Capacitive Sensing of Electrocardiographic Potential Through Cloth From the Dorsal Surface of the Body in a Supine Position: A Preliminary Study

Akinori Ueno, Yasunao Akabane, Tsuyoshi Kato, Hiroshi Hoshino, Sachiyo Kataoka, and Yoji Ishiyama

Abstract—A method for obtaining electrocardiographic potential through thin cloth inserted between the measuring electrodes and the skin of a subject’s dorsal surface when lying supine has been proposed. The method is based on capacitive coupling involving the electrode, the cloth, and the skin. Examination of a pilot device which employed the method revealed the following: 1) In spite of the gain attenuation in the high frequency region, the proposed method was considered useful for monitoring electrocardiogram (ECG) for nondiagnostic purpose. 2) Electrode area was the factor that had most influence on the signal, compared with other factors such as cloth, and the skin. Examination of a pilot device which employed the insulating textile electrodes in a bed set. Park [8] proposed a prototype intelligent garment that can collect, process, store, and transmit information about the wearer, such as ECG parameters, and Catrysse [9], Scilingo [10], and Paradiso [11] have implemented this idea by using textile or fabric sensors. Since a noninvasive and awareness-free approach is essential for long-term home monitoring, the idea of sensors embedded in furnishings or daily clothing is quite rational. However, there are still some challenges to be addressed regarding the utility of such systems.

In conventional ECG measurement, an electrolytic paste or a conductive adhesive is almost always required for maintaining reliable ohmic contact with the skin. Therefore, ECG measurement for a long period using conventional methods causes irritation and discomfort, and is a potential cause of skin allergy and inflammation. Despite these disadvantages of conventional ECG detection, sufficient consideration has not been given to skin-to-electrode coupling [8]. Previous authors [6], [7], [9]–[11] have addressed this problem by employing textile electrodes, which do not require any electrolytic paste or conductive adhesive for measurement, to relieve potential irritation and discomfort. Nevertheless, the potential for metal allergy or discomfort upon direct long-term contact of the skin with metal material has not yet been addressed.

With the above background in mind, the authors have focused on the principle of capacitive sensing capable of detecting alternating electrical potential through an inserted thin insulator, and have applied the principle for measurement of electrocardiographic potential through commonly available cloth from the dorsal surface of the body when the subject is supine. In order to explore the feasibility and applicability of this idea, the authors have fabricated a pilot measuring device and conducted some preliminary experiments and model studies using it.

II. CAPACITIVE SENSING

Capacitive sensing is based on the principle of the capacitive (or insulated) electrode [12]–[18], which is an electrode not requiring the use of electrolytic paste or conductive adhesive. The electrode relies on capacitive coupling, conventionally involving a metal electrode, an insulator, and the skin, as shown in Fig. 1(a). The electrode can carry an alternating bioelectric current equivalently through the capacitance of the coupling. In previous studies, various insulators, such as anodized aluminum [12], [13], silicon dioxide [14], [15], pyre varnish [16], anodic insulated tantalum oxide [17], or barium titanate [18], have been tested as these materials exhibit high permittivity. In the present study, conventional systems for ECG monitoring are unsuitable for long-term monitoring at home over periods of weeks or even months. Since conventional methods for ECG monitoring are unsuitable for long-term studies, previous researchers have focused on developing dedicated systems for long-term monitoring in daily life [2]–[11]. For example, Kwatra [2], [3], Ishijima [4], and Tamura [5] have developed systems for ECG or heart rate monitoring during bathing. Ishijima [6], [7] developed an automated system for ECG monitoring during sleep using embedded textile electrodes in a bed set. Park [8] proposed a prototype intelligent garment that can collect, process, store, and transmit information about the wearer, such as ECG parameters, and Catrysse [9], Scilingo [10], and Paradiso [11] have implemented this idea by using textile or fabric sensors. Since a noninvasive and awareness-free approach is essential for long-term home monitoring, the idea of sensors embedded in furnishings or daily clothing is quite rational. However, there are still some challenges to be addressed regarding the utility of such systems.

