

# New algorithm for QT interval analysis in 24-hour Holter ECG: performance and applications

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## 1 Introduction

CARDIOVASCULAR DISEASES are the primary cause of death in the adult population. Half of cardiac deaths occur as sudden death. Many studies have been performed in recent years to show the relationship between QT interval and prognosis in postmyocardial infarction patients (PMIPs) (PUDDU *et al.*, 1981; PUDDU and BOURASSA, 1986; SCHWARTZ and WOLF, 1978; AHNVE *et al.*, 1978; TAYLOR *et al.*, 1981). An abnormal QTc (QT corrected with Bazett's formula (BAZETT, 1920)) prolongation in surface can be associated with serious ventricular arrhythmias, syncope and sudden death. We have also demonstrated that the dynamic behaviour of QT measurements made by hand in Holter tapes could be a marker of malignant ventricular arrhythmias (MVAs) (MARTÍ *et al.*, 1988). Thus QT interval measurement in Holter tapes assumes great importance in diagnostic ECG, and several methods are proposed in the literature to implement QT measurement, trying to solve the problem of detecting QRS-wave beginning and T-wave end (ALGRA *et al.*, 1987; PISANI *et al.*, 1985; CRITELLI *et al.*, 1982). The most difficult problem is to define the T-wave end, because of its low frequency components. Some criteria for T-wave end definition (ALGRA *et al.*, 1987; CRITELLI *et al.*, 1982) are based on deflection with reference to the baseline, whereas others make use of first derivative and some threshold (PISANI *et al.*, 1985).

We present a new algorithm which makes use of the first derivative and morphology of each T-wave, and compare its performance with that of manual measurements on a set of approximately 40 beats at the beginning of 18 selected Holter tapes. As an example we present the results obtained from a 24-hour Holter ECG signal.

## 2 Methods

We analysed the data (lead II) recorded on a Holter system (ICR-6500) with a frequency response of 0.05–100 Hz (–3 dB). Holter tapes were played back on a DR

tape recorder (Teac model R-61). Analogue signals were digitised using a 12-bit resolution data-acquisition system installed on a Compaq Model III computer, at an equivalent sampling rate of 250 samples s<sup>-1</sup>. Data were stored in the computer for subsequent detailed analysis. We have developed a new algorithm which is composed of several steps: preprocessing, QRS detection to position beats, QRS onset and T-wave end definition, and selection of possible noisy beats in order to remove them.

### 2.1 Algorithm

**2.1.1 Preprocessing:** In a first step we use a low-pass differentiator (LPD) for QRS detection with the following transfer function

$$G_1(z) = 1 - Z^{-6} \quad (1a)$$

The amplitude response is

$$|G_1(\omega T)| = 2 |\sin(3\omega T)| \quad (1b)$$

where  $T$  is the sampling period, gain is  $6T$ , and the filter processing delay is three samples. We will refer to  $d(k)$  as the differentiated ECG signal.

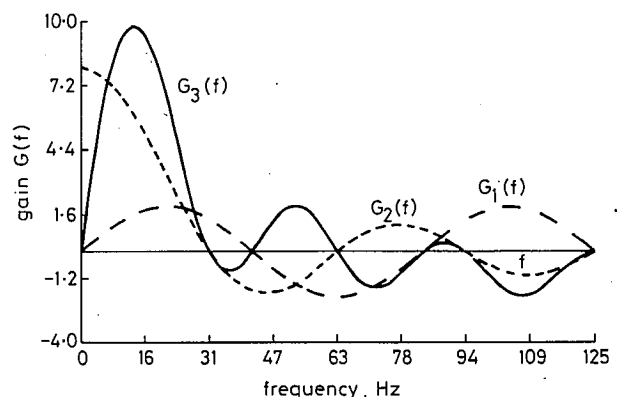


Fig. 1 Transfer functions for differentiator filter  $G_1(f)$ , low-pass filter  $G_2(f)$  and combination of both  $G_3(f)$

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After that we filter the signal with a first-order low-pass filter to avoid residual noise and intrinsic differentiation noise. LYNN (1977) suggested the transfer function of such an integer coefficient filter:

$$G_2(z) = \frac{1 - Z^{-8}}{1 - Z^{-1}} \quad (2a)$$

The amplitude response is

$$|G_2(\omega T)| = \left| \frac{\sin(4\omega T)}{\sin(1/2\omega T)} \right| \quad (2b)$$

DC gain is 8, cutoff frequency at  $-3$  dB is about 20 Hz, and filter delay is  $4\frac{1}{2}$  samples. Fig. 1 shows transfer functions for the differentiator  $G_1(f)$ , low-pass filter  $G_2(f)$  and the combination of both  $G_3(f) = G_1(f)G_2(f)$ .

In future references we will refer to the processed signal  $f(k)$  as the differentiated plus low-pass filtered signal.

**2.1.2 QRS detection:** Once the signal is differentiated and filtered, we implement a QRS detector based on an adaptive threshold as described in PAN and TOMPKINS (1985), with some modifications to increase processing speed and make Q and R-wave peak definition easier.

We define a threshold at the  $n$  beat as  $H_n$  and we detect QRS as the first maximum or minimum whose absolute value is larger than  $H_n$ . As a first step we look for the highest positive or negative peak in the first 2 s signal, and define  $PK_1$  as the absolute value of this initial peak. Then we make  $H_1 = 0.8PK_1$  (80 per cent of peak value).

