Five day attachment ECG electrodes for longitudinal bio-sensing using conformal tattoo substrates

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***Abstract*—State-of-the-art ECG (electrocardiography) uses wet Silver/Silver-Chloride (Ag/AgCl) electrodes where a conductive gel is used to provide a resistive, low impedance, connection to the skin. These electrodes are very easy to apply, but have a significant number of limitations for personalized and preventative healthcare. In particular that the gel dries out giving a limited connection time. This paper presents ECG electrodes manufactured using the inkjet printing of Silver nanoparticles onto a conformal tattoo substrate. The substrate maintains a high quality connection to the body for many days at a time allowing ECG monitoring over periods not previously possible without electrode re-attachment. The design and manufacture of the conformal electrodes is presented, together with detailed characterization of the electrode performance in terms of the Signal to Noise Ratio and baseline wander. The Signal to Noise Ratio is shown to still be over 30 dB five days after the initial electrode attachment.**

***Index Terms*—Electrocardiogram (ECG), Conformal sensors, Inkjet printing, Longitudinal bio-sensing.**

I. INTRODUCTION

Conformal electronics, also known as epidermal electronics, are seen as the next step in the rapid advancement of computing power and microelectronic miniaturization [1], [2], see Fig. 1. Unlike current wearable devices, conformal devices use non-permanent *tattoo*-like substrates to connect directly to the skin without requiring a strap or similar. The result is that the devices maintain a high quality connection to the body for many days at a time, and because they follow the contours of the skin they get a much higher contact area. This gives better quality signals and maintains Signal to Noise Ratios (SNRs) even at very small sensor sizes. As such they intrinsically overcome the limitations of current wearable devices, giving better signal quality, longer term connections to the body, and a more discrete profile for better social acceptability.

The challenge now is to enable a wide range of sensing modalities on conformal substrates in order to realize these potential benefits, particularly for personalised and preventative healthcare. It is estimated that 90% of type 2 diabetes, 80% of heart diseases and 70% of strokes could be avoided with the use of suitable preventative techniques [3], and longitudinal bio-sensing is a key enabling technology for this.

Current non-invasive heart monitors for fulfilling this aim are generally based upon Photoplethysmography (PPG) which shines a light into the wrist and measures the amount of light reflected, which changes with blood flow. This is now common in smartwatches such as the Apple Watch [4] and Samsung Simband [5]. However, the light source required necessarily consumes a large amount of power (of the order of 1 mW) limiting the monitoring lifetime due to battery constraints. In addition, the accuracy of current wearable PPG is debated [6], and PPG signals are heavily corrupted by the presence of motion artifacts which mean that only average heart rates (typically averaged over 5–8 s windows) are reported [7]. This prevents the morphological features of the heart beat waveform from being observed and prevents wearable Heart Rate Variability (HRV) analysis. Even at a fixed average heart rate the time between each individual heart beat is not constant, it is modulated by vagal nerve and sympathetic nervous system activity [8]. This gives rise to Heart Rate Variability measures which are direct markers of autonomic activity of significant interest for clinical, non-exercise, situations such as myocardial infarction and diabetic neuropathy [8].

The Electrocardiogram (ECG) is the well known alternative method for monitoring the activity of the heart and is used widely clinically. Small metal electrodes are placed on the chest and used to sense the micro-Volt sized electrical activity from the sino-atrial node and heart muscle contractions that cause the heart pumping action. However, wearable ECG monitoring is much more challenging than wearable PPG due to the electrode contact required with the body, the difficulty in maintaining this robustly over time, and the motion artifacts that are introduced. During such artifacts individual heart beats may be seen (unlike with the PPG), allowing heart rate and heart rate variability analyses, but smaller ECG morphological components such as the *P wave* and *T wave* [9] are obscured.

State-of-the-art ECG uses *wet* Silver/Silver-Chloride (Ag/AgCl) electrodes where a conductive gel is used to provide a resistive, low impedance, connection to the skin. These electrodes are very easy to apply, but have a significant number of limitations for personalized and preventative healthcare:

1) Over time the conductive gel dries out, lowering the quality of the body contact and introducing more noise into the collected signal. This limits the recording time. [10] recommends changing electrodes every day for the best signal quality.

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Fig. 1. The miniaturization of computing power from large desktop PCs in the 1980s to highly portable smart-watches in the 2010s. *Conformal electronics* are envisioned as the next step in this miniaturization process, giving electronic sensors that attach directly to the skin and provide very high quality data over long time periods. Conformal electronics picture taken from [1].

2) Ag/AgCl electrodes are held in place with a strong adhesive, which is painful to remove. Repeated electrode removals, day-after-day, lead to skin reddening, and eventual tearing. In turn this produces user discomfort, the potential for infection, and an inability to record from the same location. As examples, [11], [12] report skin reddening when using Ag/AgCl electrodes for between one and two days.
3) Chest hair can block the electrical connection, and for high quality recordings it is common to shave the whole chest in order to provide a large hair-free region for electrode attachment.
4) The electrodes come in standard sizes and shapes. There is no potential for personalization to the user, for example to help minimise the amount of shaving required.

Conformal ECG electrodes offer the potential to overcome these challenges and to enable long term out of the lab ECG monitoring.

