Technical note

High-pass filtering of ECG signals using QRS elimination

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1 Introduction

THE AUTOMATIC or visual measurement and interpretation of the electrocardiogram depends strongly on the quality of the memorised or recorded signals. The absence of mains interference and baseline drift are the two main conditions for a high-quality signal. The 50 Hz interference has been successfully eliminated by an adaptive subtraction procedure (LEVKOV *et al.*, 1984), further improved to be applicable even for real-time processing (CHRISTOV and DOTSINSKY, 1988).

The baseline drift is a less disturbing interference, except in stress-test ECG, but more difficult to eliminate or even suppress. Inadequate analogue high-pass filtering has been known to introduce distortions in the ST segment (BERSON and PIPBERGER, 1965), therefore the high-pass time constant of 3·2 s has been universally accepted in diagnostic ECG instrumentation. An example is shown in Fig. 1*a*, comparing 'QRS-like' triangular test signals after 0·3 s and 3·2 s high-pass time constant filtering. Another example using real ECG (Fig. 1*b*) shows an elevation of the initial part of the ST-segment (best seen as a reduced falling part of the R-wave) and a distortion at the end of the T-wave, applying the same 0·3 s and 3·2 s time-constant filtering.

Recently, digital filtering has been successfully applied. Recursive high-pass filters are relatively fast, but their phase distortions might be unacceptable. If the filtering procedure is not to be accomplished in real time, a backward processing can be applied (two-pass filtering) to compensate for phase distortions. Nonrecursive high-pass filters do not introduce distortions and permit the achievement of a high-pass cutoff frequency up to near the heart rate (VAN ALSTÉ *et al.*, 1986), but the procedure could be rather slow, or would need special hardware. Moreover, in our opinion, the relatively high cutoff frequency will cause distortions of large monopolar QRS complexes (e.g. in cases with ventricle hypertrophy or certain types of blocks), which can be observed in some of the authors own records.

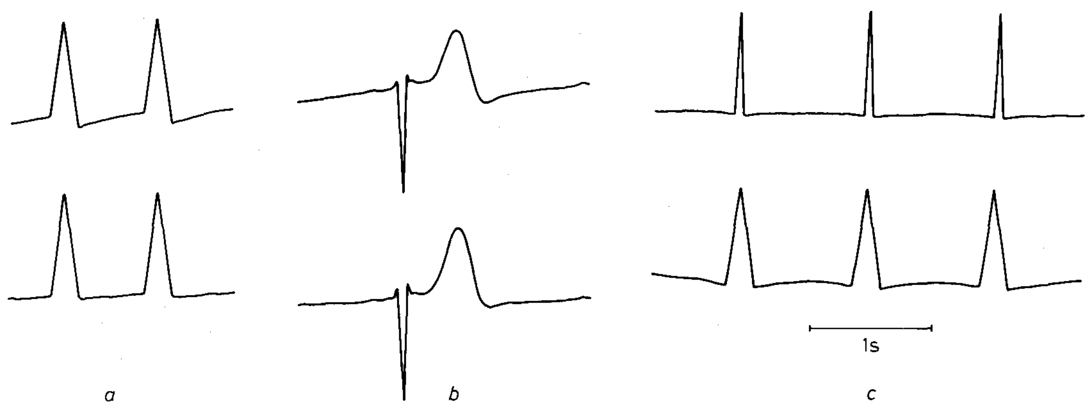

Fig. 1 *Comparison of different types of filtering. (a) Triangular 'QRS-like' test signal filtered by a first-order high-pass filter of 0·3 s time constant (upper trace) and of the standard 3·2 s time constant. (b) The same filter used on a QRS-T complex. Note the slight ST-segment elevation and the distortion of the falling part of the T-wave, due to the 0·3 s time-constant (upper trace). (c) A '100 ms R-wave' test signal (upper trace), filtered by a two-pass digital filter (forward and backward), each pass a first-order 0·64 Hz cutoff frequency filter. The same filter applied to a '250 ms R-wave' test signal (lower trace)*

Our group successfully applied linear-phase two-pass digital and hybrid filtering for drift suppression, keeping the cutoff frequency to about 0·64 Hz, to avoid the above-mentioned distortions (DASKALOV, 1988). However, with large QRS complexes, distortions can appear. An example is shown in Fig. 1*c*, where triangular test signals of 80 ms duration (a normal QRS complex) and 240 ms (as a large QRS or extrasystole) are compared. Distortions appear in the case of 240 ms signals.

The high-pass filtering of ECG signals can also be achieved by fast low-pass averaging-type filters with no phase distortions (AHLSTROM and TOMPKINS, 1985), for filtering out the drift and subtracting it from the original

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signal. This can be done with a certain short and constant delay, thus obtaining a delayed (or a quasi-real-time) filtered signal.