If  $PK_n$  denotes the maximum absolute value of the QRS at the  $n$  beat in the processed signal, we calculate the next threshold  $H_{n+1}$  as

$$H_{n+1} = 0.8H_n + 0.2(0.8PK_n) \quad (3)$$

We also implement a recursive search back for missing beats, with a lower threshold  $H'_n (H'_n = 4/5H_n)$ , as in PAN and TOMPKINS (1985), when the next beat is farther than 1.8 times the current R-R interval average ( $RR_{av}$ ). We also calculate the actual  $RR_{av}$  from the last R-R interval ( $RR$ )

$$RR_{av} = \begin{cases} 0.8RR_{av} + 0.2RR & 1.5RR_{av} > RR > 0.5RR_{av} \\ RR_{av} & \text{otherwise} \end{cases} \quad (4)$$

**2.1.3 R and Q-waves definition:** After detecting the  $PK_n$  position in the processed signal  $f(k)$ , we search for the nearest peak forward ( $PK_a$ , peak after) and backwards ( $PK_b$ , peak before) as described in Fig. 2. To define the R-position  $R_p$ : we know that the R-wave has the highest slope in the QRS complex; thus  $PK_n$  will be the maximum slope value on either the upward-going side or downward-going side of the R-wave. The other highest slope side of the R-wave will be either  $PK_a$  or  $PK_b$ , depending which has the larger absolute value. Then we define the R-position  $R_p$  as the zero-crossing between  $PK_n$  and the highest absolute value of  $PK_a$  or  $PK_b$ . Fig. 3 shows the flowchart of this algorithm.

In some QRS morphologies the Q or S-wave can be consistently detected as the R-wave. As we are interested in the R-R interval and in normal beats the Q-Q, R-R, or S-S intervals have the same value, we do not take this into consideration.

We define the Q-position  $Q_p$  as the zero-crossing preceding the  $R_p$  position in the differentiated signal (Fig. 2), and not in the filtered ECG, because the Q-wave has high frequencies that cannot be present in a low-pass filtered signal. When the  $R_p - Q_p$  interval exceeds 80 ms no Q-wave is considered. This occurs when the Q-wave has been detected as an R-wave or when no Q-wave is present.

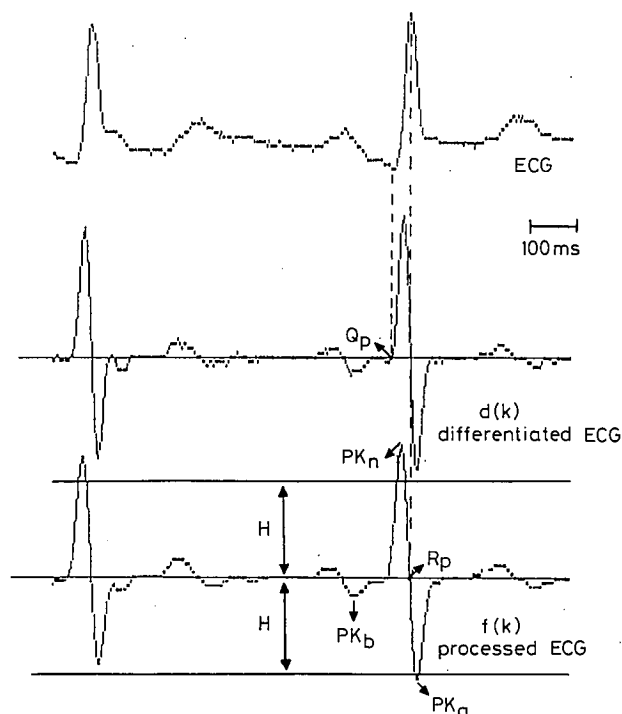


Fig. 2 Original ECG played back from a Holter tape, output signal of the differentiated preprocessing (differentiated ECG) and output of the low-pass filter after differentiation (processed ECG). The threshold  $H$  for QRS detection and  $PK_n$ ,  $PK_a$ ,  $PK_b$ ,  $R_p$  and  $Q_p$  positions are shown as described in text

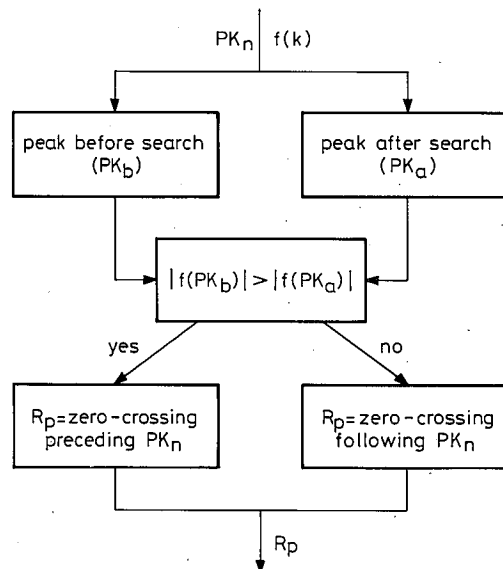


Fig. 3 Flowchart for R-wave position definition  $R_p$ , where the inputs are  $PK_n$  (detected peak) and  $f(k)$  (processed signal)

**2.1.4 QRS onset definition:** In the next stage we define QRS onset ( $QRS_1$ ) as the beginning of the Q-wave (or R-wave when no Q-wave is present). The Q-position  $Q_p$  and the R-position  $R_p$  have already been detected. Then from the  $Q_p$  (or  $R_p$ ) point we search backwards, in the differentiated signal  $d(k)$ , for a point  $Q_i$  (or  $R_i$ ) of maximum slope in the ECG signal (Fig. 4). With this point we define a threshold  $H_q$  (or  $H_r$ ) as the value of the differentiated signal in  $Q_i$  (or  $R_i$ ) divided by a constant  $K$  which takes the value  $K = K_q$  when the Q-wave is present, and  $K = K_r$  when no Q-wave is present;  $H_q = d(Q_i)/K_q$  (or  $H_r = d(R_i)/K_r$ ). From this point we define QRS-begin ( $QRS_1$ ) as the backwards threshold crossing point from  $Q_i$  (or  $R_i$ ). The flow chart for this implementation is also shown in Fig. 5. The algorithm performs best for  $K_q = 2$  and  $K_r = 5$ .

