Knitted ECG electrodes in relaxed fitting garments

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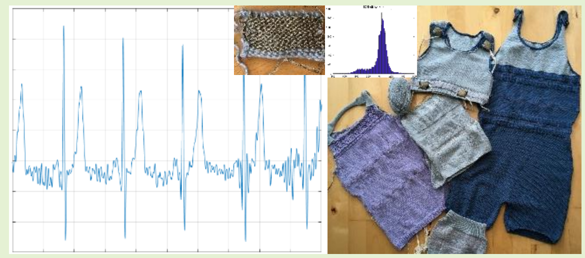
***Abstract*—A wide range of signal quality indices (SQIs) related to statistical methods, are used to guide the optimisation process of knitted textile electrodes for ECG recordings. The electrode structure and composition as well as their integration into a garment are evaluated in view of a fully knitted garment. The dry electrodes with the best SQIs are obtained by using conductive yarn only with a compact knit structure and a medium level of roughness. The best SQIs for the e-garment were obtained by sewing electrodes at their edges only, into the knitted garment. This implementation outperforms the intarsia and double-knit method as it allows the garment some independent movement from the electrodes, reducing motion artifacts. Tests done on a healthy volunteer demonstrate excellent system performance under gentle ambulation. The advantage of using SQIs in the optimisation process of dry textile ECG electrodes is that they offer a quantitative benchmark against which to compare other approaches. The fully knitted clothing allows for more relaxed e-garments when gentle ambulation is considered.**



***Index Terms*—ECG (electrocardiogram), health monitoring, knitting, textile electrodes, wearables.**

I. Introduction

ELECTROCARDIOGRAM (ECG) recordings are a standard technique to observe and analyse the heart function of patients [1]. Up to 10 electrodes are connected to the torso to record a 12 lead ECG. This gives a wealth of information on the electrical activity of the heart [2]. Even implementations with only 1 or 2 leads [3,4] can give good information, useful for wearable ECG recorders [5]. The challenge is that the electrical signals related to the heart function, recorded on the surface of the body, are small and influenced by different noise sources such as breathing, movement and mains interference [6]. The signals of many of these artifacts also overlap in the frequency range relevant to the ECG data. In a clinical setting, noise is minimised by having the patient at rest and using mains powered, sophisticated equipment. For e-garments, this is no longer possible as the person will be moving while recordings occur. It is therefore essential to optimise the electrode as well as its implementation into the garment to limit interference of these noise sources. Reduction of noise is often achieved by using tight fitting solutions [7]. This approach may well exclude some people, e.g. infants and the elderly who might not feel comfortable wearing tight garments [8]. Thus, a study of ECG recordings using less tight-fitting garments is of interest to the community. Knitted garments have some specific advantages compared to wovens and non-wovens. It is lightweight, deformable, comfortable to wear [9], commercial knitting equipment is advanced and with careful design, yarns in weft knits can be recycled. Wirings to the electrodes can be knitted in, maintaining full deformation potential of the knit and avoiding push buttons on the electrodes [10]. When considering contact-type dry ECG electrodes, their electrical requirements are [11]: low skin-electrode contact impedance, robustness to motion artifacts, stability and lifetime of electrodes, biocompatibility

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with the skin and wearing comfort. The deformability of knits, however, makes this approach sensitive to motion artifacts. Embroidery of conductive thread was suggested to minimise the movement artifacts as it produces a non-deformable area [12,13]. However, it still needs a tight fit and requires non-knitting steps to integrate the electrodes. Printing conductive inks on knitted fabrics is another way to define the electrodes [14]. While excellent ECG signals were obtained using this technique, it still needs tight fits or elastic belts. Nigusse et al. [11] created padded electrodes with sandwiched structures to reduce the required tightness. Alternatively, the roughness as well as the density of stitches can be varied by choosing stitch type and needle size to improve performance [9,15]. Optimisation of the electrodes only is not sufficient, the full garment performance needs to be considered. Our work builds on garment implementations in [16,17,18] and improves the dry electrode performance. A breathing implementation as in [19] can be easily added [20].

Our optimisation work relies on Signal Quality Indices (SQIs) derived from the ECG data. We use Ag-plated polyamide yarn for the electrodes with electrical resistivity $\rho < 80\ \Omega/m \pm 30\ \Omega/m$ [21] and 316L stainless steel, $\rho < 36\ \Omega/m \pm 30\ \Omega/m$ [22]. Ag-plated polyamide yarn has shown acceptable robustness in washing cycles [23] and is well tolerated by the skin. Future development in conductive yarns will make alternatives available that can optimise abrasion resistance.

The manuscript is organised as follows: section 2 discusses data acquisition and SQIs to select the best electrodes. Section 3 describes the electrodes and their performance. The best electrodes were integrated in a knitted garment and evaluated in section 4. Their performance under different ambulatory conditions is evaluated in Section 5, where we also compare the results of the looser fit to that with an elastic band.