This paper presents the design and performance characterization of electrodes for ECG monitoring manufactured on a conformal tattoo substrate. We use inkjet printing to deposit Silver nanoparticles in order to make the electrode area. As a rapid prototyping, additive manufacturing approach, inkjet printing can allow personalization of electrode sizes and shapes in near real-time. The nanoparticles are deposited on to a conformal tattoo-like substrate which can follow the contours of the skin in order to give a very high quality signal contact and Signal-to-Noise Ratio. In turn, the electrode is very light and the tattoo substrate allows very long term connections to the body to be made with minimal adhesion force, easing removal. Our results demonstrate the use of the new electrodes over a five day period with high SNRs obtained throughout. This is a step change from the typical one day lifetimes obtained by current electrodes and gives substantial potential for new longitudinal bio-sensing. To our knowledge we present the first demonstration of capacitively connected inkjet printed conformal ECG electrodes, the first quantification of the performance of inkjet printed conformal ECG electrodes in terms of SNR and baseline wander, and the first detailed characterisation and quantification of the sensing performance of any form of conformal ECG electrode when used for recording over multiple days.

Section II introduces our electrode manufacturing technique and the mode of operation of the conformal electrodes. Section III evaluates the performance of our prototypes, and this is discussed in Section IV with reference to other recent developments in ECG electrode design. A preliminary version of this work was presented in [13], which only considered the performance of the electrodes in a single application. In the current paper we describe the manufacturing process in full, have improved our performance evaluation procedure (particularly for baseline wander estimation), and characterized the use of the electrodes over periods of up to five days.

II. Conformal ECG electrodes

A. Electrode manufacturing

Our conformal electrodes consist of a printed Silver nanoparticle layer on a conformal substrate with an adhesive film layer placed over the electrode surface to provide a strong attachment to the skin. This gives a capacitively coupled electrode connection, as shown in Fig. 2. Importantly the substrate is conformal to the skin and provides adhesion under the entire electrode surface. This leads to a very robust and long term attachment to the body, and avoids the artifacts commonly associated with capacitive sensors [14]. Most capacitive ECG approaches, for example embedding conductive material in a T-shirt, do not maintain intrinsic adhesion to the body. As a result the electrodes can move relative to the body, particularly in the orthogonal direction. In turn this alters the distance between the body and the electrode, altering the coupling capacitance and hence the signal gain and SNR. Our conformal approach overcomes this. In addition, this arrangement means that the printed material is not in direct contact with the subject for bio-compatibility, and that the electrodes can be placed directly over small amounts of hair. A layer of hair increases the gap between the conductors inside the body and electrode metal (Fig. 2), which will decrease the coupling capacitance and reduce the signal amplitude, but the capacitive connection will still be present allowing signals to be recorded. Any such effects are included within our results presented here.

The electrodes are inkjet printed using an in-house process with three layers of electrically conductive Silver nanoparticle paint, using similar procedures to [15] previously used for printing antennas [16]. The substrate is temporary transfer tattoo inkjet paper with an adhesive film laid over it [17]. The wanted electrode shapes were drawn in Adobe Illustrator and exported as bitmap files. To manufacture the wanted shape bitmap patterns were processed using a quadrant pixel mask

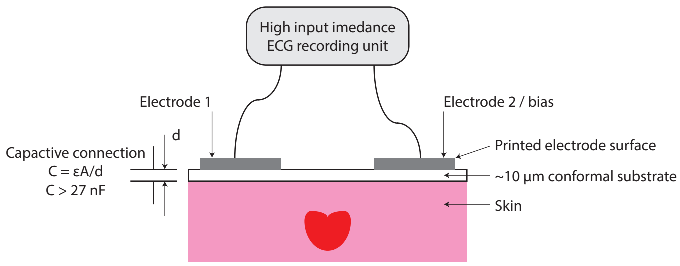

Fig. 2. Our electrodes maintain a capacitive connection to the body with an adhesive conformal substrate behind the entire electrode surface.

in the GNU Image Manipulation Program (GIMP). The masks were designed to remove 75% of the image pixels reducing the amount of ink printed in any one layer. This improves the printed quality, and reduces the amount of ink required to lower cost. In this manner the pattern was split into four pixelated images which when overprinted comprise one whole image. Silver nanoparticle ink suspension (736465 Sigma-Aldrich) was printed using a 10 pL print head (DMC-11610) on a Dimatix DMP-2800 inkjet printer (Dimatix-Fujifilm Inc., USA). Each quadrant pattern was printed twice in succession onto temporary tattoo paper heated to 60°C to form one complete image. This was then repeated twice more to yield a three layer image. Finally, the paper was heated to 150°C for 30 minutes to sinter the metal layers.

After printing, the adhesive film layer was placed over the electrode surface to provide a firm attachment to the skin. This adhesive is pain free to remove [17] and FDA approved [18]. Upon application to a subject the backing paper of the temporary tattoo transfer material was removed by wetting, leaving behind the printed ink a 10 μm thick substrate. To give a two electrode ECG recording electrodes are manufactured in pairs, and connected to a high input impedance amplifier, Fig. 2.

B. Electrode design

For the initial prototype devices our printed electrodes were designed as a 25 mm circle with a 7 mm tab for connecting recording wires to. ECG standard 1.5 mm touchproof connectors were attached to this tab using Silver loaded paint, and in some cases a layer of surgical tape placed over the connection point to strengthen the contact. As an additive manufacturing approach with the advantages of rapid prototyping and little waste, inkjet printing offers the potential for personalising the electrode shapes to each individual person. This potential will be the subject of future work, see Section IV, once the initial prototypes presented here have demonstrated the fundamental operation of the conformal ECG electrodes using more basic shapes.