As mentioned above, in our opinion, the main problem in high-pass filtering distortions is the QRS complex, especially when it is monopolar, of high amplitude and large duration. We have shown that partial differentiation of such QRS complexes often occurs, resulting in an after-effect component superimposed on the initial part of the ST segment, changing its shape (Figs. 1*a* and 1*b*). This effect has often been wrongly attributed to the fact that the ST-segment is the component of lowest frequency content in the signal spectra; therefore it is the first to be distorted with high-pass filtering. In many medical ECG interpretation textbooks (e.g. WARTAK, 1975), nonischaemic ST shifts opposite to high and large monopolar QRS complexes are described as additional signs of hypertrophy, some types of blocks etc.

This effect is stronger with higher cutoff frequencies of the high-pass filters. It was studied for the purpose of ECG screening analysis using a filter time-constant of 0·5 s (DASKALOV, 1977). Time constants in the range 0·3–1·5 s are often used in bedside monitors and Holter devices.

We have attempted to solve this problem by using an analogue adaptive filter, where the high-pass cutoff frequency was shifted to lower values depending on the differentiated ECG signal. In other words, the appearance of a QRS complex (or other high slew rate component) automatically increased the high-pass time constant (DASKALOV, 1975). Unfortunately, the analogue solution was of limited efficiency, due mainly to the inherent phase distortions.

Taking into account these considerations, we propose an algorithm for high-pass filtering of the ECG signal including means for elimination of the QRS complex from the filtering procedure.

2 Algorithm

The ECG signal, excluding the QRS complexes, is low-pass filtered by an averaging filter (other types of low-pass filtering are possible, depending on the processing speed) and the signal thus obtained is subtracted from the original.

The number N of samples to be averaged depends on the sampling frequency f and the first zero-frequency of the filter F:

$$N = f/F$$

In our example we use $f = 200$ Hz, synchronised with the mains frequency and $F = 1{\cdot}6$ Hz, corresponding to a -3 dB frequency (cutoff frequency) F_c of about 0·8 Hz. Experiments with $F = 1{\cdot}2$ Hz ($F_c = 0{\cdot}6$ Hz) and $F = 0{\cdot}8$ Hz ($F_c = 0{\cdot}4$ Hz) were carried out too.

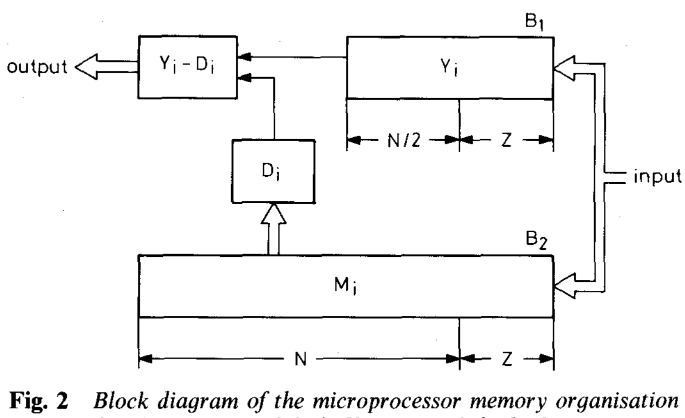

Fig. 2 *Block diagram of the microprocessor memory organisation for the purpose of drift filtering with high slew-rate wave suppression*

With $f = 200$ Hz and $F = 0{\cdot}8$ Hz, N equals 250. For convenience (faster processing), $N = 256$ was chosen (thus the exact frequencies are $F = 0{\cdot}78$ Hz and $F_c = 0{\cdot}39$ Hz). For $N = 171$, $F = 1{\cdot}2$ Hz (precisely, $F = 1{\cdot}17$ Hz) is obtained and for $N = 128$, $F = 1{\cdot}6$ Hz (precisely, $F = 1{\cdot}56$ Hz). The sampling is of 10-bit resolution, with a sensitivity of 12·5 μV per bit referred to the input.

The current input signal samples Y_i are stored in the memory, organised as two 10-bit shift-register buffers B_1 and B_2 (Fig. 2). B_1 has a length of $(N/2) + Z$ samples and B_2 is of $N + Z$ samples. The number of samples Z (the 'length' of the Z-part of the B_1 and B_2 buffers) is chosen to be capable of storing any large QRS complex or even a joined QRS-T interval. The samples Y_i are named M_i in B_2, because they may be further modified according to the algorithm.

The drift component D_i is obtained by low-pass filtering the data by averaging N values of M_i. It is then subtracted from the original signal: $Y_i - D_i$.

The suppression of the high slew-rate components of the signal (normally the QRS complexes) is done by an algorithm applied over the Z samples.

The entire algorithm is as follows.