We used different  $K$ -values in the case of Q-wave presence ( $K_q$ ) or only R-wave ( $K_r$ ) due to different maximum slope value for both waves. That considers the

R-wave beginning at a position where the slope value reaches  $\frac{1}{2}$  ( $K = 5$ ) of the maximum, and  $K = 2$  for the Q beginning.

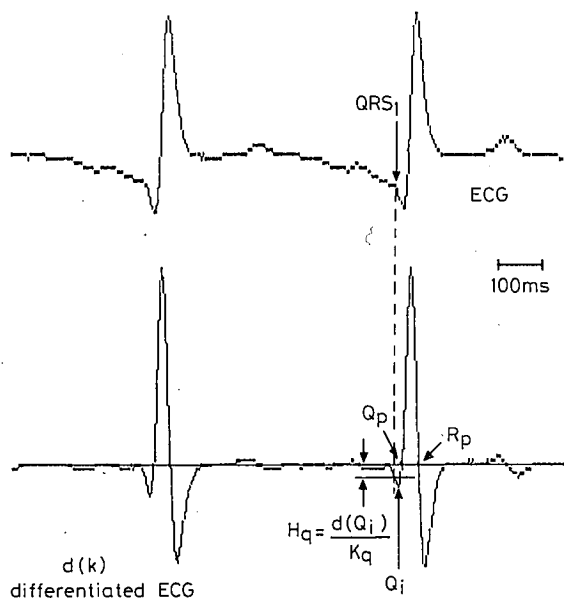


Fig. 4 Original ECG and differentiated ECG where the QRS onset ( $QRS_1$ ) point definition is shown, with the Q-position  $Q_p$ , the point of maximum slope  $Q_i$  and  $H_q$  threshold value

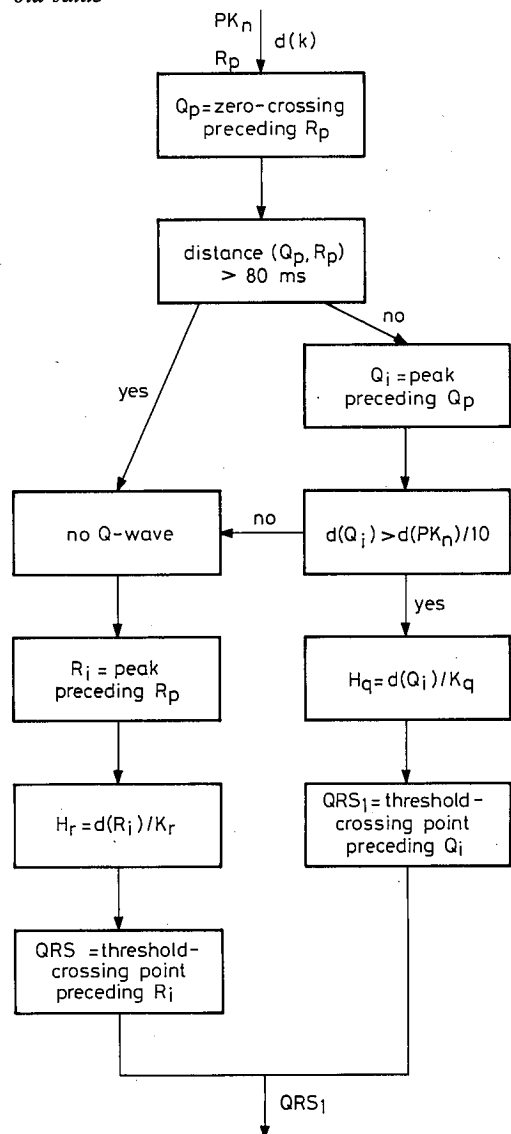


Fig. 5 Flowchart for QRS onset definition  $QRS_1$ , where the inputs are  $PK_n$ ,  $R_p$ , and  $d(k)$  (differentiated signal)

2.1.5 T-wave peak and T-wave end definition: From the R-position we define a search window whose limits are  $bwind$  and  $ewind$  (Fig. 6).

$$(bwind, ewind) = \begin{cases} (140, 500) \text{ ms} & RR_{av} > 700 \text{ ms} \\ (100, 0.7RR_{av}) \text{ ms} & RR_{av} < 700 \text{ ms} \end{cases} \quad (5)$$

We decrease the window size when R-R decreases, to avoid the next P-wave being detected as a false T-wave.  $RR_{av}$  is used to avoid strong changes in the R-R interval. Four different kinds of T-waves are going to be considered: normal T-wave (upward-downward), inverted T-wave (downward-upward), only-downward T-wave, and only-upward T-wave.

We search for the maximum ( $max$ ) and minimum ( $min$ ) values of the processed signal  $f(k)$  in the defined window, as shown in Fig. 6. If the  $max$  position is before the  $min$  position, then the T-wave may be upward-downward or only upward;  $|max| > 4|min|$  is the selected condition to characterise the only-upward T-wave.

