10	20.86 ± 2.34	2.41 ± 1.34
120	1	10	7.82 ± 9.83	1.24 ± 0.26
200	1	100	20.71 ± 2.77	0.443 ± 0.189
120	1	100	20.63 ± 3.32	0.829 ± 0.296
200	10	10	21.57 ± 3.22	1.67 ± 0.49
200	10	20	15.64 ± 8.89	1.19 ± 2.16
120	10	20	21.09 ± 1.96	1.05 ± 1.51

Values are expressed as mean ± standard deviation.

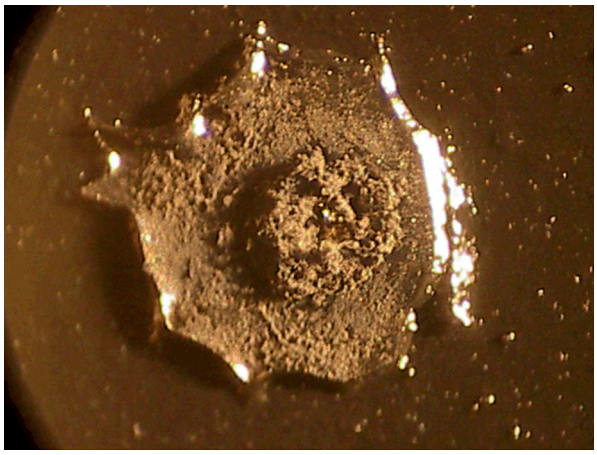

FIGURE 3. Impedance of carbon/salt/adhesive and Ag/AgCl electrodes.

were used to calculate time- and frequency-domain indices were calculated, for sitting and standing positions. A summary of time- and frequency domain indices is presented in Table 4. Figure 4 shows a short duration HR time series obtained by both media.

These indices are similar to those reported by the HRV task force.[8] Non-significant differences were found for all indices except for the mean NN interval index for both body positions. While the mean NN interval index is significantly different, the actual difference is quite small and it is physiologically insignificant.

ECG Morphology Comparison

Figure 5 represents the average of an ensemble template of ECG aligned by the R-wave peak, using the two media in the same body locations (not simultaneous measurement, however), for a given subject. Such ECG ensemble was computed for each subject during sitting and standing positions. Note the similarity in the morphologies of the ECG waveforms between the two media. The R-wave peak amplitude, ST-segment elevation and QT interval duration time were computed for each ensemble and two-tail paired t test was performed to compare between CSA and Ag/AgCl results. Table 5 includes these results. We found only the R-wave peak amplitude of the Ag/AgCl electrodes showed significantly higher value than the CSA electrodes. There was no significant difference in the ST-segment elevation and QT interval time duration between the two electrode media.

Motion Artifacts

Figure 6 shows an example of the data collected during introduction of motion artifacts. Data from both types of electrodes were simultaneously collected using 2 Holter monitors for each subject. Note that motion artifacts affect differently to CSA and Ag/AgCl electrodes. For comparison of motion artifacts on both types of electrodes, the R-wave peak amplitude misdetections were computed. Table 6 shows the percentage of R-wave peak misdetections found for each type of motion. In general, we found that CSA electrodes exhibited a similar performance to the wet Ag/AgCl electrodes in terms of miss-detected R-wave peak.

DISCUSSION

Novel CSA electrodes were developed and their characteristics were compared to the standard Ag/AgCl electrodes. It was found that our dry electrodes are comparable to the industry gold standard Ag/AgCl electrodes in terms of impedance and ECG morphological characteristics. The optimal mixture to fabricate electrodes was determined to be 10% carbon when

TABLE 4. HRV indices.

	Sitting		Standing	
	Carbon/salt/adhesive	Ag/AgCl	Carbon/salt/adhesive	Ag/AgCl
Time domain				
Mean NN (ms)	1186.103 ± 205.9*	1186.11 ± 205.8*	1330.76 ± 238*	1330.77 ± 238*
SDNN (ms)	100.5 ± 62.2	99.9 ± 60.2	107.1 ± 41.3	106.9 ± 40.6
RMSSD (ms)	30.38 ± 48.24	30.003 ± 46.75	26.08 ± 34.14	25.77 ± 33.13
NN50	61.5 ± 126.3	61.5 ± 126.6	41.6 ± 67.6	41.6 ± 67.1
pNN50	0.039 ± 0.077	0.039 ± 0.07	0.025 ± 0.040	0.025 ± 0.040
Frequency domain				
LF (ms^2)	2701 ± 2149	2696 ± 2150	3981 ± 3153	3983 ± 3154
HF (ms^2)	3151 ± 8179	3040 ± 7716	1463 ± 1956	1438 ± 1846
Total (ms^2)	12208 ± 21116	11921 ± 19957	11074 ± 10979	10958 ± 10520
LF/HF	2.80 ± 2.57	2.82 ± 2.62	4.59 ± 3.84	4.61 ± 3.87

Values are expressed as mean ± standard deviation.
* Statistically significant difference.