II. Data Recording and Processing

All ECG data were obtained using the Cyton board from OpenBCI [24] that can measure up to 8 channels simultaneously. The data was sampled at 250 Hz and was wirelessly transmitted via Bluetooth to a laptop. Matlab [25] was used to process the raw data. Basic signal processing includes removal of the data at the start (hardware settling time, ~2 s) and end (motion interference from switching off hardware, ~10 s) of the recording. A Butterworth IIR bandpass filter (order 8, 0.5 Hz < f < 50 Hz) was used (except where otherwise indicated) followed by an elliptic notch filter at 50 Hz (UK mains).

An ECG recording consists of peaks and troughs that reflects the heart's electrical activity [1]. The different minima and maxima are labelled P, Q, R, S, T in order of occurrence during one heartbeat and recur periodically in healthy people (see Fig. 3). The R-peaks normally have the highest amplitude and are used to derive heart rate. R peaks in our recordings were identified by looking for the maxima in the signal at least 0.7 s apart. The median time difference between R peaks (heart rhythm) is used to define the heart cycle time window, denoted h. The total recorded data was then split in different cycles with one PQRST signal in each cycle by using windows of [-0.2h, 0.7h] around each identified R peak.

To determine the best implementation, data acquired from a healthy volunteer was compared to simultaneous readings with commercial red dot gel ECG electrodes (3M) and SQIs were extracted.

One SQI is the standard variation of the position of the R peaks, σ_R. This choice is acceptable as the volunteer showed consistent regular heart signals in 1 – 2 min recordings taken over several days (5×) within a month using commercial ECG electrodes. For these measurements, σ_R were smaller than 0.04 s. Thus, larger standard variations in R peak position can be assumed to be from the electrode system rather than from the biosignal, making this approach acceptable for the classification of the electrodes.

The difference between the mean and median cycle was taken as a second SQI. This SQI identifies the impact of outliers and is derived as:

$$MM = \frac{\sum_{t=1}^{n} |\bar{S}(t) - \tilde{S}(t)|}{\sum_{t=1}^{n} |\tilde{S}(t)|}$$

with t the sample number and n the total number of samples in the cycle. $\bar{S}$ is the mean and $\tilde{S}$ the median.

The third SQI is the signal-to-noise ratio (SNR). The SNR was determined by first subtracting each measured ECG cycle, $S_i(t)$ (i is cycle number) by the median, $\tilde{S}(t)$, giving an approximate value for the noise at each sample point, $N_i(t)$:

$$N_i(t) = S_i(t) - \tilde{S}(t)$$

The total signal-to-noise ratio is then calculated as:

$$SNR = \frac{C \times \sum_{t=1}^{n} \tilde{S}(t)^2}{\sum_{i=1}^{C} \sum_{t=1}^{n} N_i(t)^2}$$

with i the cycle number and C the total number of cycles. The extraction of $N_i(t)$ assumes that the median is noiseless. This somewhat overestimates noise because of small variations in the position of the R peaks due to the finite sampling time. This makes $N_i(t)$ artificially larger in the regions with the largest signal amplitudes and thus underestimates the SNR.

The value $\widetilde{RS}$ is the difference in amplitude between the S and R peaks. This value is an indirect indicator of the resistance of the skin-electrode contact in the frequency range given by the bandpass filter. The higher the voltage difference, the lower the contact resistance. Good quality contacts have $\widetilde{RS}$ > 500 μV.

The final SQIs are the kurtosis (kSQI) and skewness (sSQI) [26] of the signal, given by:

$$kSQI = \frac{1}{N}\sum_{j=1}^{N}\left(\frac{S_j-\bar{S}}{\sigma}\right)^4$$

$$sSQI = \frac{1}{N}\sum_{j=1}^{N}\left(\frac{S_j-\bar{S}}{\sigma}\right)^3$$

where the sum, mean and standard deviation are computed over all sample points N. These parameters, associated with data statistics, are used to evaluate how far the ECG signal distribution deviates from a normal Gaussian. It assumes that the noise has a Gaussian distribution while an ECG signal does not. This is illustrated in Fig. 1, giving a nearly noiseless and a motion-artifact-influenced histogram of ECG recordings.

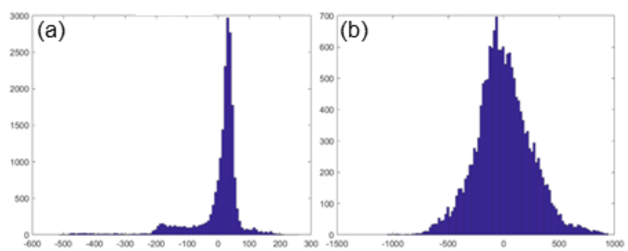

Fig. 1. Histogram of ECG recordings. (a) A nearly noiseless recording kSQI = 8.6 and sSQI = -2.18. Volunteer sitting still, using commercial electrodes. (b) A recording influenced by motion artifacts due to walking, using dry knitted electrodes, kSQI = 3.6 and sSQI = 0.24.

A high kSQI value (kSQI > 3) is representative for a non-Gaussian ECG signal, thus one where the noise is smaller than the signal. Similarly, a good quality ECG signal has a non-symmetric distribution, requiring |sSQI| > 0. In cases where the histograms have long tails, these values evaluated separately do not necessarily give a clear insight into the quality of the signal. Nardelli *et al.* [27] introduced a higher order statistical SQI, hSQI, that combines kSQI and sSQI:

$$hSQI = |sSQI| \times \frac{kSQI}{5}$$

Based on observations in this work, and in accordance with literature, we choose kSQI > 5 and |sSQI| > 1 for good electrodes. However, very high kSQI means that the histograms have heavy tails or the data many large outliers. Thus, we limit acceptable kurtosis to 5 < kSQI < 20. For good ECG signal quality, we choose 1 < hSQI < 10.