A picture showing the final fabricated electrodes when attached to the body is given in Fig. 3. Fig. 3(a) shows the printed ECG electrode on the left, together with a standard Ag/AgCl ECG electrode (Lessa, Spain) which is used for comparison. This traditional pre-gelled disposable Ag/AgCl ECG electrode has a 7 mm diameter electrode at the center of a 50 mm diameter adhesive patch. A UK 10 p piece is included in Fig. 3(a) for scale. Fig. 3(b) then shows two electrodes placed on a subject in two electrode ECG locations, one electrode on either side of the heart, with wires to connect the electrodes to an ECG recording unit.

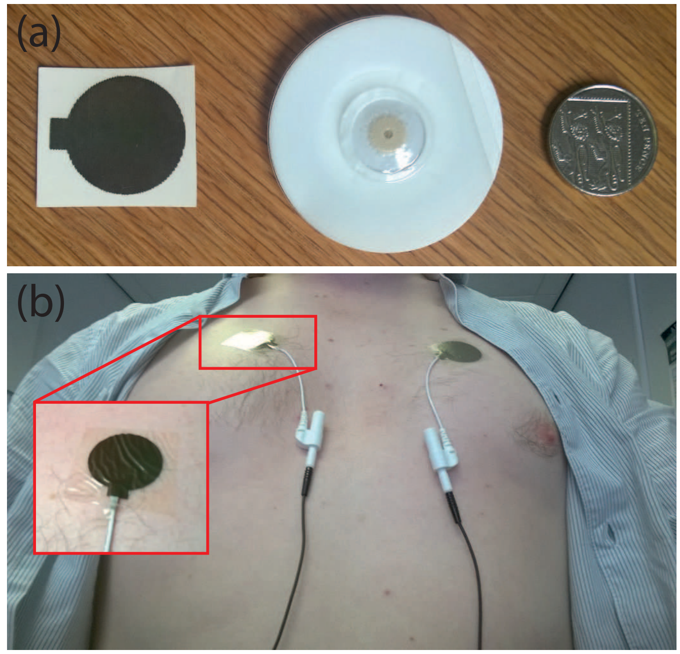

Fig. 3. (a) One of our printed conformal tattoo ECG electrodes, a traditional Ag/AgCl ECG electrode, and a UK 10 p piece for scale. (b) Example attachment to a subject.

III. Performance analysis

A. Overview

To assess the performance of our new electrodes they were connected to a high quality two electrode conventional ECG recorder (CamNtech Actiwave (Cambridge, UK)) and used to perform ECG measurements. Both electrodes in Fig. 3(b) are used to generate one ECG trace, and in any one recording both electrodes are of the same type (tattoo or Ag/AgCl). There is no explicit third/ground/driven right leg electrode in addition to the two recording electrodes. Common mode driving is provided simultaneously through one of the two recording electrodes. (See for example [19] on two electrode ECG measurements).

The ECG recorder is made using discrete components and is not integrated onto the conformal substrate, allowing us to characterize the performance of the conformal electrodes separately from the performance of the electronics that must also be mounted conformally in order to make a complete final system. The collected signals are then analysed as detailed in each section below to extract the electrode properties, principally the SNR and baseline wander. The electrode impedance is also assessed. These analyses are done in the time domain to reflect the electrode performance when visually inspecting the signals, and also in the time–frequency domain to reflect the electrode performance when automatically extracting information. In all cases the time–frequency domain information is calculated using the Continuous Wavelet Transform (CWT) power, found as the squared wavelet coefficients using a scale 10 Mexican hat mother wavelet.

Five two hour ECG recording tests were performed using the conformal ECG electrodes, during which the subject was stationary and using a computer for standard office work. In a separate recording session conventional Ag/AgCl electrodes were attached to the subject and used to record two hours of ECG for comparison purposes. This recording was done non-simultaneously with the conformal electrodes in order to avoid the ECG common mode driving from one electrode set up interfering with the common mode as seen by the other electrodes. If both are used simultaneously the Ag/AgCl electrodes measure and apply back a common mode voltage which affects the assessed performance of the conformal electrodes as when used in isolation they would measure and feedback a different common mode signal.

At the end of the recording the electrodes were either removed, or the wires to the electrodes removed and the conformal electrodes kept in place for 5 days. On each subsequent day, at the same start time to within half an hour, the wires were re-attached (see Section IV) and repeat measurements using the conformal electrodes were taken to assess their potential for very long term recordings.

All recordings used a 10 bit resolution, 1024 Hz sampling rate, downsampled to 256 Hz with a 50 Hz powerline notch filter prior to analysis in MATLAB.

B. Signal quality

To illustrate the collected ECG signals two sections of data collected using the conformal electrodes are shown in Fig. 4 in the time domain, and in the CWT domain to highlight the R peaks. A section of ECG recorded using an Ag/AgCl electrode is included for comparison. In both conformal cases a high quality ECG trace is seen with a clear QRS complex and R peak which is readily highlighted in the time–frequency domain. Although the amplitude of the entire ECG trace is in general smaller than when using Ag/AgCl electrodes, the P and T waves can be seen in the conformal traces.