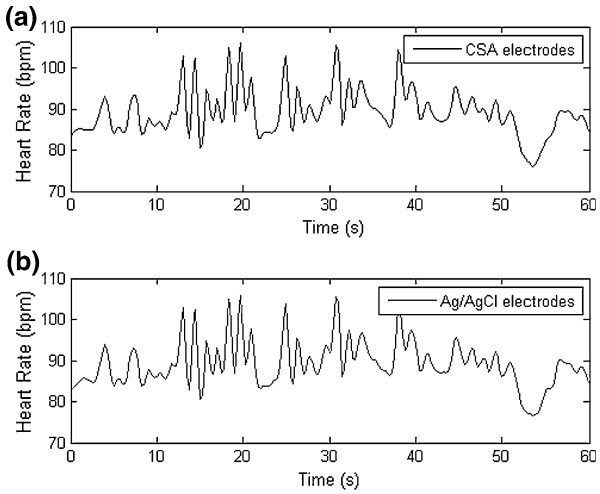

FIGURE 4. Segment of interpolated HR time series signal from ECG signals acquired by using Ag/AgCl (a) and carbon/salt/adhesive electrodes (b).

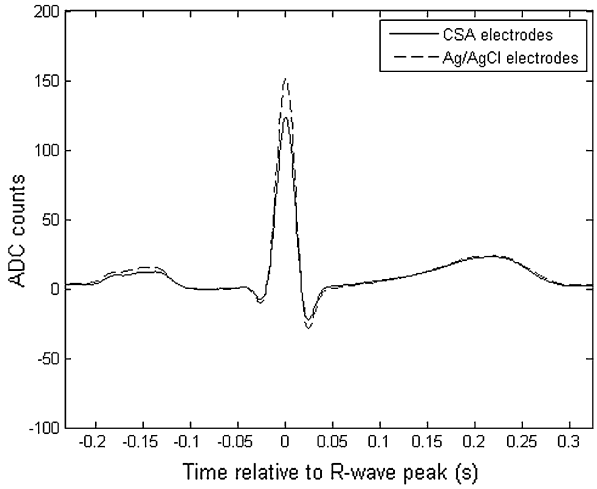

FIGURE 5. Averaged ensembles of ECG signals acquired using Ag/AgCl and carbon/salt/adhesive electrodes, for the same subject.

TABLE 5. ECG morphology measurements over ensemble averages.

	CSA	Ag/AgCl
R-peak amplitude (ADC counts)	151.2 ± 37.6*	165.99 ± 13.06
ST segment elevation (ADC counts)	0.45 ± 6.12	−0.17 ± 4.97
QT time segment (s)	0.24 ± 0.02	0.24 ± 0.018

Values are expressed as mean ± standard deviation.
* Statistically significant difference with respect to Ag/AgCl signals.

combined with 90% visco-elastic polymeric adhesive. This mixture resulted in the lowest post-activation impedance, which also meets AAMI requirements. These electrodes need to be activated prior to use in order to reduce their impedance; pre-activation impedance is in the range of mega ohms. Once the electrodes are activated, they are permanent. The optimal activation parameters were found to be 200 V, 100 mA current, and a 100 ms discharge time using 1 μF capacitor. CSA electrodes address two of Ag/AgCl electrode's issues: dehydration and high cost. The CSA electrodes are theoretically more economical than Ag/AgCl electrodes since carbon is cheaper than silver. Further, the CSA electrodes have an infinite shelf life whereas the Ag/AgCl electrodes dry out and the standard practice is to discard them after 1 month of storage. Finally, our CSA electrodes meet all AAMI requirements.

Electrode–skin impedance measurements showed much lower values for CSA electrodes when compared to wet Ag/AgCl electrodes. Moreover, the impedance of our CSA electrodes is even lower than those

reported for carbon nanotube-polydimethylsiloxane (CNT-PDMS) flexible electrodes.[9]

All HRV indices derived from the CSA electrodes were found to be no different than Ag/AgCl derived indices. Statistically significant differences were found only for the mean NN interval index for both body positions. However, the differences are minimal and not physiologically significant.