III. Optimisation of Electrodes

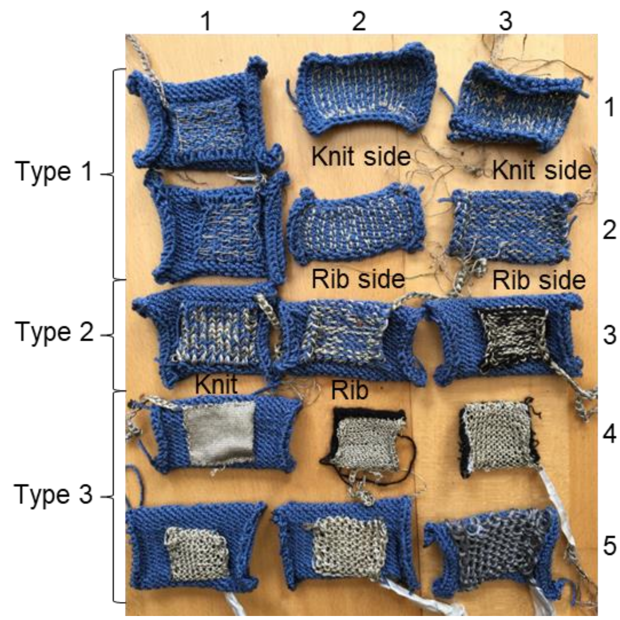

Fig. 2. Different knit implementations ordered in rows and columns (x,y). Type 1 compares embroidery, crochet and a hybrid knit. Type 2 are hybrid knitted structures with conductive and non-conductive yarn. Type 3 are conductive yarn implementations with increasing thickness. (3,5) is knitted in stainless steel yarn.

While multiple knitted ECG electrodes have been presented in literature [11,12,28], in our work we select the knit-in-garment concept and systematically optimise their performance. Ag-coated polyester yarn [21] was used in all implementations (ShY), except (3,5) uses stainless-steel yarn (SsY) [22]. The non-conductive yarn (ncY) is 92% acrylic and 8% PBT elastane. Fig. 2 shows the different implementations of the electrodes. Three types of electrodes were implemented ordered in 3 columns and 5 rows and referred to as (x,y) with x the column and y the row.

Type 1 are hybrid structures combining ncY and one thread of ShY using different techniques: embroidery, crochet on top of the knit or by knitting two yarns simultaneously. Previous studies [9,15] have shown that a rougher contact can lead to better skin-contact impedance. Therefore, we evaluate both knit and rib side. Similarly, a higher knit density reduces the electrode's resistance and leads to better skin contacts [28]. Thus, type 2 electrodes are knitted with three threads of ShY and in combination with either an elastic or non-elastic yarn. The type 2 implementations are examples of double knits (two layers knitted simultaneously to insulate the electrode at the front). Type 3 are all non-hybrid implementations. The electrode side contains only ShY and the front is ncY. The electrodes are knitted with different number of ShY threads, the needle size is adapted to the yarn diameter to maintain good compactness of the stitches. Increasing the number of ShY increases the roughness of the rib side of the knit. In addition, a commercial machine-knitted fabric with smooth surfaces [29] is used by sewing it on top of the knit. While the ShY feels comfortable on the skin, the SsY might, for some people, lead to skin irritation.

TABLE I

Parameters extracted from the ECG recordings on the wrist for all electrodes in Fig. 2. CG is the commercial gelled electrode ; * is non-elastic yarn. $\overline{\Delta R}$ is the mean distance between the R-peaks and σ_R is standard deviation. $\widetilde{RS}$ is the voltage difference between R and S peak. MM is the normalsed difference between mean and median. SNR** = SNR/C is a value for the signal-to-noise ratio. kSQI is the kurtosis, sSQI the skewness and hSQI the higher order SQI. The bold numbers identify reaching the signal quality criteria.