If the  $min$  position is before the  $max$  position we search for the minimum ( $mina$ ) between the  $max$  position and the end of the window. If  $mina$  is comparable in absolute value with  $max$  we consider again the upward-downward shape. If not, we compare  $min$  and  $max$ , and if they have similar values, T downward-upward is considered, otherwise T only-downward is assumed. We have obtained a good performance for classification of the four defined T-wave morphologies using a factor of four for the comparison of  $mina$  with  $max$ , and  $min$  with  $max$ .

if  $|max| < 4|mina|$ , then normal T, otherwise next criterion

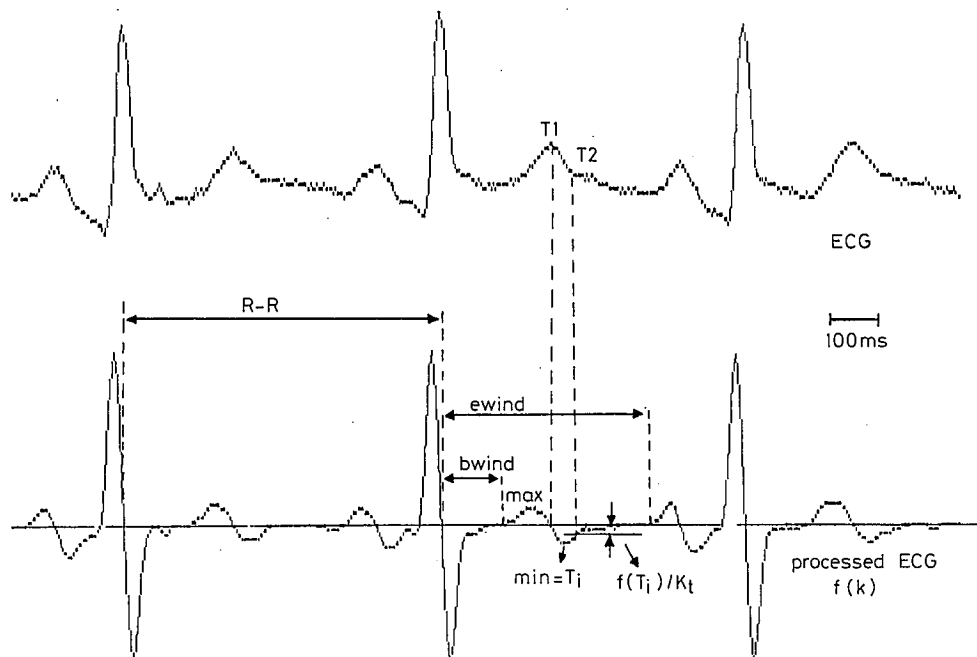
if  $|min| > 4|max|$ , then only-downward T-wave, otherwise T-wave inverted

With these considerations we have the last highest slope point of the T-wave (downward or upward), and from this point  $T_i$  we search for the T-end point as follows.

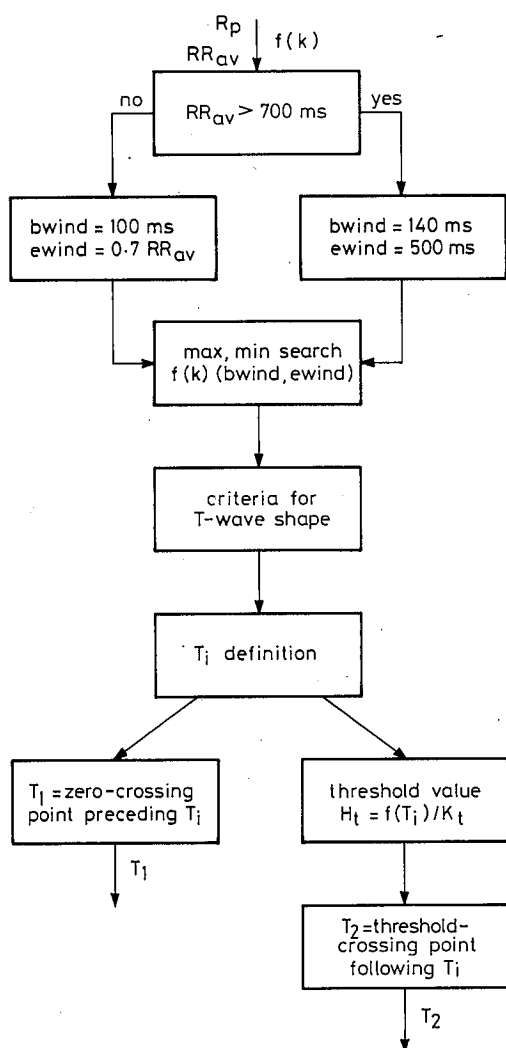
Let  $f(T_i)$  denote the value of the processed signal at this point (sometimes it will be  $min$ ,  $max$  or  $mina$  depending on the T-wave morphology). This value has information on the T-decay rate. We define the T-wave end point  $T_2$  as a forward point from  $T_i$  where the downward (or upward in T-inverted) processed signal reaches a threshold value  $H_t$ ;  $H_t = f(T_i)/K_t$ , where  $K_t$  is an experimental value that in our study has its best performance for  $K_t = 2$ .

This is the most important step of the algorithm. Note that other methods make use of a differentiated signal (PISANI *et al.*, 1985) and a threshold that has no information on the T-decay rate. In our case, when the T-wave has a higher or lower slope value,  $f(T_i)/K_t$  also has a higher or lower value to reach the T-end point. Low-frequency baseline interferences have great importance in methods for T-wave end definition based on baseline considerations (ALGRA *et al.*, 1987; CRITELLI *et al.*, 1982). In our method these have poor influence, because these baseline interferences are of lower frequency than those of the T-wave, and the differentiator has very poor gain for these frequencies in comparison with its gain for T-wave frequencies.

The T-wave peak  $T_1$  is now defined as the first zero-crossing backwards from the  $T_i$  position in the processed signal. Fig. 7 shows a flowchart for this procedure.