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Capacitive sensing is also based on an impedance matching circuit to mediate the high impedance of the coupling with a low impedance required by the subsequent circuitry. For example, a field-effect transistor...
Fig. 1. Schematic models of (a) conventional and (b) modified capacitive electrodes coupled to the skin. \( \varepsilon_a \) and \( \varepsilon_b \) indicate dielectric constants of the inserted insulator and the cloth, respectively. \( \varepsilon_0 \) is the permittivity of free space.

(FET) source follower has been used as an impedance matching circuit and actively mounted in an electrode housing [15], [16] for reducing common mode interference caused by power lines. In the present study, an instrumentation amplifier [19] composed of operational amplifiers with high input impedance (National Semiconductor, LF356, 1000 \( \Omega \)) according to the specification sheet\(^1\) was employed in order to match the greater impedance of the coupling due to the smaller dielectric constant of the inserted cloth in comparison with conventional insulators. The impedance matching circuit was not mounted in the electrode housing to avoid hard contact with the skin.

III. MATERIALS AND METHODS

A. Fabric Electrode

The electrocardiographic signal is picked up by electrodes made of a square sheet of conductive fabric with conductive acrylic adhesive (3M, 2191FR). The electrodes are stuck to a mattress with the adhesive and covered by a cotton bed sheet. Two lead electrodes are located under the right and left scapulae of the subject when lying in a supine position, as shown in Fig. 2. A reference electrode is placed beneath the right lumbar region. Skin-cloth-electrode coupling is held by the subject’s weight on the cloth and by repulsive force from the mattress. The lead electrodes are connected to a measuring device, as described in the next section, by using shielded wires 2.0 m long. Three sets of measuring electrodes, each with an area of 70, 40, or 10 \( \text{cm}^2 \), were prepared for examining the influence of the electrode area on an obtained signal.

B. Signal Extraction

The pilot measuring device with filtering and amplification circuitry was fabricated using off-the-shelf components. The device consists of an instrumentation amplifier, a high-pass filter, two notch filters, a low-pass filter and an inverting amplifier. A block diagram of the device is shown in Fig. 2. The instrumentation amplifier is employed not only as a differential amplifier but also as the impedance matching circuit. The circuit elements of the high-pass filter and the low-pass filter are determined in order to obtain a cutoff frequency of 0.05 and 100 Hz, respectively. A Butterworth filter (NF, CF-2BE-HB-50Hz-Q5) is used for the former notch filter in order to reduce 50-Hz interference. The device is powered by four series D batteries to obviate the possibility of electric shock. A dc-dc converter (Datel, BST-12/105-d5)\(^2\) is used to boost the nominal 6.0 V supply to a regulated ±12 V.

C. Simultaneous Measurement With a Conventional Device

One male subject, aged 22, naked from the waist up, was instructed to lie supine on the mattress bearing the electrodes with an area of 70 \( \text{cm}^2 \). A cotton bed sheet 395 \( \mu \text{m} \) thick was used to cover the electrodes and the mattress. Electrocardiographic potential was recorded using the developed device from the dorsum of the subject through the inserted cotton sheet. The output signal from the device was digitized at 1 kHz by an analog-to-digital converter and stored in a personal computer using a data acquisition system (Biopac Systems, MP-150 system). As a reference signal, the ECG was recorded using a conventional bioamplifier (Teac Instruments, BA1104-CC) and disposable electrodes (Nihon Kohden, Vitrode F-150M).\(^3\) The disposable electrodes were attached directly to the dorsal surface of the subject at positions close to each electrode on the mattress. As another reference, the Lead I ECG was measured using a standard limb lead with the same conventional bioamplifier. Both reference signals were wirelessly transmitted and received with a telemeter unit (Teac Instruments, TU-4), and then digitized simultaneously with the output signal of the developed device.


\(^2\)[Online]. Available at http://www.datel.com/data/power/bst-3w.pdf.

D. Experimental Long-Term Monitoring Using the Device

In order to examine a stability of the capacitive coupling of our device, an experiment to examine long-term monitoring was conducted using the same volunteer. The subject, naked from the waist up, was instructed to lie supine at rest for 7 h on the mattress, which was covered by the same cloth, in the same manner as that described in Section III-C. An initial measurement was conducted for 15 min, and then every hour thereafter.