Electrode	Name	Technique	side	#threads yarn	Needle mm	$\overline{\Delta R}$ s	σ_R s	$\widetilde{RS}$ mV	MM	SNR**	kSQI	sSQI	hSQI
CG	CG	commercial	NA	NA	NA	1.02	**0.04**	**591**	**0.04**	**0.437**	12.55	-2.38	**5.96**
(1,1)	1ShYe	embroidery	knit	1 ShY	3	1.07	0.29	250	0.76	0.002	14.85	-1.27	**3.77**
(2,2)	1ShYcr	crochet	rib	1 ShY	3	1.22	0.84	388	0.62	0.003	98.31	6.07	119
(3,1)	1ShYk	knitted	knit	1 ShY	3	1.11	0.53	334	0.43	0.003	44.42	-0.17	**1.53**
(3,2)	1ShYr	knitted	rib	1 ShY	3	1.05	0.20	392	0.21	0.018	63.19	-4.28	54
(1,3)	3ShYk	knitted	knit	3 ShY	3	1.09	0.45	**528**	0.19	0.015	87.56	-5.71	100
(2,3)	3ShYr	knitted	rib	3 ShY	3	0.98	**0.08**	**502**	0.28	0.005	78.76	5.33	84
(3,3)	3ShYr*	knitted	rib	3 ShY*	3	0.98	0.16	476	0.19	0.014	67.88	3.35	46
(1,4)	ShF	commercial	knit	NA ShY	NA	1.03	0.14	**508**	0.22	0.011	13.89	0.98	**2.73**
(2,4)	1ShY	knitted	rib	1 ShY	1	0.96	0.18	477	0.40	0.005	49.92	3.49	35
(3,4)	2ShY	knitted	rib	2 ShY	1.75	0.93	**0.09**	**543**	**0.07**	0.061	15.96	-2.19	**6.98**
(1,5)	3ShY	knitted	rib	3 ShY	2	0.99	**0.04**	**504**	**0.03**	**0.291**	6.58	-1.66	**2.19**
(2,5)	4ShY	knitted	rib	4 ShY	2.5	1.03	0.25	**545**	0.47	0.004	78.54	5.12	81
(3,5)	2SsY	knitted	rib	2 SsY	2.5	1.00	**0.11**	**506**	**0.05**	**0.125**	6.62	-1.68	**2.23**

The electrode surface areas are approximately equal and of a size like the CG electrodes. Previous studies [9] have shown that larger contacts improve the skin-electrode contact impedance and thus the ECG signal. The variations in area in Fig. 2 are sufficiently small to be negligible. All knitted contacts have a resistance from electrode to the Cyton board of $R_{tot} < 0.7\ \Omega$.

Measurements to determine the quality of all electrodes were done by clamping them underneath a gym wrist sweatband and measuring between right and left wrist (lead I of the Einthoven triangle [27]). This will generate approximately the same pressure on all contacts. Commercial red dot gel ECG electrodes were measured in the same way. All measurements can be found in the supplementary information (S1).

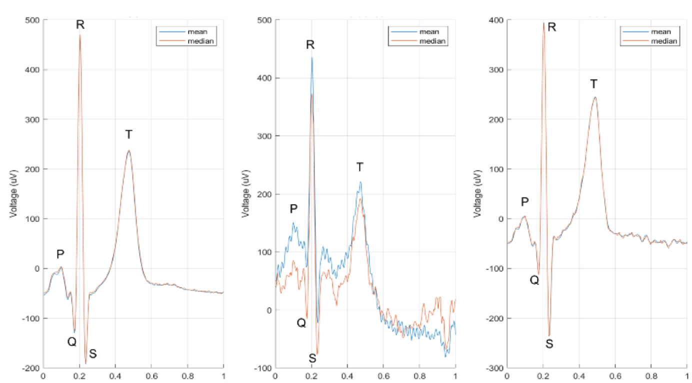

Fig. 3. The mean and median of 1 min ECG recordings taken on the wrist. From left to right: commercial gelled ECG contact CG, contact (2,2) 1ShYcr and contact (1,5) 3ShY.

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Fig. 3 are examples of the mean and median for three different implementations. In recordings with good quality electrodes, the difference between the two is small and the noise in the time regions after the T peak: ~0.6 s < *t* < 1 s, is small. This shows that comparing the mean and median of the ECG cycles gives a quick insight into the quality of the electrode and justifies the MM as quality factor.

Fig. 4. Knitted garment with integrated electrodes in border (identified by square dashed lines). 1. Intarsia. 2. Double knit with cotton padding. 3. Separately knitted contacts.

Table 1 summarises the SQIs for the different electrodes. The distance of the R-peaks is approximately the same for all electrodes, indicating that no R peaks are missed (also visually confirmed in the full 1 min. measurements). To select the optimal electrode, the SQIs must obey the following criteria:

- $\sigma_R < 0.1$ s
- MM < 1
- SNR > 0.1
- 1 < hSQI < 20

The values in Table I that fulfil these criteria are in bold. We conclude that the best electrodes are 3ShY (1,5) and 2SsY (3,5), with 2ShY (3,4) nearly achieving all criteria. Observations are:

- Non-hybrid implementations perform better than hybrid implementations because the effective contact area is increased, increasing the signal amplitude differences. The noise decreases (see hSQI) because the electrical characteristics of non-hybrid structures vary less with movement than hybrid structures.
- The rib face is better than the knit side. This is associated with an improved electrode-skin contact. However, too high roughness, e.g. 4ShY (2,5), deteriorates the contact quality, limiting the allowed roughness.
- The commercially knitted fabric [29] performed worse than the rougher knits and showed more baseline wander. Baseline wander is also present for the rib side of hybrid knitted electrodes.

IV. Implementation in Garment

The best non-hybrid knitted electrode (3ShY; (1,5)) was selected for integration into a knitted top (Fig. 4). Two electrodes were integrated on the knit border at the front of the chest, in line with recommendations in [30]. The top was knitted in the round (no seams) using 93% acrylic & 7% PBT yarn and 2.5 mm needles for the body and 2 mm needles for the border where the electrodes are integrated. Using 2 mm needles reduced the deformability of the knit resulting in less motion artifacts. The electrodes were added in 3 different ways (Fig. 4): intarsia (jacquard, cfr. [16]); double knitting (front layer ncY and electrode layer: 3ShY) and separately knitted electrodes sewn at their edges onto the border. These electrodes have an electrode layer in 3ShY and front layer in ncY and were knitted in one piece, then folded and sewn. Cotton padding was added between back and front layer.