**Fig. 6** Original ECG and processed ECG. In the processed signal the T-end definition algorithm is described where  $bwind$  is the distance from the R-position to the beginning of the search window,  $ewind$  is the distance from the R-position to the end of this window,  $max$  and  $min$  are maximum and minimum positions of the processed signal in this window, and  $f(T_i)/K_t$  is the threshold for the T-wave search. Broken lines indicate R-positions, window limits and correspondence with the T-wave peak  $T_1$  and T-wave end  $T_2$  in the original signal



**Fig. 7** Flowchart for T-wave peak  $T_1$ , and T-wave end  $T_2$  definition, where inputs are  $R_p$ ,  $f(k)$  and  $RR_{av}$  (current R-R interval average)

## 2.2 QT values selection

Once the QRS onset ( $QRS_1$ ), T-wave peak  $T_1$  and T-wave end  $T_2$  have been defined, we can calculate the QT interval by subtracting the  $QRS_1$  time from  $T_2$ . We also define QTP as an interval between the beginning of the QRS-wave and the T-wave peak (CRITELLI *et al.*, 1982). Finally, we can apply Bazett's formula to obtain the QT and QTP-intervals corrected for R-R interval variability:  $QT_c$  and  $QTP_c$ , respectively.

$$QT = T_2 - QRS_1 \quad QT_c = \frac{QT}{\sqrt{RR}} \quad (6a)$$

$$QTP = T_1 - QRS_1 \quad QTP_c = \frac{QTP}{\sqrt{RR}} \quad (6b)$$

where  $RR$  is the previous R-R interval.

As ECGs from Holter tapes unfortunately have an important high-frequency noise contamination, errors can occur in QT-interval definition, due fundamentally to changes in the end of the T-wave leading to wrong T-end definition. Also ectopic and other abnormal beats are sources of error in QT interval measurements. To avoid this problem we only select three beats in each set of five measured beats. We reject the highest and the lowest QT intervals in this five-beat set. Also we reject those whose QT value is higher or lower than 15 per cent current average.

Following this procedure we plot  $QT$ ,  $QT_c$ ,  $QTP$ ,  $QTP_c$  and  $RR$  measurements obtained from 24-hour Holter ECG signal recording. The percentage distribution of measurements is also presented as explained below.

## 3 Validation and results

To evaluate the performance we compare automatic and manual measurements. We have taken a set of  $n$  beats at the beginning of 18 tapes, and they have been automatically analysed. Graphic recordings of the same tapes are played back from the Holter system on a paper record at a speed of  $25 \text{ mm s}^{-1}$ , and manual measurements on the same beats are made by two experts ( $expert_1$ ,  $expert_2$ )

(MARTÍ *et al.*, 1988). Manual precision is usually 10 ms, and we know that this value also depends on the observer. The differences between two observers can have values even higher than 10 ms in some cases (PISANI *et al.*, 1985; AHNVE, 1985).

Table 1 shows our validation results. We present the number of beats  $n$  measured in each tape, the mean difference  $d^*$  between automatic and manual measurements, or between experts, and the standard deviation of these differences  $SD^*$  for each patient.  $SD$  and  $d$  are values after rejection of maximum and minimum in five beats. Note that  $SD$  improves from 17.7 to 14.3 ms (*automatic - expert<sub>1</sub>*) and from 17.4 to 14.4 ms (*automatic - expert<sub>2</sub>*) when we reject these beats, and mean error improves from 2.4 to 2.0 ms, and from 0.9 to 0.4 ms respectively. As noted before, the standard deviation between observers is around 10 ms, similar to that reported in the literature.

The standard deviation of the error between the manual and automatic measurements has a value in the range 10–15 ms. We conclude that reproducibility of the QT interval by computer measurements in this way is comparable to human measurement.

Fig. 8 plots various QRS complex patterns in the 18 tapes along with the QT intervals identified by our algorithm. Note that for patients 2, 3, 4, 5, 6 and 12, no Q-wave is present. Fig. 9 also shows results in some especially noisy beats where we can see the performance in these cases with processing signal and definition points.

We have analysed 18 Holter tapes and as an example we present the results obtained from a 24 hour dynamic QT study in a patient with postmyocardial infarction and malignant ventricular arrhythmia. Fig. 10b plots  $QTP$  and  $QT$  trends for this patient over 24 hours; Fig. 10c  $QTP_c$  and  $QT_c$  trends; and Fig. 10a the RR trend for this patient in 24 hours. Each point is the average of measured beats in each 2.5 min period. We can see that the correction with Bazett's formula leads to good results, but there is a remainder  $QT_c$  and  $QTP_c$  dependence with RR interval variations. This had already been demonstrated by recent works (CRITELLI *et al.*, 1982; SADEH *et al.*, 1987) where discrepancies from Bazett's formula are shown and other correction expressions are based on

$$QT = c(RR)^b \quad (7)$$

Table 1 Validation results with two manual measurements and between observers on 18 Holter tapes, where  $n$  is the number of beats measured on each tape,  $d^*$  is the mean difference between automatic and manual QT interval measurements, and  $SD^*$  is the standard deviation.  $d$  and  $SD$  are values after rejection of maximum and minimum QT values in each five-beat set. Units for  $d$ ,  $SD$ ,  $d^*$  and  $SD^*$  are ms