To quantify the signal quality, and to summarize the quality of more than just the short section of data that can be illustrated visually, the SNR of each heart beat has been calculated. There is no standardized formula for ECG SNR, here we have calculated it as:

1) The estimated ECG baseline wander is removed (see Section III-C).
2) The location of each R peak is identified in the CWT domain, thresholding points where the CWT power is over 10^{-25} for the conformal electrodes and 0.5×10^{-24} for the Ag/AgCl electrodes, and then searching within 10 samples of this in the time domain to find the maximal signal point.
3) The middle third of the ECG between R peaks is extracted, which avoids ECG signal components, other than the T wave, being in this section.
4) This middle third section of data is detrended using a similar procedure as for the baseline wander removal (see Section III-C) to remove low frequency and T wave components.

TABLE I
TIME DOMAIN SIGNAL-TO-NOISE RATIO OVER TWO HOURS.

Electrode	Record	SNR / dB Mean	Max	Min	Standard deviation
Ag/AgCl	Application 1	51.1	55.8	24.2	2.3
	Application 2	53.2	57.3	18.8	1.8
	Application 3	51.0	56.4	18.5	2.8
	Application 4	50.0	56.3	16.7	3.2
	Application 5	50.2	56.7	06.3	3.4
Tattoo	Application 1 day 1	34.6	45.4	14.7	3.3
	Application 1 day 2	37.8	49.6	22.4	3.1
	Application 2 day 1	29.3	48.8	02.4	5.2
	Application 2 day 2	26.8	43.7	09.5	4.5
	Application 3 day 1	30.0	45.8	03.6	7.7

5) The Root Mean Square (RMS) of this detrended data section is found and considered as the noise present.
6) The SNR of the ECG is calculated for each heart beat as 20log(Found R peak height after baseline wander removal / detrended RMS noise between R peaks).

The detected R peaks were checked by eye and incorrect detections eliminated from the SNR calculation. Results are given in Table I and Table II for the time domain and CWT domain results respectively, with a total of total of 27,193 beats present for the 5 Ag/AgCl electrode data records and 23,654 for the 5 conformal electrode data records, giving large data set for robustly finding the SNR. For interpreting the results note that any residual T wave activity in the noise section may lead to an increased estimate of the noise present and so an under reporting of the SNR values here, and that each new application corresponds to the electrodes being removed and a new set attached on a subsequent day. Neither electrode type is re-usable after removal from the skin.

The Ag/AgCl electrodes outperform the new conformal electrodes in the time domain, with higher mean SNRs present. Nevertheless the mean SNR of the conformal electrodes is very high, in excess of 26 dB in all cases which allows clear identification of the R peaks. In general the reduction in SNR is due to both a decrease in the R peak amplitude (mean 431 μV and 945 μV for conformal and Ag/AgCl respectively) and an increase in the noise floor (14.2 μVrms and 2.9 μVrms respectively). A similar pattern is seen in the CWT domain. Here the SNRs are in general higher than in the time domain. The Mexican Hat wavelet can be seen as a (semi-)matched filter for QRS complexes, and therefore increases SNR. Although the SNR is higher for the Ag/AgCl cases the mean for the conformal electrodes is always greater than 38 dB. Again this allows easy signal detection for a very large number of R peaks.

C. Baseline wander

Many methods for quantifying and removing baseline wander are available in the academic literature. In this article the residual baseline wander is estimated by performing a nine level Discrete Wavelet Transform (DWT) with the db4 wavelet as this allows the same procedure to be used for both baseline wander estimation and detrending the noise component in