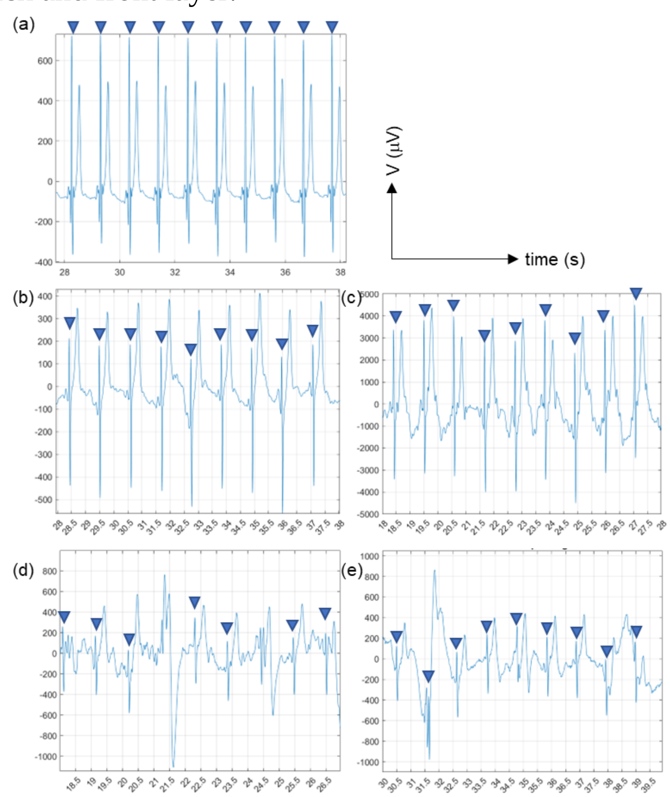

Fig. 5. A 10 s snapshot of the ECG signal recorded with the different electrode implementations in the garment. The knitted contacts are all 3 ShY. (a) commercial gel electrode. (b) separately knitted and sewn-in electrodes. (c) intarsia. (d) and (e) double knitted contact without and with cotton padding, respectively. Arrows identify R-peaks.

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The first and second implementations are advantageous because the electrodes are knitted at the same time as the top and, when using weft knitting, can be recycled by pulling on the single yarn with which the whole knit was carried out. The third method has the advantage that the electrode can be attached to the garment in a way that allows the garment some independent movement from the electrode. This helps reduce motion artifacts but makes recycling more cumbersome.

Specific electrode placement can help to improve the response of the dry electrodes by minimising motion artifacts as described in [31]. In our design, the electrodes' position was chosen in accordance with the traditional Holter method.

ECG recordings were taken with the volunteer sitting down (static position). All recordings were done simultaneously with gelled commercial (CG) contacts in slightly shifted positions. ~10 s snapshots of the ECG recordings are given in Fig. 5. All allowed identification of the R-peaks, except the double knit implementations (Fig. 5 d,e). To facilitate extraction of the SQIs and other performance parameters, these signals were filtered with a 3 Hz to 50 Hz band pass filter (standard band pass was 0.5 – 50 Hz).

TABLE II
SIGNAL QUALITY PARAMETERS EXTRACTED FROM THE ECG RECORDINGS WITH THE DIFFERENT ELECTRODE IMPLEMENTATIONS IN THE BORDER OF THE TOP. ALL WERE DONE UNDER STATIC CONDITIONS. THE NUMBERS CORRESPOND TO THOSE IN FIG. 4. THE LOW PASS FREQUENCY OF THE BAND PASS FILTER IS INCREASED TO 3 HZ FOR 2 AND 2*. * IS THE DOUBLE KNIT WITH COTTON PADDING.

	$\overline{\Delta R}$ s	σ_R s	MM	SNR**	kSQI	sSQI	hSQI
CG	1.08	0.03	0.02	0.528	8.99	-2.03	3.65
1	1.08	0.08	0.05	0.054	4.45	-0.7	0.62
3	1.1	0.04	0.05	0.221	5.94	-0.56	0.67
2	1.08	0.27	1.13	0.001	46.41	1.1	10.2
2*	1.05	0.26	0.93	0.001	49.24	0.8	7.84

All results are given in Table II and show that implementation 3 is the best electrode, closely followed by the intarsia approach.

V. PERFORMANCE UNDER GENTLE MOVEMENT

It is well known that ECG recordings using dry electrodes, especially implemented in a more relaxed garment, suffer from motion artifacts. To evaluate the performance of our optimal implementation, four different levels of movement were tested: lying down, sitting, standing and a gentle walk. In these measurements, the left electrode in the garment was shifted ~2 cm further to the left to improve robustness of the recordings by insuring opposite polarities of the biopotential under the electrode during movement [32]. CG electrodes were stuck on the wrist for simultaneous control recordings. The SQIs are summarised in Table III. All measurements can be found in the supplementary information (S3).