Tape	$n$	Automatic - expert <sub>1</sub>				Automatic - expert <sub>2</sub>				expert <sub>1</sub> - expert <sub>2</sub>	
		$d^*$	$d$	$SD^*$	$SD$	$d^*$	$d$	$SD^*$	$SD$	$d^*$	$SD^*$
1	34	4.3	2.4	33.9	16.9	7.0	-0.8	36.0	12.6	3.6	12.6
2	42	3.5	3.0	15.1	12.0	1.7	-0.1	11.5	9.8	-0.3	14.7
3	26	-5.9	-8.5	21.0	19.5	-7.7	-7.4	18.8	20.9	2.7	18.5
4	19	-2.3	-2.8	7.7	7.3	-7.2	-7.5	7.3	8.1	4.7	6.8
5	38	4.5	3.2	10.3	8.6	4.5	3.2	10.3	8.6	1.9	10.2
6	49	-21.0	-20.0	10.4	9.0	-22	-21.0	14.4	15.4	0.0	10.7
7	39	-7.2	-10.0	20.4	15.0	-26	-28.0	23.7	17.4	19.2	14.4
8	41	1.0	2.4	13.1	13.3	-1.1	-1.2	14.6	14.5	2.1	14.1
9	29	16.5	16.8	18.7	19.3	16.7	16.5	16.9	17.7	-1.0	5.5
10	39	-8.8	-8.5	13.1	12.4	-7.1	-7.3	14.1	14.9	-1.3	7.6
11	42	9.5	9.1	14.0	11.6	6.8	6.3	11.0	7.5	2.6	7.9
12	45	-17.0	-16.0	16.7	14.6	-18	-15.0	19.1	16.9	0.9	8.4
13	36	8.8	9.4	17.8	15.2	9.4	11.4	19.0	17.5	-0.6	5.7
14	34	7.7	10.9	17.0	14.8	12.3	14.6	17.7	15.7	-4.4	7.3
15	37	8.2	5.1	44.7	22.1	6.1	5.8	39.8	21.9	2.2	15.4
16	35	11.7	11.7	16.0	16.0	9.7	8.5	12.5	11.5	1.7	12.8
17	40	19.6	17.9	13.2	12.9	21.9	21.6	12.3	11.9	-0.2	6.5
18	35	9.5	9.3	15.9	17.5	9.9	8.4	14.8	16.7	-0.3	5.0
Mean	37	2.4	2.0	17.7	14.3	0.9	0.4	17.4	14.4	1.9	10.2

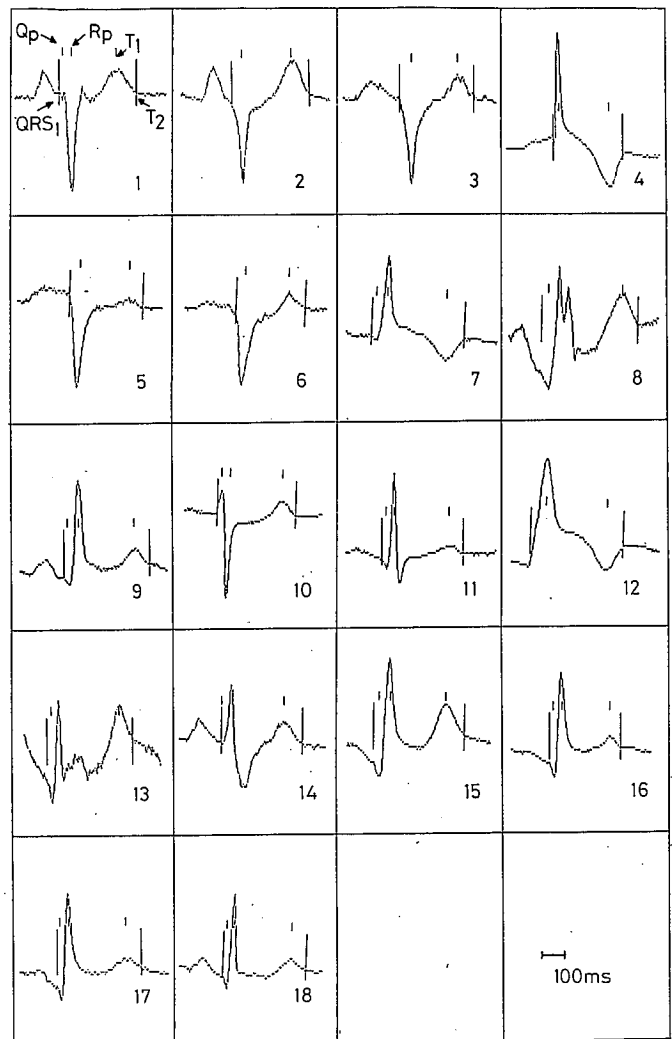
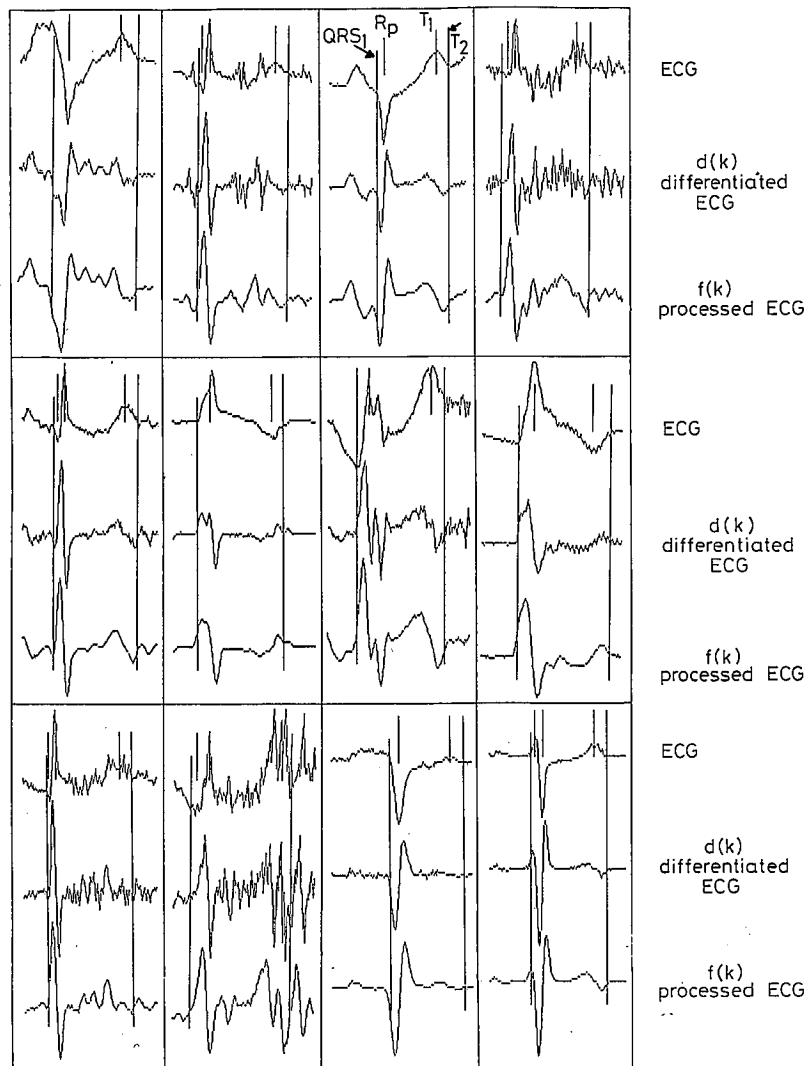
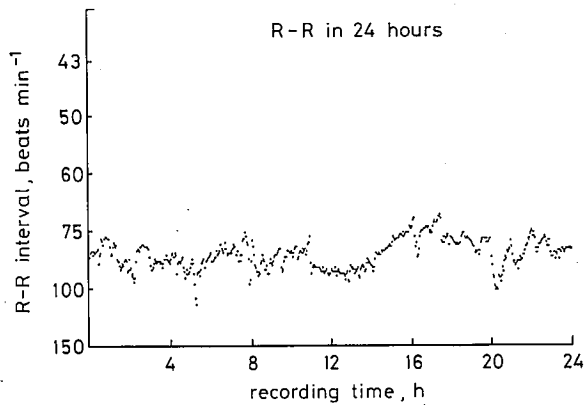