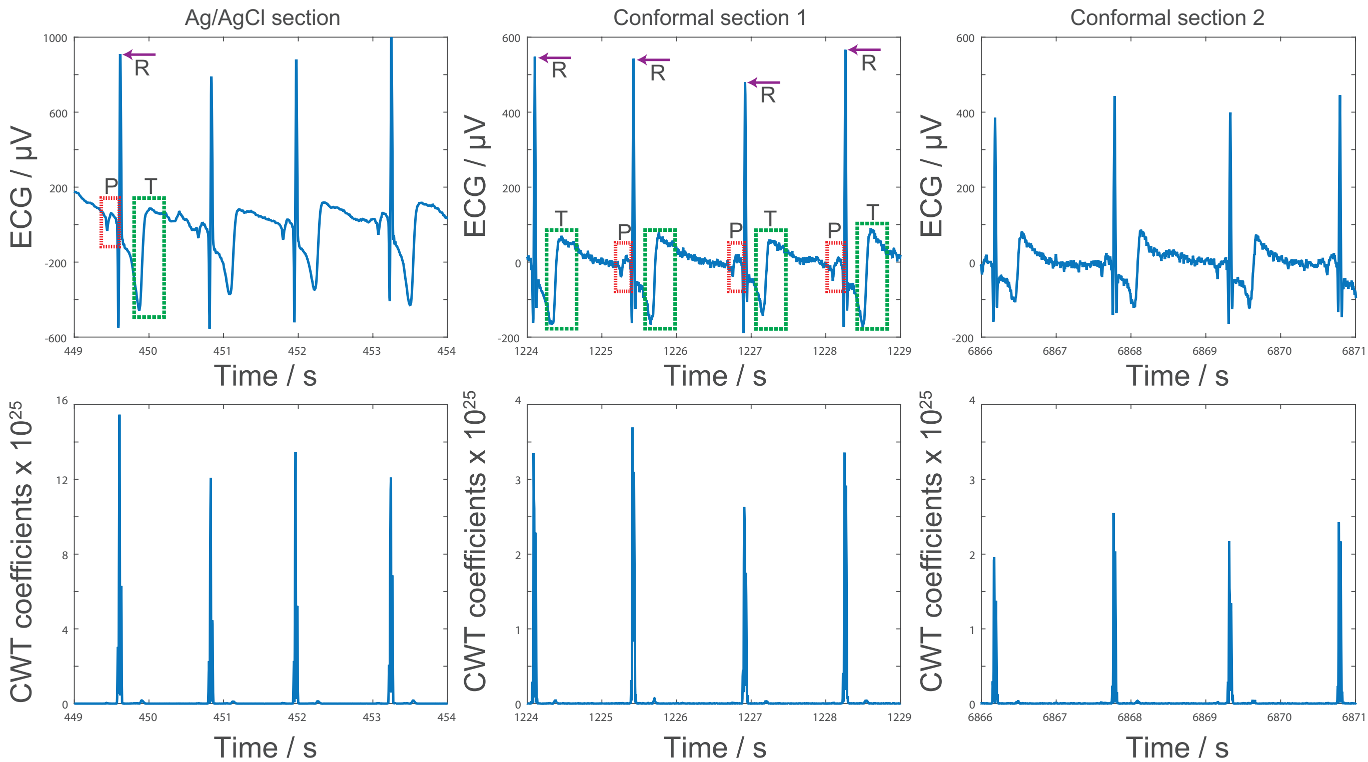

Fig. 4. Two illustrative sections of ECG data recorded using conformal electrodes to demonstrate the data quality. These are taken from the same recording at different points in time. R peaks are clearly seen in both the time and CWT domains. R peaks, and the P and T waves are annotated on the conformal section 1 trace, and a section of ECG data recorded with Ag/AgCl electrodes is included for comparison.

TABLE II
CWT DOMAIN SIGNAL-TO-NOISE RATIO OVER TWO HOURS.

Electrode	Record	SNR / dB			
		Mean	Max	Min	Standard deviation
Ag/AgCl	Application 1	87.0	094.0	40.2	03.7
	Application 2	90.3	097.2	22.0	03.2
	Application 3	88.1	095.8	27.3	05.0
	Application 4	85.8	094.4	20.5	05.7
	Application 5	89.9	101.8	19.7	06.3
Tattoo	Application 1 day 1	52.7	068.6	06.8	06.9
	Application 1 day 2	57.7	081.1	21.9	05.9
	Application 2 day 1	44.3	067.8	00.0	10.0
	Application 2 day 2	39.9	086.9	−05.4	09.0
	Application 3 day 1	38.7	080.0	−16.0	18.9

TABLE III
RESIDUAL BASELINE WANDER OVER TWO HOURS.

Electrode	Record	Baseline wander / μV				
		Mean	Max	Min	SD	MAD
Ag/AgCl	Application 1	−0.03	0379	−421	12.78	6.26
	Application 2	−0.02	0044	−035	06.01	4.65
	Application 3	−0.05	0108	−185	08.26	6.18
	Application 4	−0.03	0266	−237	16.83	8.10
	Application 5	−0.02	0462	−718	15.51	7.74
Tattoo	Appl. 1 day 1	−0.03	0045	−0056	04.17	2.83
	Appl. 1 day 2	−0.03	0260	−0213	05.53	2.02
	Appl. 2 day 1	−0.03	0634	−0417	20.98	7.21
	Appl. 2 day 2	−0.09	1192	−1646	54.07	12.47
	Appl. 3 day 1	−0.02	0226	−0235	18.03	12.02

Section III-B. The only difference is that in Section III-B a four level analysis is used.

The level 9 approximations at our sampling frequency correspond to frequency content in the range 0–0.25 Hz, and the DWT approximation signal is then found from a single level reconstruction using the DWT approximation coefficients [20]. This single scale estimation of the baseline wander is similar to the DWT baseline wander estimation methods reported in [21], [22]. Alternatively, as a single DWT scale corresponds to a fixed frequency range (0–0.25 Hz here) the baseline estimation could be implemented as a single FIR (Finite Impulse Response) filter as reported in [23], [24].

The DWT is applied to the two hour recorded traces, excluding the first and last 10 s, and the approximation signal is taken as the estimate of baseline wander and used to find the mean, max, min, Standard Deviation (SD) and Mean Absolute Deviation (MAD) of the residual signal components. The values of these parameters are given in Table III. For baseline wander removal the approximation signal estimate of the wander is subtracted from the raw ECG trace and this procedure is found to give good baseline wander removal performance, provide compact support via the db4 wavelet, and is very efficiently implemented in hardware [20] and in MATLAB. Our DWT processing is offline, and so although filters and mother wavelets with smaller group delays are avaiable this is not a factor of importance and we make use of the very common db4 mother wavelet.

All of the records have very small mean levels of baseline wander, zero on average, showing that no long term drift/offset is present. The values for the conformal electrodes are in the same range, $< 0.1\ \mu$V, as the Ag/AgCl electrodes. The standard deviation values are also similar, in the range $< 20\ \mu$V, with two of the tattoo records having smaller standard deviations over 2 hour periods than any of the Ag/AgCl recordings. The record from application 2 day 2 is an exception to this. It shows a substantially higher mean and standard deviation than the other conformal electrode records. The start of this record has a section of data with a series of very large artifact transients, most likely substantial motion or a knocking of the recorder which briefly saturates the amplifier and causes a refractory period due to the hardware filter responses. If the first two minutes of this record are discarded the extracted baseline wander values are very similar to those for other records.