(i) The samples are kept unchanged in B_2 ($M_i = Y_i$) as long as the absolute difference D between the current sample Y_i and the one eight conversion steps before (Y_{i-8}) remains equal to or less than a threshold value $L = 8$. This condition is met for low slew-rate signal components. The threshold used ($L = 8$ bits for eight samples, or 2·5 μV ms^{-1}) is not sensitive to 50 Hz interference, due to the fact that the interval of eight samples (40 ms) is a multiple of the 50 Hz period (20 ms). Small fluctuations between adjacent samples also do not reach the threshold.

(ii) When D becomes greater than L, the value $Y_{i-8} = M_{i-8} =$ BEG (beginning) in the zone Z of B_2 is taken as the beginning of a high slew-rate wave.

(iii) Furthermore, when D has been lower or equal to L for forty consecutive values, the value $Y_{i-40} = M_{i-40} =$ END is taken as the end of the high slew-rate wave. Thus the classification of horizontal ST elevation or depression intervals as isoelectric segments is avoided.

(iv) Finally, a linear interpolation between BEG and END is accomplished.

3 Results and discussion

The algorithm described above has been implemented on a multichannel microprocessor electrocardiograph in a real-time recording mode. As mentioned above, the sampling frequency is 200 Hz and the resolution is of 12·5 μV per bit referred to the input. This relatively low sampling frequency was selected for a real-time mode of recording three channels simultaneously, including 50 Hz interference elimination (according to the method described by CHRISTOV and DOTSINSKY, 1988) and the present drift suppression algorithm. This mode is for emergency recordings, whereas in normal operation a sampling rate of 400 Hz is applied.

The influence of high slew-rate waves suppression on the filtering error can be assessed by comparing filtering without and with suppression (Figs. 3*a* and 3*b*), using $F_c = 0{\cdot}4$ Hz. The error signals can easily be seen on the third trace—the difference between filtered and unfiltered signals. Each wave, identified by the program as a high slew-rate wave, is marked by short upward and downward directed lines (Fig. 3*b*), for illustration of the suppressed intervals. Note that the distortions appearing on the difference trace are delayed with respect to the original signal

QRS complexes due to the constant delay of the filtering procedure.

$F_c = 0{\cdot}4$ Hz proved to be distortionless for every one of our 400 recordings (Fig. 3*b*). Errors of less than 0·05 mV (difference between unfiltered and filtered signals) were obtained using $F_c = 0{\cdot}6$ Hz (Fig. 3*c*).

The selection of F_c depends on the specific application. $F_e = 0{\cdot}8$ Hz was adopted after extensive testing of more

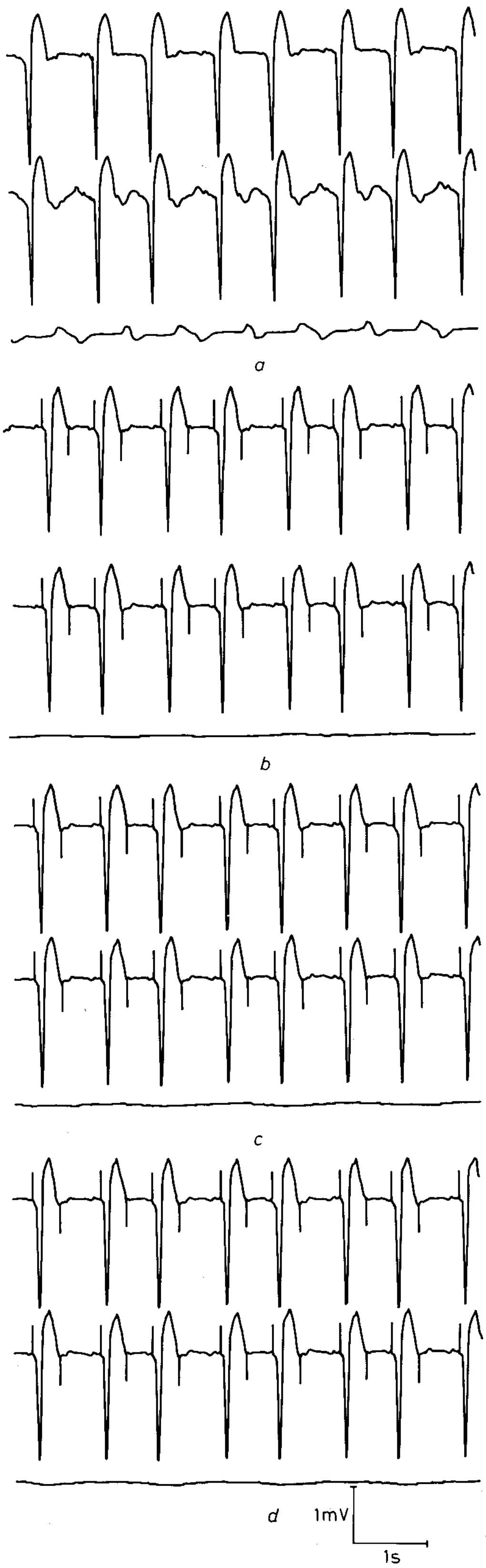

Fig. 3 *Filtering (a) without and (b–d) with high slew-rate waves suppression. Cutoff frequencies of (b) 0·4 Hz, (c) 0·6 Hz and (d) 0·8 Hz are used. First trace: unfiltered signal; second trace: filtered signal; third trace: difference between both signals. Each high slew-rate wave, identified by the program, is delineated by a pair of upward and downward directed marks*