TABLE III
SIGNAL QUALITY PARAMETERS EXTRACTED FROM THE ECG RECORDINGS UNDER GENTLE AMBULATORY CONDITIONS USING THE TOP WITH THE OPTIMAL DRY ELECTRODE IMPLEMENTATION. CG ARE THE COMMERCIAL GEL ELECTRODES. "FILTERED" MEANS THAT THE LOW PASS FREQUENCY IS INCREASED TO 3 HZ. TIGHT IS THE IMPLEMENTATION WITH THE ELECTRODES UNDER AN ELASTIC BAND.

	$\overline{\Delta R}$ s	σ_R s	MM	SNR*	kSQI	sSQI	hSQI
Lie CG	0.93	0.03	0.02	0.453	11.17	-2.56	5.73
Lie knit	0.93	0.03	0.01	1.15	6.82	-1.88	2.57
Sit CG	1.00	0.04	0.02	0.91	8.59	-2.18	3.74
Sit knit	1.00	0.1	0.07	0.05	5.35	-1.36	1.46
Stand CG	0.96	0.03	0.02	0.428	8.46	-2.12	3.6
Stand knit	0.98	0.08	0.05	0.044	5.44	-1.27	1.38
Walk CG	0.93	0.14	0.05	0.053	6.88	-1.67	2.31
Walk knit	1.11	0.31	0.33	0.005	3.56	0.24	0.17
Walk knit filtered	0.93	0.12	0.09	0.026	8.45	-1.43	2.42
Walk tight knit	1.03	0.14	0.03	0.227	8.76	-1.93	3.38
Stairs up tight knit	0.88	0.07	0.08	0.193	6.86	-1.47	2.02
Stairs down tight knit	0.78	0.03	0.07	0.253	6.64	-1.21	1.6

The table shows good static performance (lying, sitting and standing) of the dry knitted electrodes in the relaxed fit. They all predicted the same heartrate, and the performance parameters were within the selection criteria. When walking, standard deviations on R peaks increased both in the knitted as well as the commercial electrodes (arms were kept still during the gentle walk). While the performance parameters for the CG electrodes were still within the acceptable range, it is clear from the higher value of $\overline{\Delta R}$ that R peaks are missed in the knitted implementation. However, increasing the low frequency of the band pass to 3 Hz recovers the heart rate and other parameters in the median ECG signal.

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Thus, under gentle ambulation and with minimal signal processing the ECG recordings from our relaxed fitting garment allow extracting relevant physiological parameters. In the next step, the ambulation was increased by walking up and down the stairs. The time taken to walk up/down the stairs was ~20 s each. Thus, the analysis of movement on the stairs was done for recordings of 20 s only. With this dynamic activity, the ECG signals deteriorated such that simple signal processing was insufficient to extract physiological parameters.

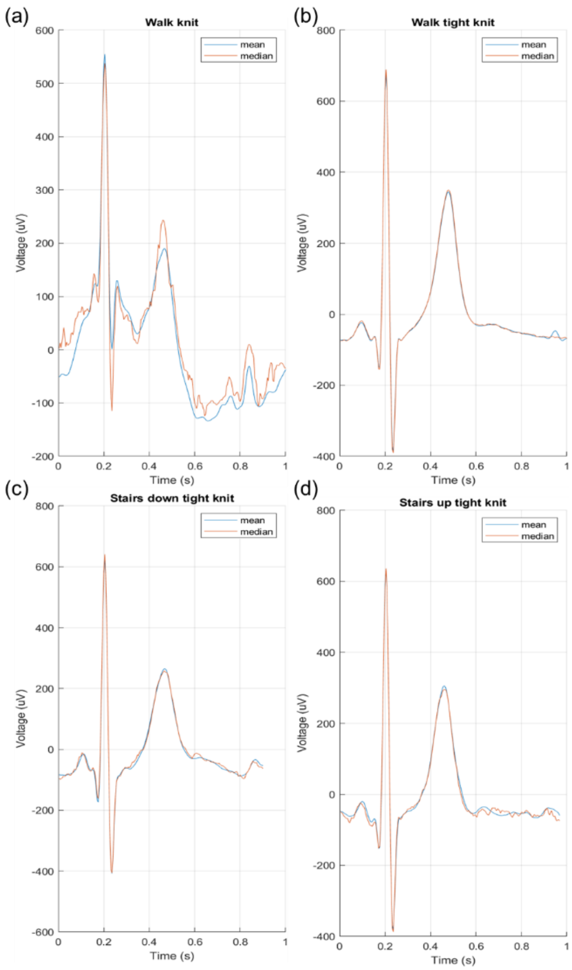

Fig. 6. Mean and median of the ECG recordings under gentle motion. The top two figures are for a gentle walk, (a) relaxed and (b) tight fit. The total recording time was ~2 min. (c) and (d) are for the tight fit for walking up and down the stairs, respectively

Although the study focuses on relaxed garments, for completeness, the pressure between the electrodes and the skin was increased by using an elastic band on the border of the knit, increasing contact pressure. The SQIs of these recordings are also included in Table III and demonstrate that dry knitted electrodes can be used under increased ambulation when the tightness of the fit is increased.

