An Automated System for *ST* Segment and Arrhythmia Analysis in Exercise Radionuclide Ventriculography

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***Abstract*—A computer-based system for interpretation of the electrocardiogram (ECG) in the diagnosis of arrhythmia and *ST* segment abnormality in an exercise system is presented. The system was designed for inclusion in a gamma camera so that ECG diagnosis could be combined with the diagnostic capability of radionuclide ventriculography. Digitized data are analyzed in a beat-by-beat mode and a contextual diagnosis of underlying rhythm is provided. Each beat is assigned a beat code based on a combination of waveform analysis and *RR* interval measurement. The waveform analysis employs a new correlation coefficient formula which corrects for baseline wander. Selective signal averaging, in which only normal beats are included, is done for an improved signal-to-noise ratio prior to *ST* segment analysis. Template generation, *R* wave detection, *QRS* window size, baseline correction, and continuous updating of heart rate have all been automated. *ST* level and slope measurements are computed on signal-averaged data. Arrhythmia analysis of 13 passages of abnormal rhythm by computer was found to be correct in 98.4 percent of all beats. 25 passages of exercise data, 1-5 min in length, were evaluated by the cardiologist and found to be in agreement in 95.8 percent in measurements of *ST* level and 91.7 percent in measurements of *ST* slope.**

Introduction

We have designed a fully automated system for interpretation of the electrocardiogram (ECG) in the diagnosis of arrhythmia and *ST* segment abnormality for use in exercise radionuclide ventriculography. In addition, the software will provide a smart trigger for control of image acquisition. Computer techniques have been applied to the exercise electrocardiogram (ECG) for the past 20 years [1], [2] and several techniques have become somewhat standard for processing multilead signals in real time. Signal averaging of successive beats to improve the signal-to-noise ratio is a commonly used technique [3] and measurement of *ST* segment amplitude [1], [2], [4], [5], slope [4], [5], vector [6], and time-voltage integral [7] are typically computer derived and analyzed. Signal averaging which allows inclusion of abnormal beats will produce erroneous results in the measurement of *ST* features. Typically rejection of abnormal beats is done by *RR* interval measurement. We have developed a computer system which analyzes each beat individually for detection of abnormal rhythm or morphology, and selectively averages only those beats found to have normal *RR* intervals and normal *QRS* waveform. Waveform analysis is based on a new correlation coefficient. This *weighted* correlation coefficient gives better performance than previous correlation methods [8]-[11] on the exercise ECG because it corrects for baseline drift, yet its computation time is much less than the traditional cross correlation in which a mean value is required. Therefore, this new method allows real-time execution of the analysis system. The system provides arrhythmia diagnosis as well as *ST* amplitude and slope measurements. It automatically generates a template for waveform analysis, individually determines a window size for *QRS* analysis, derives a baseline for purposes of *ST* measurements, computes the weighted correlation coefficient for each beat, and provides a running average *RR* interval. All *ST* calculations are made on the selective signal-averaged beat, and on-line individual beat analysis and *ST* measurements are given. A statistical summary is reported after all beats are processed which includes a beat-by-beat record of rhythm abnormalities, contextual diagnosis of underlying arrhythmias, and measurements of *ST* level and slope. In addition to providing an automated analysis of the exercise electrocardiogram, the ECG software was specifically designed for inclusion in a gamma camera for sophisticated control of image acquisition in radionuclide ventriculography. During gated blood pool studies where an ECG trigger typically controls acquisition of a series of images of the heart during a cardiac cycle, the exclusion of "bad beats" is critical. A simple-minded trigger can reject beats only on *RR* interval information. Our algorithm, which does a beat-by-beat recognition of seven classes of abnormal beats, will provide a smart trigger to direct image acquisition of only identical cardiac cycles.

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Methods and Material

Hardware

Patient electrocardiogams (ECG) were recorded at the Heart Station at University of Michigan Hospital using the HP 3964A FM instrumentation recorder. Patients were connected to a 12-lead electrocardiograph (Cambridge, model CM3000) set to a gain of 1 (10 mm/mv) and three channels of unfiltered electrocardiographic data (AVF, V2, V5) were tape-recorded at speed of $3\frac{3}{4}$ ips. A strip chart of the ECG was taken during the entire exercise test for later validation. A 1 mv calibration signal was recorded prior to the recording of each patient's ECG signal. The data were subsequently played back for off-line processing in the Medical Computing Laboratory at the University of Michigan. The computer system used for processing is an Intel 8086/8087 processor within an APEX medical computer system (Elscint, Ltd. Model 215). The configuration consists of the following components: 1 Mbyte of on-board memory, an analog-to-digital (A/D) converter, an ECG hardware trigger circuit, a 30 Mbyte hard disk, and two CRT screens. This special purpose system (designed for nuclear imaging) was specifically configured in order to incorporate data acquisition of the ECG signal. Data were digitized at a sampling rate of 250 Hz, with an input range of ± 5 V, 8 bits resolution, and were triggered internally for recognition of the *QRS* waveform. During digitization, strip chart recordings were made from tape recorded data at paper speed of 50 mm/s Gould 8-channel, Model Brush 481). After processing, the digitized samples were displayed on the strip chart recorder via a digital-to-analog (D/A) subsystem (Tecmar Lab Master).