ECG morphological waveforms are well captured with CSA electrodes. ST segment elevation and QT segment time showed no significant differences. However, statistically significant differences in R-wave peak amplitudes were found between ECG signals collected using CSA and Ag/AgCl electrodes. Although CSA electrodes exhibited lower impedance than Ag/AgCl, amplitude and morphology of the ECG signals seem to be influenced by capacitive components of the impedance. The elimination of such reduction is an aim for future generations of CSA electrodes. During stable conditions (no movement), such amplitude reduction did not affect the R-wave peak detection (see Table 6).

We examined the quality of both types of electrodes during body movements. Specifically, we examined miss-detection of R-wave peak amplitudes during various body movements consisting of side-to-side and up-and-down torso movements as well as ECG data collection during normal walking condition. In general, CSA electrodes performed comparably to the wet Ag/AgCl electrodes in correct identification of R-wave peaks. During torso rotations, CSA electrodes presented less problems for R-wave peak detection. For vertical movements and random walking, CSA electrodes produced more misdetections than Ag/AgCl electrodes, but such differences were not dramatic. These findings involving CSA show better results with respect to other dry electrodes, which have been reported to have poor performance in comparison to Ag/AgCl in the presence of motion artifacts.[3,7,9,13,16] However, more thorough side-by-side comparison of CSA to other dry electrodes will be needed to make such definitive claim.

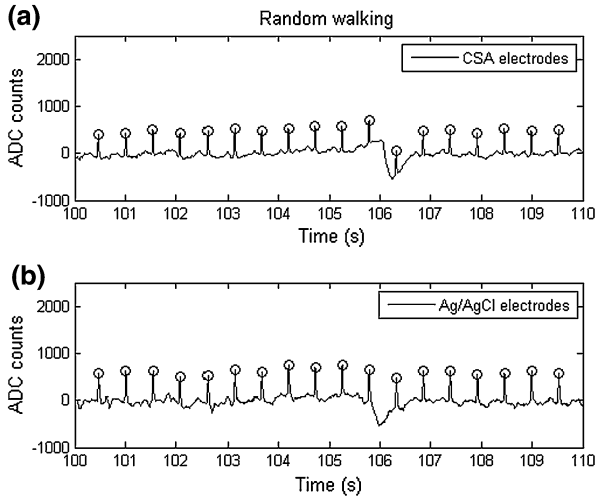

FIGURE 6. Simultaneous ECG signals to show CSA and Ag/AgCl electrodes' motion artifact response.

Based on 30 min of exposure to CSA electrodes, none of the 17 subjects showed any negative reactions to electrode–skin contact. Further, in our previous study involving development of hydrophobic ECG electrodes based on a mixture of carbon-black powder and polydimethylsiloxane (CBPDMS), it was shown that CB/PDMS electrodes are not cytotoxic to L929 connective tissue fibroblasts or neonatal human keratinocytes.[12] Given that our current electrodes are also composed of carbon, we do not expect that they are cytotoxic to skin. Certainly, further experiments will need to be performed to determine the long-term biocompatibility of the CSA electrodes. Moreover, CSA electrodes will need to be tested in the future for their sensitivity to moisture as it is well known that perspiration can lead to degradation of the quality of ECG signals. While we have not tested our electrodes after storing them for more than a year, we have found no ECG signal degradation with 2-month old CSA electrodes (not shown). For future work, we aim to test CSA electrodes after storing them for more than 1 year.

In summary, we demonstrated that dry ECG electrodes can be fabricated from a mixture of carbon powder, salt, and visco-elastic polymeric adhesive. All ECG morphological waveforms and HRV indices were found to be nearly identical to Ag/AgCl electrodes. Moreover, the CSA electrodes can also be applicable for defibrillation usage but the main advantage is their infinite shelf life which lowers both supply chain handling costs and scrap costs when compared to Ag/AgCl electrodes since the latter have a shelf life of only a

TABLE 6. R-wave peak misdetection percentage for CSA and Ag/AgCl electrodes in the presence of motion artifacts.

	No movement $N = 249$ beats		Torso rotation $N = 261$ beats		Vertical movement $N = 267$ beats		Random walking $N = 255$ beats	
	CSA	Ag/AgCl	CSA	Ag/AgCl	CSA	Ag/AgCl	CSA	Ag/AgCl
R-wave peak misdetection %	0%	0%	1%	3.43%	6.67%	3%	9.06%	7.91%

month. Hence, our CSA electrodes have the potential to be a viable and cost-effective alternative to standard Ag/AgCl electrodes.

CONFLICT OF INTEREST

FLEXcon supported the present study and might be interested in commercializing CSA electrodes in the future. FLEXcon employees (KB and JP) may have conflict of interest as they are employed by the company.

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