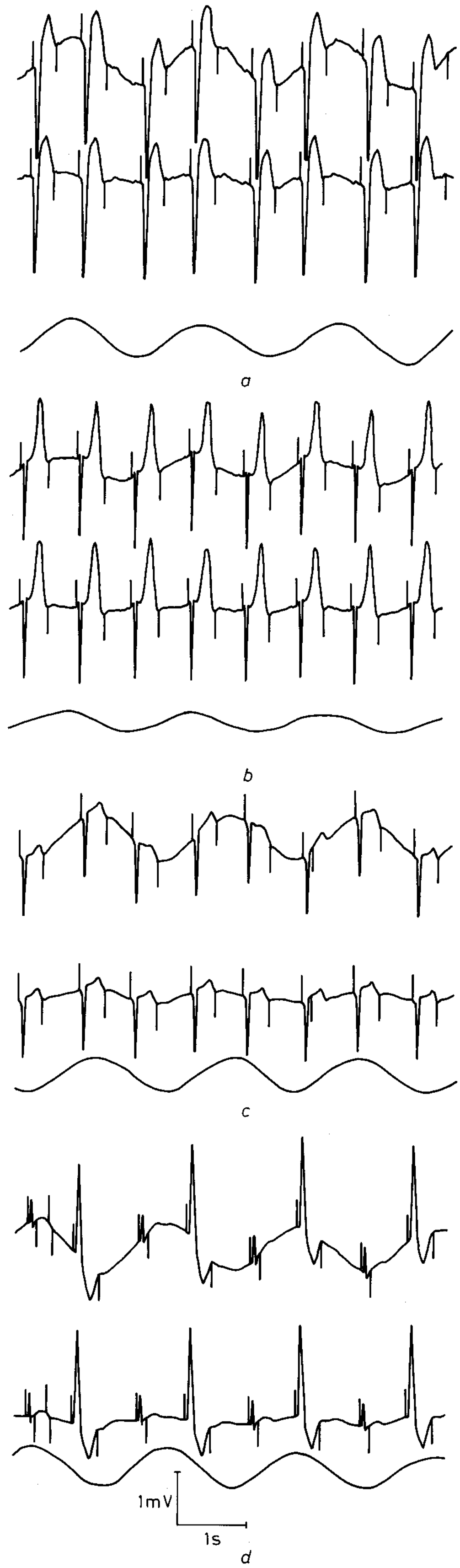

Fig. 4 *Drift filtering with $F_c = 0{\cdot}8$ Hz of ECG signals of various shapes. A 0·5 Hz 0·7 mV sinusoidal 'drift' is superimposed on the original signal. The suppressed waves are marked as in Fig. 3. An example of bigeminy is shown in Fig. 4d*

than 400 12-lead ECG recordings, paying special attention to cases with large monopolar QRS complexes.

An error of 0·1 mV was obtained with $F_c = 0{\cdot}8$ Hz in the worst case of our database, shown in Fig. 4*d*. Such an error, occurring in an ST-segment of 0·6 mV can be judged as acceptable, considering its diagnostic significance. Lesser errors were found in the other recordings of the database, concerning mostly large monopolar complexes, extrasystoles or joined QRS-T complexes with large ST-elevations or depressions.

The assessment of the drift reduction was made by adding a 0·5 Hz 0·7 mV sinusoidal 'drift' to the original signals. This relatively high drift frequency can be considered as a heavy experimental condition. Moreover, the introduction of a known drift to the signal permits an accurate assessment of the filtering procedure. The ideal case would be a difference trace signal equal to the inserted one.

The results can be seen in Fig. 4. The difference traces do not contain distortions correlated with the heart rate. However, the difference traces are not quite sinusoidal. This is due to the fact that the high slew rate intervals of the original signal are eliminated by linear interpolation, whereas with superimposed drift some of these intervals become nonlinear.

In our opinion, $F_c = 1$ Hz or even $F_c = 1{\cdot}2$ Hz could be used, e.g. in monitoring and even in stress ECG testing, but special attention should be paid to the possible errors and their eventual compensation in or influence on the corresponding diagnostic decisions.

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