Fig. 8 Patterns of one beat for each of the 18 tapes used in validation. Long lines show the Q wave beginning  $QRS_1$  and T-wave end  $T_2$  points on these beats as defined by the algorithm; and short lines show Q, R and T peak positions, in this order, when all are present. When there are just two, no Q-wave presence is detected, and R and T-wave peak positions are indicated



**Fig. 9** Patterns of 12 especially noisy beats, with differentiated and low-pass filtered signals where we can see definition points on every processed signal. The long lines are the QRS onset  $QRS_1$  and T-wave end  $T_2$  points, and the short lines show the Q or R-positions  $Q_p$  or  $R_p$  and T-wave peak  $T_1$



**Fig. 10** Dynamic results obtained from a 24 hour Holter tape. (a) R-R interval translated to  $\text{beats min}^{-1}$ ; (b) QT and QTP intervals in ms; (c)  $QT_c$  and  $QTP_c$  intervals in  $\text{ms s}^{-1/2}$ . Each point on the graph is the average of measured interval values in periods of 2.5 min

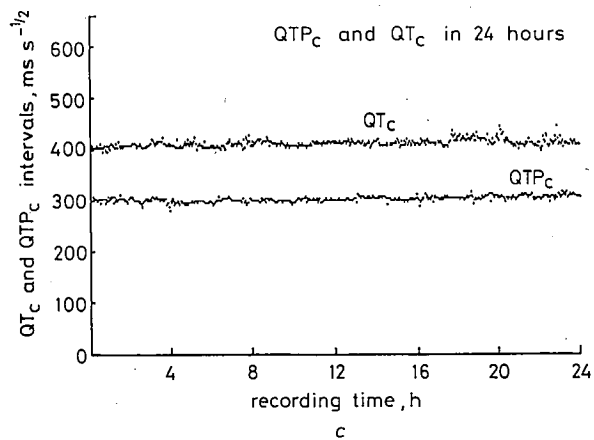
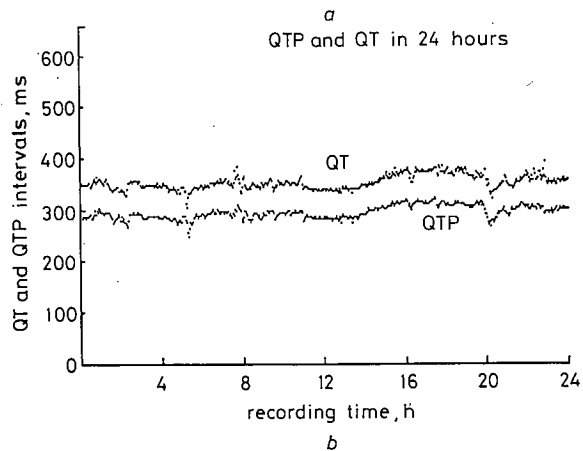


Table 2 Results from a patient. QT, QT<sub>c</sub>, QTP, QTP<sub>c</sub> and R-R intervals are plotted as percentage of beats in 20 ms intervals for QT values, and intervals of 100 ms for RR values. Results are rounded to nearest whole integer numbers

Interval, ms	QT, beats per cent	QTP, beats per cent	QT <sub>c</sub> , beats per cent	QTP <sub>c</sub> , beats per cent	Interval, ms	RR, per cent
240-260	—	1	—	—	400-500	—
260-280	—	21	—	—	500-600	—
280-300	—	53	—	3	600-700	—
300-320	—	23	—	38	700-800	15
320-340	—	2	—	48	800-900	49
340-360	4	—	—	10	900-1000	24
360-380	32	—	2	1	1000-1100	9
380-400	38	—	26	—	1100-1200	3
400-420	23	—	55	—		
420-440	3	—	15	—		
440-460	—	—	2	—		

Bazett's formula uses  $b = \frac{1}{2}$  and SADEH *et al.* (1987) proposed  $b = 0.13$ . These discrepancies suggest future investigation in a higher number of patients, and dependency on pathology, age etc.