E. Influence of Coupling Condition on the Obtained Signal

If the coupling between an electrode and the skin can be modeled as in Fig. 1(b), the capacitance $C$ of a coupling having a distance of $d$ meters, a coupling area of $S$ square meters, and a coupling dielectric constant $\varepsilon$ is given by

$$C = \frac{\varepsilon S}{d}. \quad (1)$$

Since any change in the capacitance of the coupling is related to a change in impedance (i.e. voltage loss) at the coupling, the output signal of the developed device is considered to be affected by the area of the coupling (i.e. electrode area), the distance between the electrode and the skin, and the dielectric constant of the coupling. Then, the influence of coupling condition on the output signal can be examined by changing the electrode area, the thickness of the inserted cloth, and the pressure on the coupling.

1) Influence of Electrode Area: Measuring electrodes having an identical area of 70, 40, or 10 cm$^2$ were placed on the mattress according to the configuration shown in Fig. 2. The electrodes were covered with the cotton sheet 395 mm thick. Then, three synthetic human dorsal surfaces made of square pieces of conductive fabric having the same area of the measuring electrode were attached to the cloth at locations just above each of the measuring electrodes, so that the whole area of each measuring electrode was involved in capacitive coupling with each synthetic dorsal surface. The electrodes, the sheet, and the synthetic dorsal surfaces were subjected to 1751 Pa pressure using a weight to mimic the pressure resulting from a supine adult subject 1.65 m in height and weighing 60 kg (dorsal area 0.336 m$^2$). The simulated ECG signal was input from an ECG generator (Nihon Kohden, AX-201D) to the synthetic dorsal surfaces. The output signal from the device, and also the frequency-gain response from 0.01 to 500 Hz, was measured for each of the different electrode areas.

2) Influence of Thickness of the Inserted Cloth: Cotton sheets with different thicknesses of 395, 463, or 1020 $\mu$m at atmospheric pressure were inserted between the electrode and the skin, and the output signal from the device was measured for each thickness. Three 70-cm$^2$ electrodes, three synthetic dorsal surfaces, the ECG generator and a pressure of 1751 Pa were used in a similar way to that in Section III-E1. Also, the frequency responses were measured respectively for each sheet thickness.

3) Influence of Pressure on the Coupling: The simulated ECG and the frequency response for each thickness of cotton sheet were measured at a pressure of 1751, 1083, or 578 Pa. The 70-cm$^2$ electrodes, the synthetic dorsal surfaces and the ECG generator were used in a similar way to that described in Section III-E1.

F. Frequency-Impedance Response of the Inserted Cloth

In order to investigate the interaction between the electrical properties of the inserted cloth and the obtained signal, the frequency-impedance response was measured for each cloth with a precision LCR meter (Agilent Technologies, 4284A). The cloth was fixed at a constant pressure by a dielectric test fixture (Agilent Technologies, 16451B). The frequency was varied from 20 Hz, which is the lower limit of the instrument, to 300 Hz. To complement the deficit of the raw data below 20 Hz, we used the model of capacitively coupled cloth shown in Fig. 3, and then estimated the model parameters for each cloth by fitting the following equation to the measured impedances of the cloth, $Z_{\text{cloth}}$:

$$|Z_{\text{cloth}}(f)| = R_{cn} + \frac{R_{cl}}{\sqrt{1 + (2\pi f C_d R_{cl})^2}} \quad (2)$$

where $C_d$ and $R_{cl}$ are the capacitance and dc resistance of the cloth, $R_{cn}$ is the contact resistance between the cloth and conductive surfaces, and $f$ is the frequency.

G. Input Capacitance of the Developed Device

We expected that the input capacitance of the device would eventually degrade the performance in the high frequency region. Since the input capacitance was too low to be measured with conventional equipment, we devised a method for assessing it. This procedure was performed as follows. First, two off-the-shelf capacitors with the same capacitance were connected at the front ends of the differential input of the device instead of the two lead electrodes, and the frequency response while the capacitors were inserted was measured. Then, the gain in decibels at each frequency with the capacitors inserted was subtracted from the corresponding original gain measured without the capacitors. Second, the developed device connected to the two capacitors was modeled by an equivalent circuit, as shown in Fig. 4. In the figure, the gain of the device with the capacitors, $G_{\text{cap}}(f) = V_{\text{out}}/V_{\text{in}}$, can be expressed using the original gain of the device, $G_{\text{org}}(f) = V_{\text{out}}/V_{\text{in}}$, and circuit components in Fig. 4, as follows:

$$G_{\text{cap}}(f) = \frac{j2\pi f R_{in} C_0}{2 + j2\pi f R_{in} (2C_{in} + C_0)} G_{\text{org}}(f) \quad (3)$$

where $C_0$ is the capacitance of the externally inserted capacitors, and $C_{in}$ and $R_{in}$ are the input capacitance and input resistance of the device, respectively. According to (3), the gain difference in decibels between 20 log $|G_{\text{cap}}|$ and 20 log $|G_{\text{org}}|$, i.e. 20 log ($|G_{\text{cap}}|/|G_{\text{org}}|$), can be described as follows:

$$20 \log \left[\frac{|G_{\text{cap}}(f)|}{|G_{\text{org}}(f)|}\right] = 20 \log \frac{2\pi f R_{in} C_0}{\sqrt{1 + 4\pi^2 f^2 R_{in}^2 (2C_{in} + C_0)^2}}. \quad (4)$$

![Fig. 3. Equivalent circuit of the capacitively coupled cloth. $R_{cn}$ is the contact resistance between the cloth and the conductive surfaces such as electrode or skin. $R_{cl}$ and $C_d$ are dc resistance and capacitance of the cloth, respectively.](image-url)
Fig. 4. Equivalent circuit of the input part of the developed device in which capacitors are inserted at the front ends. $C_0$ is the capacitance of the inserted capacitor. $C_{in}$ and $R_{in}$ are the input capacitance and input resistance of the device, respectively.

Fig. 5. Recordings obtained simultaneously using the present device with a cotton sheet 395 μm thick from the dorsum (top), using a conventional bioamplifier without a cloth from the dorsum (middle), and using a conventional bioamplifier and a standard limb lead without a cloth (bottom). The top recording was subjected to a 20-point moving average to eliminate power line noise.

Since the value of $C_0$ is known and $R_{in}$ can be obtained from the specification sheet of the IC used in the device, we can consider (4) as a function of $f$ and $C_{in}$. Therefore, we can estimate $C_{in}$ by fitting (4) to the experimentally obtained gain difference between $20 \log |G_{ref}|$ and $20 \log |G_{det}|$.

In the actual experiment, ceramic 102-pF or 478-pF capacitors were connected individually via the lead wires, and the two frequency responses were measured. Moreover, frequency responses not including the wires were also measured. The capacitance of the capacitors was measured with the precision LCR meter.

IV. RESULTS

A. Simultaneous Measurement With a Conventional Device

Fig. 5 shows recordings typical of those obtained. The output signal of the developed device showed periodic waveforms synchronized to, and similar to, the reference ECGs. Therefore, the proposed method was considered useful at least for detecting heart rate and/or other periodic parameters of the ECG even with the cloth inserted. However, the ratios of the peak amplitude of the P-wave and T-wave to the R-wave in the top recording appeared to be slightly greater than those in the reference ECGs. This distortion was probably due to inconstant gain through the pass band of our device when the cloth was inserted.

B. Experimental Long-Term Monitoring Using the Device

As can be seen in the top recording in Fig. 6, depressed R-waves and skewed T-waves were observed at the beginning of the experiment. However, they had disappeared before measurement was conducted at the 1-h point. In addition, there was a marked decrease in hum noise in recordings after 4 h, approaching a level as low as that in the directly measured ECG.

Fig. 7 shows typical recordings contaminated by motion artifacts during long-term measurement. (a) Recording made at the beginning, and (b) 2 h later. The top recording in each figure is that obtained using the present device from the dorsum with a cotton sheet 395 μm thick inserted, and the bottom recording is the wirelessly measured lead I ECG.
This phenomenon was often observed in recordings after 2 h. Therefore, there was no significant adverse effect of long-term measurement on the quality of the output signal from the developed device.