D. Contact impedance

For the conformal electrodes the adhesive substrate has a relative permittivity of approximately 2.5, and the ink carrier a relative permittivity of approximately 3 [25]. At 30 Hz these suggest a contact impedance for our electrode size and distance from the skin of approximately 4 MΩ, beyond our electro-physiologically safe measurement capability.

For some measurements an additional electrode was placed on the mid-line, between the two recording electrodes in order to allow electro-physiological standard three electrode impedance measurements using a SIGGI II meter (EasyCap, Germany), and this confirmed that the conformal electrodes had impedances above 200 kΩ, beyond our measurement capability, at the start and end of the recordings.

In contrast, for non-prepared skin (no alcohol swab, no outer layer abrasion) to give a comparable set up case to the tattoo electrodes which do not require any skin preparation, the Ag/AgCl electrode impedance at 30 Hz was 96 kΩ at first attachment, falling to 67 kΩ after two hours of use. Electrode impedance is related to the amount of powerline (50/60 Hz) interference [26], and large amounts of powerline noise is seen in the raw recorded traces for our electrodes. However this is easily removed by a notch filter, and is not found to be a significant source of interference, as can be seen in the ECG traces in Fig. 4.

E. Longevity

To investigate the multi-day sensing capabilities of our electrodes Fig. 5 illustrates five short segments of ECG recordings performed at approximately the same time on subsequent days. In all cases a clear QRS complex is seen, allowing heart rate and heart rate variability analyses to be performed. In these sections of data the height of the R peak decreases over time and more high frequency artifacts are seen in the trace, particularly on day 5. Nevertheless a high SNR is clearly maintained.

To quantify this performance, Fig. 6 shows the SNR of the collected data on subsequent days, calculated using the same analysis procedure as in Section III-B. Here the SNR is calculated using 10 minute sections of ECG data which do not

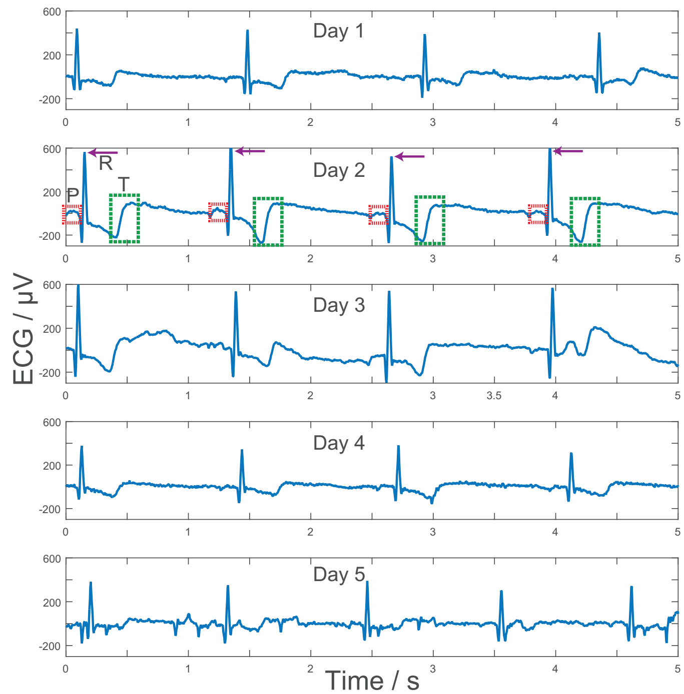

Fig. 5. Illustration of the ECG recording performance over 5 days. Some P and T waves are present, but more difficult to observe at this scale, and their presence decreases over time as the ECG SNR decreases. They are most readily observable, and annotated, in the day 2 trace.

include motion or wire connection artefacts (Section III-F). In most cases the wires were securely attached for much longer than 10 minutes, but in some records the wire disconnected, giving this as the maximum duration recording possible. This wire connection is the weak point in the current system design, and is discussed fully in Section IV. Section III-B uses the same amount of data from all records for consistency, allowing Fig. 6 to show the quality of the ECG signal itself over time when no motion or wire connection artefacts are present.

The results in Fig. 6 clearly demonstrate the ability of the electrodes to collect high quality data over multiple days. There is a systematic decrease in SNR on days 4 and 5, but still above 30 dB on average. On days 2 and 3 the average SNR was higher than on day 1. This could indicate an electrode settling time where the electrodes get better contact after a settling period. Such a settling is commonly seen in electro-physiology electrodes [27] and would not be reflected in the results of Table I where for most of the recordings the data was collected immediately after the electrodes were attached.