Data are also analysed by classifying the measurements obtained in intervals of 20 ms for QT, QT<sub>c</sub>, QTP and QTP<sub>c</sub>, and in intervals of 100 ms for RR interval. Table 2 presents the percentage of beats in each interval.

#### 4 Discussion and conclusions

A new algorithm is presented for QT interval measurement. Its accuracy is comparable with results of other QT algorithms (ALGRA *et al.*, 1987; PISANI *et al.*, 1985) and to manual measurement, but it has a better immunity to baseline interferences when the baseline differs from a flat line after T-end, a situation quite common in Holter ECG recordings. Other algorithms for QRS onset (Q-beginning) involve a significantly higher amount of calculation for a comparable performance (NYGÅRDS and SÖRNMO, 1983).

At present this system is implemented in 'turbo Pascal' language and runs on a PC compatible, taking about 90 min processing time in a Compaq III computer for 24 hour recorded ECG. No ectopic beat rejection is implemented in the QRS detector, thus the subsequent selection is supposed to remove these abnormal beats. Problems can be found in ventricular tachycardia or ventricular fibrillation; in these cases the algorithm will give nonsense values because no QT interval is defined. Further work to detect and remove this kind of interference is required.

Also, in sequences with very low signal-to-noise ratios wrong values can result. Moreover, if these sequences are only of one or a small number of beats, the problem is minimised by rejection criteria. Now the problem is only important in recordings where there is an electrode problem or continuous interference occurs.

The availability of 24 hour ECG signal processing, to make use of the QT interval as a marker of MVA in PMIP, is presented. The relationship and dependence of QT and QT<sub>c</sub> in patients with different pathologies suggests further investigation to show the real value of these parameters as predictors of sudden death and other heart diseases (MARTÍ *et al.*, 1988); this is the reason for the presentation as a percentage of beats in some defined intervals. The interpretation of the QT<sub>c</sub> interval to quantify prognosis in PMIP should take into account at least the inaccuracy of measurements. A study of the relationship between QTP<sub>c</sub> and QT<sub>c</sub> will be interesting, because if both give the same information QTP measurement is more accurate than QT.

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#### References

- AHNVE, S., LUNDMAN, T. and SHOALEH-VAR, M. (1978) The relationship between QT interval and primary ventricular fibrillation in acute myocardial infarction. *Acta Med. Scand.*, **204**, 17-19.
- AHNVE, S. (1985) Errors in visual determination of corrected QT (QT<sub>c</sub>) interval during acute myocardial infarction. *J. Am. Coll. Cardiol.*, **5**, 699-702.
- ALGRA, A., LE BRUN, H. and ZEELLENBERG, C. (1987) An algorithm for computer measurement of QT intervals in the 24 hour ECG. In *Computers in Cardiology*. IEEE Computer Society Press, 117-119.
- BAZETT, H. C. (1920) An analysis of the time relation of electrocardiograms. *Heart*, **7**, 353-370.
- CRITELLI, G., MARCIANO, F., MAZZARELLA, M. and MIGAUX, M. L. (1982) QT interval measurements of long-term ECG recordings. Application to an automatic Holter analysis system. In *Computers in Cardiology*, IEEE Computer Society Press, 481-480.
- LYNN, P. A. (1977) Online digital filters for biological signals: some fast designs for a small computer. *Med. & Biol. Eng. & Comput.*, **15**, 534-540.
- MARTÍ, V., BAYÉS DE LUNA, A., ARRIOLA, J., SONGA, V., GUINDO, J., DOMINGUEZ DE LAS ROZAS, J., MARRUGAT, J., SANZ, F., THAKOR, N., MIN, Y., CAMINAL, P. and LAGUNA, P. (1988) Value of dynamic QT<sub>c</sub> in arrhythmology. Proc. 8th. Int. Congr. The New Frontiers of Arrhythmias, Marilleva, Italy, 683-691.
- NYGÅRDS, M.-E. and SÖRNMO, L. (1983) Delineation of the QRS complex using the envelope of the e.c.g. *Med. & Biol. Eng. & Comput.*, **21**, 538-547.
- PAN, J. and TOMPKINS, W. J. (1985) A real time QRS detection algorithm. *IEEE Trans.*, **BME-32**, 230-236.
- PISANI, E., PELLEGRINI, E., ANSUINI, G., DI NOTO, G., RIMATORI, C. and RUSSO, P. (1985) Performance evaluation of algorithms for QT interval measurements in ambulatory ECG recording. In *Computers in Cardiology*, IEEE Computer Society Press, 459-462.
- PUDDU, P. E., JOUVE, R., TORRESANI, J. and JOUVE, A. (1981) QT interval and primary ventricular fibrillation in acute myocardial infarction. *Am. Heart J.*, **101**, 118-119.
- PUDDU, P. E. and BOURASSA, M. G. (1986) Prediction of sudden death from QT<sub>c</sub> interval prolongation in patients with chronic ischemic disease. *J. Electrocardiol.*, **19**, 203-212.
- SADEH, D., SHANNON, D. C., ABOUD, S., AKSELROD, S. and COHEN, R. J. (1987) A new technique to determine the correlation between the QT interval and heart rate for control and sides babies. In *Computers in Cardiology*, IEEE Computer Society Press, 125-127.
- SCHWARTZ, P. J. and WOLF, S. (1978) QT interval prolongation as predictor of sudden death in patients with myocardial infarction. *Circ.*, **57**, 1074-1077.
- TAYLOR, G. J., CRAMPTON, R. S., GIBSON, R. S., STEBBINS, P. T., WALDMAN, M. T. G. and BELLER, G. A. (1981) Prolonged QT interval at onset of acute myocardial infarction in predicting early phase ventricular tachycardia. *Am. Heart J.*, **102**, 16-24.