C. Influence of Coupling Condition on the Output Signal

1) Influence of Electrode Area: As shown in Fig. 8(a), the periodic R-wave was evident in all recordings, even for the smallest electrode area of 10 cm$^2$. However, the waveform of the PQRST complex became distorted as the electrode area decreased. Frequency responses in Fig. 8(a) suggests that this distortion was attributable to a tilted frequency-gain response in the 0.1–100 Hz region. Since the gain in the high-frequency domain decreased as the area of the electrode decreased, the R-wave was depressed as the area decreased and the S-wave was absent in the recording obtained with the 10 cm$^2$ electrodes. Therefore, the area can be reduced to 10 cm$^2$ if only heart rate is being detected.
TABLE I

<table>
<thead>
<tr>
<th>Thickness [μm]</th>
<th>(R_{cm} [\Omega])</th>
<th>(R_{cd} [\Omega])</th>
<th>(C_{cd} [\text{pF}])</th>
<th>Relative Permittivity</th>
</tr>
</thead>
<tbody>
<tr>
<td>395</td>
<td>1.93</td>
<td>209</td>
<td>59</td>
<td>2.3</td>
</tr>
<tr>
<td>463</td>
<td>3.96</td>
<td>152</td>
<td>71</td>
<td>3.3</td>
</tr>
<tr>
<td>1020</td>
<td>8.23</td>
<td>121</td>
<td>39</td>
<td>4.0</td>
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</tbody>
</table>

2) Influence of Thickness of the Inserted Cloth: As can be seen in Fig. 8(b), the PQRST complex was sufficiently recognizable even with a cloth thickness of 1020 μm, and the quality was similar to that of other recordings obtained with thinner cloth. Although the gain had a tendency to decrease as the frequency increased, all the frequency responses obtained with the cloth indicated a close gain for each frequency. In fact the responses obtained with the 463 μm cloth and the 1020 μm cloth were almost the same. This may be because the coupling impedance of these cloths is almost the same at a pressure of 1751 Pa. On the other hand, the least gain was obtained for the 395 μm cloth, despite the fact that this was the thinnest. This minimal gain for the 395 μm cloth was presumably due to the low dielectric constant and the high resistivity of the cloth. Thus, it appears that the thickness of the inserted cloth at atmospheric pressure can be increased to at least 1000 μm as long as the electrode is coupled firmly with sufficient pressure.

3) Influence of Pressure on the Coupling: As indicated in Fig. 8(c), reduction of the pressure from 1751 to 578 Pa had little influence on both the S/N and the waveform of the recordings for the 395 μm cloth. The frequency responses, shown in Fig. 8(c), agreed with the results. With regard to the other two cloths, little effect was also confirmed. Moreover, the R-wave could be recognized for each cloth thickness even when no weight was placed on the electrodes. Thus, the developed device shows promise for application not only to adult subjects, but also to neonates for heart rate monitoring.

4) Influence of Electrode Conductivity: No gain attenuation of the electrodes was confirmed from 0.01 to 500 Hz. Therefore, the gain attenuation observed in Fig. 8 was not due to gain loss at the electrode but to that at the coupling involving the cloth.

D. Frequency-Impedance Response of the Inserted Cloth

In accordance with the results shown in Fig. 9, impedance of the 395 μm cloth was estimated to be the highest in a frequency region below 10 Hz. Table I indicates this was due to the largest dc resistance and the least relative permittivity of the cloth. Thus, the least gain for the 395 μm cloth in Fig. 8(b) can be attributed to these electrical properties of the cloth.

478 μF, with the wires
478 μF, without the wires
102 μF, with the wires
102 μF, without the wires

Fig. 10. Frequency response of (a) the gain with capacitors inserted at the front ends of the system and (b) the difference in gain between (a) and the original response. The original frequency response is that measured directly from the electrodes. The solid and dotted lines in (b) are the best fitted curves obtained by applying (4).

E. Input Capacitance of the Developed Device

As shown in Fig. 10, the plots obtained when the lead wires were included were apparently smaller than those excluding the wires. As a consequence, \(C_{in}\) of 370 and 65 pF indicated the least root mean square error for the results including and excluding the wires, respectively. Thus, the lead wires had an input capacitance of 305 pF.