Fig. 6 compares the mean and median plot for the gentle walk wearing the relaxed and tight fit and for the walk up and down the stairs using the tight fit only. The histograms can be found in the supplementary information (S3).

VI. Discussion

In this work we presented different signal quality indices (SQIs) to compare the performance of a multitude of dry knitted electrodes. This offers a universal way to compare the ECG signals. We found that knitted electrodes in pure conductive yarn outperform those with mixed yarn (hybrid electrodes) giving lower noise due to smaller variation in contact character and higher signal amplitudes due to a higher number of skin-electrode contact points. This is helped by a high stitch density that gives small electrode resistance (large $\widetilde{RS}$) because of good inter-stitch connectivity and good noise performance (small *MM* and large *SNR*) because of small deformability (limited relative movement of neighbouring stitches). In our implementation we controlled stitch density by adapting the needle size to the thickness of the yarn (gauge). We confirm that roughness of the electrode improves contact quality. However, we demonstrate a limit to acceptable roughness beyond which the ECG signals deteriorate as shown by the noise related SQIs. In our implementation the roughness is controlled by increasing the thickness of the yarn. This roughness, offered by the rib side of the jersey knit, potentially reduces the motion artifacts due to breathing and muscle movement by resisting electrode movement with respect to the skin.

An important aspect in dry contact optimisation is their integration into a garment. Three possible approaches were investigated: intarsia, double knit and separately knitted electrodes. Of these, the electrodes knitted separately and then added by sewing them only with their corners into the garment was found to be best. It is suggested that this approach allowed the garment some movement with respect to the electrodes. This approach outperformed the intarsia and double-knit implementations in ECG signal quality. The disadvantage of separate electrodes sewn into the garment is that they need to be removed before garment recycling. The intarsia method would circumvent this. However, intarsia would require a protective front layer. The double-knit structures that are equivalent to such implementation show extensive motion artifacts, making them less suitable for relaxed garment implementation.

Recordings made on a healthy volunteer showed excellent reliability of the garment and electrode implementation for static positions: lying down, sitting and standing. For a gentle walk, motion artifacts begin to disrupt the signal, but physiological parameters can still be recovered by changing the low pass frequency of the bandpass filter.

Initial tests on volunteers with chest sizes between 70 cm and 85 cm showed that good ECG recordings can be obtained with the same knitted garment for recordings in static positions (results embargoed). This demonstrates the robustness of the implementation.

VII. Conclusion

In this work, we have optimised the structure, material and implementation of dry knitted ECG electrodes in a relaxed fitting garment to give high quality ECG signals under gentle ambulation. Our optimisation process is guided by the analysis of the ECG signals via signal quality indices (SQIs).

This works shows that using SQIs in design optimisation offers a universal technique to improve health monitoring garments. More relaxed e-garments offer great opportunities for health monitoring of the elderly and infants where only gentle ambulation needs to be considered.

References

[1] Noble R.J., Hillis J.S., and Rothbaum D.A., Chapter 33 Electrocardiography from A Clinical Methods: The History, Physical, and Laboratory Examinations. 3rd ed. https://www.ncbi.nlm.nih.gov/books/NBK354/

[2] Su, L., Borov, S. & Zrenner, B. 12-lead Holter electrocardiography. *Herzschr Elektrophys* **24,** 92–96 (2013). https://doi.org/10.1007/s00399-013-0268-4

[3] G.B. Moody, R.G. Mark, "Development and evaluation of a 2-lead ECG analysis program", Computers in Cardiology, pp. 39-44 (1982).

[4] H.L. Kennedy, "The evolution of ambulatory monitoring", Progress in Cardiovascular diseases 56(2), pp.127-132 (2013) https://doi.org/10.1016/j.pcad.2013.08.005

[5] P. Kamga, R. Mostafa, and S. Zafar, "The Use of Wearable ECG Devices in the Clinical Setting: a Review", Curr Emerg Hosp Med Rep (2022). https://doi.org/10.1007/s40138-022-00248-x

[6] Pérez-Riera A.R., Barbosa-Barros R., Daminello-Raimundo R., de Abreu L.C., Main artifacts in electrocardiography, 23(2), e12494 (2018) https://doi.org/10.1111/anec.12494

[7] Soroudi A., Hernández N., Berglin L., Nierstrasz V.,Electrode placement in electrocardiography smart garments: A review, *J. Electrocardio.* **2019**, 57, pp. 27-30. doi.org/10.1016/j.jelectrocard.2019.08.015.

[8] Private conversation with medical practitioners from the London Lung and Heart Institute

[9] L. Euler, L. Guo, N.-K. Persson, "Textile Electrodes: Influence of Knitting Construction and Pressure on the Contact Impedance", Sensors 21, 1578 (2021). https://doi.org/10.3390/s21051578

[10] Fobelets, K., Panteli, C., and Hammour, G., Simultaneous breathing and ECG measurements using e-knits, Engineering Proceedings (2023), submitted

[11] A.B. Nigusse, D.A. Mengistie, B. Malengier, G.B. Tseghai, L.V. Langenhove, "Wearable Smart Textiles for Long-Term Electrocardiography Monitoring – A Review", Sensors 2021, 21, 4174. https://doi.org/10.3390/s21124174

[12] G. Acar, O. Ozturk, A.J. Golparvar, T.A. Elboshra, K. Böhringer and M.K. Yapici: "Wearable and Flexible Textile Electrodes for Biopotential Signal Monitoring: A review", Electronics 2019, 8, 479 pp. 25 - doi:10.3390/electronics8050479

[13] G. Cho, K. Jeong, M.J. Paik, Y. Kwun, and M. Sung, "Performance Evaluation of Textile-Based Electrodes and Motion Sensors for Smart Clothing", IEEE Sensors Journal 11(12), pp. 3183-3193, (2011).