In Fig. 5, although the ECG recording itself was not continuous over the full 5 days, the electrodes were not removed and they stayed in place for the duration. The 5 days included all standard activities, including exercising, showering and similar. Care was taken to not accidentally put soap on or to rub-off the electrodes, particularly when towelling, but otherwise no special preparations were performed. There is thus substantial potential for printed electrodes to enable chronic ECG recordings over time spans not currently possible without having to impact the activities of daily life. As an

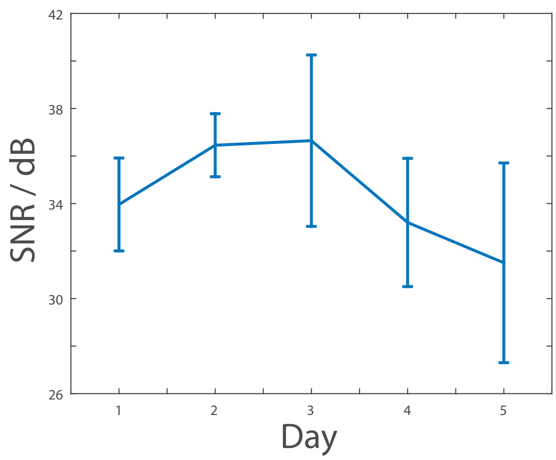

Fig. 6. Time domain SNR of ECG data collected across 5 days of electrode attachment in periods with no motion or wire connection artefacts. Figure shows the mean plus/minus one standard deviation.

illustration, one subject (data not shown in Fig. 5) did a 20 mile run and the electrodes were still attached at the end.

We did not perform a formal survey of comfort of the electrodes over time or discomfort on removal, and these should be the study of future work. Anecdotally we received no reports of discomfort from the electrodes, and removing the electrodes caused no discomfort. This is in contrast to removing conventional Ag/AgCl electrodes which is often painful, particularly if they overlap with hair and have been in place for a period of time.

F. Signal artifacts

As with all ECG electrodes our new electrodes collect a number of artifacts together with the desired signal and a representative selection of these is shown in Fig. 7. Fig. 7(a) shows a section of EMG from tensing the chest, which superimposes on top of the ECG. The R peaks are still visible in the time domain. Fig. 7(b) shows a section of data collected while walking, demonstrating that substantial motion artifacts are present. These motion artifacts will be a combination of EMG noise, respiration, movement of the skin relative to the heart during motion, cable movement, and mains and other environmental pick-up. Fig. 7(c) shows a section of recorded data while the subject was stationary and just the cables moved. This introduces dominating artefacts, with the same morphology as those in Fig. 7(b), demonstrating movement artifact due to movement of the cables used to connect the electrodes to the ECG unit, rather than intrinsic to the electrodes themselves.

Fig. 7(d) shows other examples of cable artifacts. These occur as the wire connection to the electrode varies and become increasingly frequent on later days of recording as keeping the wire attached to the electrode becomes more difficult (which is why only 10 minutes of each day is analysed in Fig. 6). The aim of the current work was to characterize the electrode performance using a high quality ECG recorder. When we integrate the recording electronics on to the conformal substrate as well, eliminating the external

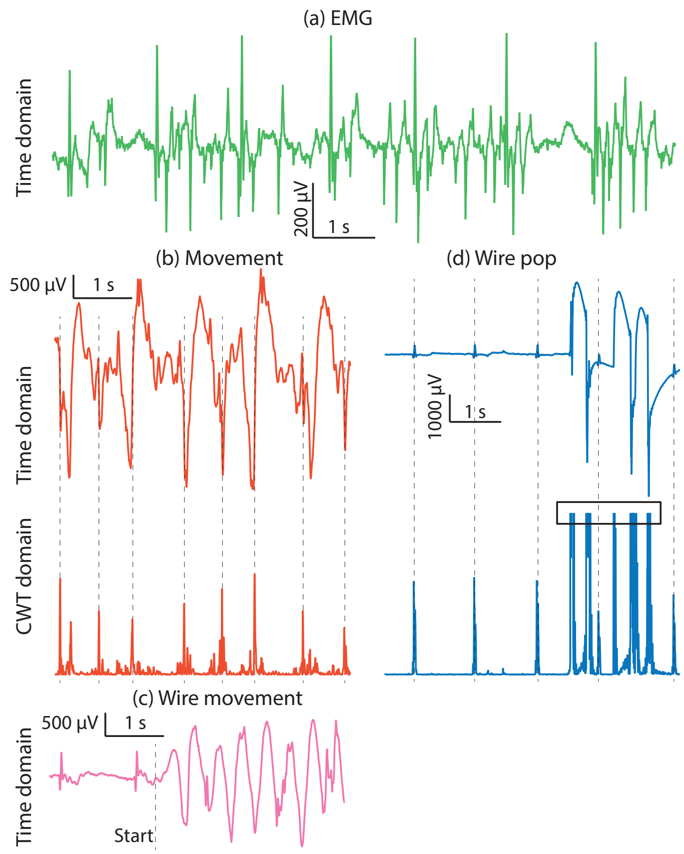

Fig. 7. A selection of ECG artifacts collected by the conformal electrodes. (a) EMG artifact in the time domain. (b) Movement artifacts during walking in the time and CWT domains. (c) Motion artefacts due to only movement of the recording wires while the subject is stationary. (d) Wire connection artifacts in the time and CWT domains. Data in the black box section has been clipped for presentation.

recording wires used in this study, many fewer of the Fig. 7(b) and Fig. 7(c) artifacts will be present.

Despite the significant potential artefacts, candidate R peaks are observable in the time domain in Fig. 7(b), and are clearly highlighted in the CWT domain. Nevertheless, the artifacts mean that we have not yet achieved reliable heart rate detection using these sections of data. At present our electrodes are only suitable for stationary or quasi-stationary long term monitoring. To our knowledge (see Section IV) there have been no previous systematic studies on the motion artefacts obtained using conformal electrodes. Combined with more sophisticated signal processing for R peak detection (particularly Kalman or particle filtering to eliminate incorrectly detected peaks [28]), or the collection of co-located artifact correlated signals (accelerometer or impedance) to remove the artifact using adaptive filter techniques already established for ECG signals [29], it is likely that a heart rate will be extractable during motion periods, but substantial algorithm development work will be required first.