V. DISCUSSION

A. Perspiration of the Subject

In a previous article, Geddes [20] reported that “dry” electrodes are really only dry when first applied because of subsequent perspiration, and that impedance at 20 min was between one-fourth and one-fifth of the initial impedance. Therefore, the improvement of the S/N with time in Fig. 6 was probably due to a decrease in both the impedance of the cloth and the contact resistance between the cloth and the skin, originating from the perspiration of the subject. In long-term measurement,
device with cloth inserted at the front ends can be modeled by substituting $C_0$ in Fig. 4 for $Z_{cloth}$ in Fig. 3, the gain with the cloth inserted, $G_{cloth}(f)$, can be described as follows:

$$G_{cloth}(f) = \frac{(j4\pi f C_{in} + 2/R_{in})^{-1} \cdot G_{eq}(f)}{R_{cn} + (j2\pi f C_{cl} + 1/R_{cl})^{-1} + (j4\pi f C_{in} + 2/R_{in})^{-1}}.$$  \hfill (5)  

We can estimate the frequency-gain responses with the cloth inserted by substituting the parameter values in (5) and by using the experimentally obtained $G_{eq}(f)$. As shown in Fig. 11, the simulated response when $C_{in}$ equals 370 pF exhibited a characteristic similar to the experimentally obtained responses shown in Fig. 8. Moreover, the simulation when $C_{in}$ equals 65 pF predicted that gain attenuation in the high-frequency regions could be suppressed by decreasing the input capacitance. Therefore, it is necessary to reduce the input capacitance, for example by shortening the wire length between the electrodes and the impedance matching circuits, in order to acquire a less distorted waveform and to make the device applicable for diagnostic use.

VI. CONCLUSION

We have proposed an approach for obtaining electrocardiographic potential from the dorsal surface of a subject lying supine under conditions whereby a thin cloth is inserted between the electrodes and the skin. We fabricated a pilot measuring device and explored the feasibility and applicable scope of the proposed method. Examination of the device yielded the following results.

1) In spite of the gain attenuation in the high frequency region, the proposed method was considered useful for monitoring ECG for nondiagnostic purposes.

2) The method was able to yield a stable ECG from a subject at rest for at least 7 h, and there was no significant adverse effect of long-term measurement on the quality of the signal obtained.

3) Electrode area had a greater influence on the signal than cloth thickness and coupling pressure, but could be reduced to 10 cm² for heart rate detection.

4) Input capacitance of the device was assumed to be the dominant factor for the gain attenuation in the high frequency region, and this should be reduced to make the device applicable for diagnostic use.

Although there is still room for improvement in terms of its practical use, the proposed method appears promising for application to bedding as a noninvasive and awareness-free method for ECG monitoring.

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REFERENCES


Wavelet Packets Feasibility Study for the Design of an ECG Compressor

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Abstract—Most of the recent electrocardiogram (ECG) compression approaches developed with the wavelet transform are implemented using the discrete wavelet transform. Conversely, wavelet packets (WP) are not extensively used, although they are an adaptive decomposition for representing signals. In this paper, we present a thresholding-based method to encode ECG signals using WP. The design of the compressor has been carried out according to two main goals: 1) The scheme should be simple to allow real-time implementation; 2) quality, i.e., the reconstructed signal should be as similar as possible to the original signal. The proposed scheme is versatile as far as neither QRS detection nor a priori signal information is required. As such, it can be thus applied to any ECG. Results show that WP perform efficiently and can now be considered as an alternative in ECG compression applications.

Index Terms—ECG compression, electrocardiogram, filter bank, subband coding, thresholding, wavelet coding, wavelet packets (WP), wavelet transform.

I. INTRODUCTION

The electrocardiogram (ECG) is widely used because it is a noninvasive way to establish clinical diagnosis of heart diseases. Therefore, ECG processing has been a topic of great interest and is most commonly used in applications such as monitoring. Long-term records have also become commonly used to extract or detect important information from the heart signals. In these cases, the quantity of data grows significantly and compression is required for reducing the storage required and transmission times.

A large amount of lossy ECG compression methods have been developed using many different techniques. In [1], a classification of compression schemes into three categories is proposed: Direct methods, transform methods, and other compression methods. Most of the methods classified in the second group are based on the wavelet transform (WT) and multiresolution analysis, enabling them to obtain good compression ratios (CRs) and real-time implementation. The set partitioning in hierarchical trees (SPIHT) algorithm [2] is the most widely known algorithm that fulfills these features. Subsequently, it has been surpassed by other approaches [3], [4] that give high CR, with the tradeoff being higher computational complexity.

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