[14] Nigusse, A.B.; Malengier, B.; Mengistie, D.A.; Tseghai, G.B.; Van Langenhove, L. Development of Washable Silver Printed Textile Electrodes for Long-Term ECG Monitoring. *Sensors* **2020**, *20*, 6233. https://doi.org/10.3390/s20216233

[15] X. An and G. K. Stylios, "A Hybrid Textile Electrode for Electrocardiogram (ECG) Measurement and Motion Tracking", Materials 2018, 11, 1887; doi:10.3390/ma11101887

[16] R. Paradiso, G. Loriga and N. Taccini, "A wearable health care system based on knitted integrated sensors," in *IEEE Transactions on Information Technology in Biomedicine*, vol. 9, no. 3, pp. 337-344, Sept. 2005, doi: 10.1109/TITB.2005.854512.

[17] Zheng, J.W., Zhang, Z.B., Wu, T.H. et al. A wearable mobihealth care system supporting real-time diagnosis and alarm. Med Bio Eng Comput **45**, 877–885 (2007). https://doi.org/10.1007/s11517-007-0221-y

[18] Mestrovic, M. A., Helmer, R. J. N., Kyratzis L., and Kumar, D., Preliminary study of dry knitted fabric electrodes for physiological monitoring, *2007 3rd International Conference on Intelligent Sensors, Sensor Networks and Information*, 2007, pp. 601-606, doi: 10.1109/ISSNIP.2007.4496911.

[19] Fobelets K., Knitted coils as breathing sensors, *Sensors and Actuators A* **2020**, 306, 111945, https://doi.org/10.1016/j.sna.2020.111945.

[20] Fobelets K., Panteli C. and Hammour G., "e-knits for biosignal recording", e-textiles conference, Nottingham, UK, 8-10 Nov (2022).

[21] Shieldex® 235/36 x2 HCB – Shieldex® – Metallized Technical Textiles

[22] see e.g. Stainless Steel Medium Conductive Thread 3 Ply 18m (60ft) - Adafruit | CPC UK (farnell.com)

[23] Ismar, E., Zaman, S.u., Tao, X. *et al.* Effect of Water and Chemical Stresses on the Silver Coated Polyamide Yarns. *Fibers Polym* **20**, 2604–2610 (2019). https://doi.org/10.1007/s12221-019-9266-4

[24] OpenBCI: https://openbci.com/

[25] MATLAB - MathWorks - MATLAB & Simulink: https://uk.mathworks.com/products/matlab.html

[26] S. Rahman, C. Karmakar, I. Natgunanathan, J. Yearwood, M. Palaniswami: "Robustness of electrocardiogram signal quality indices.", J. R. Soc. Interface 19: 20220012 (2022). https://doi.org/10.1098/rsif.2022.0012

[27] Nardelli, M., Lanata, A., Valenza, G., Felici, M., Baragli, P., Scilingo, and E.P., A tool for the real-time evaluation of ECG signal quality and activity: Application to submaximal treadmill test in horses Biomedical Signal Processing and Control 56, 101666 (pp.7) (2020). https://doi.org/10.1016/j.bspc.2019.101666

[28] Le K. , Narayana H., Servati A., Bahi A., Soltanian S., Servati P. and Ko F., Electronic textiles for electrocardiogram monitoring: a review on the structure–property and performance evaluation from fiber to fabric, Textile Research Journal 2022, 0(0) 1–33 DOI: 10.1177/00405175221108208

[29] Shieldex® Med-tex P130 – Shieldex® – Metallized Technical Textiles

[30] B.E.Jin, H. Wulff, J.H. Widdicombe, J. Zheng, D.M. Bers, J.L. Puglisi, "A simple device to illustrate the Einthoven triangle." Adv Physiol Educ., 36(4):319-24, Dec 2012. doi: 10.1152/advan.00029.2012.

[31] H. Cho & J.H. Lee, "A Study on the Optimal Positions of ECG Electrodes in a Garment for the Design of ECG-Monitoring Clothing for Male.", J Med Syst (2015) 39: 95 DOI 10.1007/s10916-015-0279-2

[32] P.L. Finka, A.S.M. Sayema, S.H. Teay, F. Ahmad, H. Shahariar, A. Albarbar, "Development and wearer trial of ECG-garment with textile-based dry electrodes.", Sensors and Actuators A 328, 112784, (2021).

[33] K. Yamada, "Body surface isopotential map: Past, Present and Future.", Japanese circulation journal 45(1) pp. 14, (1981) https://doi.org/10.1253/jcj.45.1