IV. Discussion

We have demonstrated inkjet printed ECG electrodes and the collection of high quality ECG signals up to 5 days after the electrode attachment by using conformal substrates to achieve a good connection between the electrode and the human body. This overcomes the need for a conductive gel as in conventional Ag/AgCl electrodes which dries out over time and the new electrodes are also very quick and easy to apply and remove. Our electrodes are capacitively coupled with an insulator gap present instead of a conductive gel. Similar capacitive ECG electrodes have been researched since at least the 1960's [30], can be made as the back layer of a PCB [31] for easy integration, or integrated into fabrics for smart clothing [32].

The major advantage of our conformal approach to capacitive electrodes is that it provides adhesion under the entire electrode surface. This leads to a very robust and long term attachment to the body, and avoids the artifacts commonly associated with capacitive sensors [14] as the distance between the body and the capacitive electrode varies. Further, we have demonstrated high quality recordings over 5 day periods. It is likely that even longer records are possible, the electrodes were voluntarily removed rather than falling off at the end, but we do have any data to demonstrate this at this point in time. Note that as a limitation of the current study, in this work volunteers were not pre-selected for recruitment based upon how much chest hair they had. The subject in Fig. 3(b) should be taken as representative. In Fig. 3(b) it can be seen that some hair is present underneath the electrodes, but there will be people who are much hairier than this, and there be a strong gender, ethnicity and age component. Such an investigation is not a systematic part of our study. Instead we have focused on demonstrating and characterizing ECG recordings from inkjet printed conformal substrates for the first time.

Conformal electrodes for heart [33] and brain [34] monitoring have been presented previously, particularly by John Rogers' group at Northwestern University [1]. However, these are based on Silicon processing and PDMS stamp transfers, and are not directly suitable for personalizable electrodes as enabled by inkjet printing. Since Rogers' original work on conformal bio-signal sensing [1] a number of papers using similar manufacturing methods to enable conformal ECG sensing have been presented [35]–[43], although these focus on the materials construction and performance, rather than the ECG sensing performance as considered here. For example, [36] includes a picture of ECG recorded one week after electrode application, but it does not contain any across day quantification of the SNR as is presented here for the first time.

The advantage of our inkjet printing manufacturing method is that as a rapid prototyping, additive manufacturing approach it can allow personalization of the electrode sizes and shapes. Rather than having to shave the whole chest as in conventional ECG, we anticipate shaving only a small area and printing a customized electrode that matches the size and shape of this small area, giving a significantly better experience for end users. Compared to the conventional Ag/AgCl electrode in Fig. 3(a) for which the sensing Ag/AgCl part is 7 mm in diameter, at the centre of a 50 mm diameter adhesive patch, our electrode is 25 mm in diameter, with adhesive under the entire surface, and cut out as a 30 mm square. There is thus a much higher ratio between the sensing area and the adhesive area, and as the adhesive area determines the amount of hair that must be shaved, for the same amount of shaving a larger sensing area is possible using our tattoos.

A small number of papers have investigated inkjet printing manufacturing methods for ECG sensing previously. [44] investigated using inkjet printed wires to interconnect between conventional ECG electrodes, while [45] printed the wired interconnects and wireless antenna. To our knowledge, the only papers to inkjet print ECG electrodes as we do are [46], [47]. However: these electrodes are held in place using an elastic bandage rather than a long term tattoo as we present here; the printed metal is in direct contact with the skin, which may present bio-compatibility issues unlike our capacitive connection; the measured ECG results are simply described as "*adequate to detect the weak biopotential signal*" [46], there is no quantification of SNR, baseline wander and similar; and there is no assessment of the long term or multi-day use of the devices or estimation of their performance over time. The results in the current paper comprehensively describe the long term sensing performance of conformal ECG electrodes for the first time.

Finally, here we have focused on characterizing the performance of the ECG electrode itself. To make a complete end system the recording electronics must also be integrated onto a conformal substrate. This has not yet been done and this lack of integration introduces a number of issues, particularly with regards to the wire used to connect the electrodes to the ECG recorder. Our approach of adhering the wires to the electrodes using Silver loaded paint gives a suitable method for easy prototyping of different electrode shapes and manufacturing processes, but the attachment of the wire severely limits the operation of the system: it falls out long before the electrode itself looses adhesion to the body and sensing capability. It can be re-attached, but this requires time, expertise, and introduces artifacts into the recording. In addition to fully integrating the electronics onto the conformal substrate, a strategy for robustly connecting conformal devices to conventional rigid electrodes would be highly beneficial for emerging sensor nodes to allow them to take advantage of both conventional rigid (wearable) electronics and conformal devices.

V. Conclusions

Inkjet printed Silver nanoparticle ECG electrodes can collect high quality signals, with Signal to Noise Ratios over 30 dB. Further, this performance can be maintained for 5 days, allowing longitudinal bio-sensing over timeframes not feasible using conventional Ag/AgCl ECG electrodes. This paper has presented the design and characterization of suitable electrodes, quantifying their performance in terms of Signal to Noise Ratio and baseline wander, and showed example signals including typical artifacts. These new conformal electrodes show substantial promise for the rapid and low cost manufacture of personalized electrodes for future long term ECG monitoring.

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