Patient Selection

For development of the arrhythmia analysis, we used tape-recorded data acquired prior to this study from patients in coronary intensive care [9]. 13 passages from 1 to 5 min long which contained a variety of abnormal rhythms were processed to test the rhythm analysis portion of the software. There were 1736/2741 abnormal beats of the following variety: premature ventricular beats, premature atrial beats, aberrant beats, ventricular escapes, compensatory pause, and long *RR*. In addition there were examples of 9 of our 11 contextual arrhythmias (all but the last two in Table II). The use of this special data set for rigorous testing of the rhythm algorithm was necessary because patient recordings made during exercise procedures which were used for testing of the completed system contained few abnormal rhythms.

Ten patients and five normal subjects were electrocardiographically recorded while undergoing exercise testing on a treadmill or bicycle ergometer for a period of 15–30 min. 25 passages 1–5 min in length from these 15 subjects were computer processed by the algorithm described in the following section.

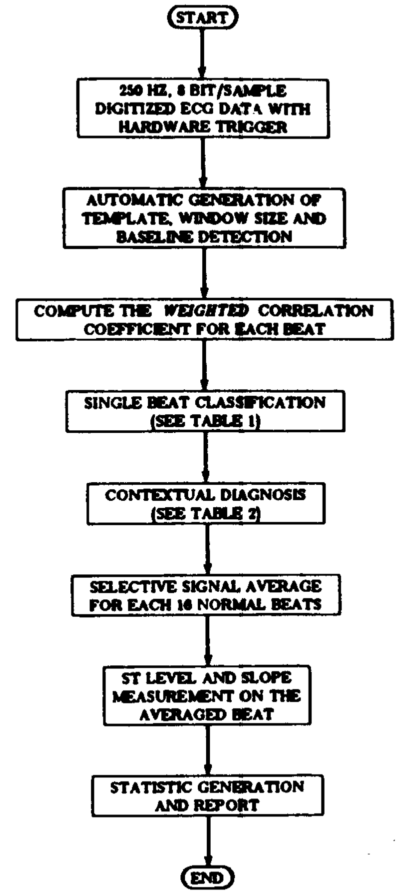


Fig. 1. Block diagram of ECG system for computer analysis of exercise with automatic arrhythmia diagnosis.

Computer Implementation

The first stage of the ECG analysis is the algorithm for individual beat classification and contextual arrhythmia analysis. This beat-by-beat classification is the basis of the complete algorithm which segregates beats by type to allow selective signal averaging of *normal* beats prior to *ST* segment measurement. Fig. 1 shows a block diagram depicting the total program.

An important feature of the system is automatic template generation in which a reference *normal QRS* complex is created for subsequent waveform analysis. The first 20 beats of each passage are used to generate automatically the template for subsequent processing. The dominant *RR* interval group G_d is determined from these 20 beats and all beats associated with the dominant *RR* are signal averaged to produce the template using the following formula.

$$A_k^m = \sum_{i=l}^{l+r} [V_i^m + \delta_i^m] \frac{1}{r+1} \qquad (1)$$

where

A_k^m = kth beat, mth point of output signal
δ_i^m = ith beat, mth correction factor

m = index for point of the current beat
$r + 1$ = number of beats in one averaging operation
k = denotes the kth beat
l = starting index of the average window
V_i^m = ith beat, mth point of input signal.

To determine the dominant RR interval group G_d we define a special partition on a group of beats based on the corresponding RR interval of the beat using the following criteria. Let E be a set of candidate beats. A partition based on the corresponding RR interval of the beat is a set Ω which consists of the exclusive subset S_i of E. Each S_i has a unique RR interval representative value, denoted by RR_i, $i = 1, \cdots, N$ where N denotes the total number of subsets in Ω and $\cup_{i=1}^{N} S_i = E$. Any beat b in E will be covered by an S_i if and only if the associated RR interval of b is in the range $[RR_i - \nabla, RR_i + \nabla]$. If b is covered by S_i, the beat b will be grouped into the set S_i. A dominant RR interval group G_d of E is the member of Ω with maximum cardinality. In general ∇ must be specified by the program so that the partition Ω is valid, i.e., no overlap exists between any two sets in the partition Ω. For actual application, the ∇ value cannot be selected so large that the partition of E contains only one subset (E itself) or too small such that the partition contains only single element subsets. For our application, the best ∇ value was found to be in the range of 10–13 percent of each RR_i. Upon initialization, RR_i, $i = 1, \cdots, N$ are computed automatically by the program and are *updated* when any new beat is grouped. Fig. 2 is the flow chart depicting this algorithm.

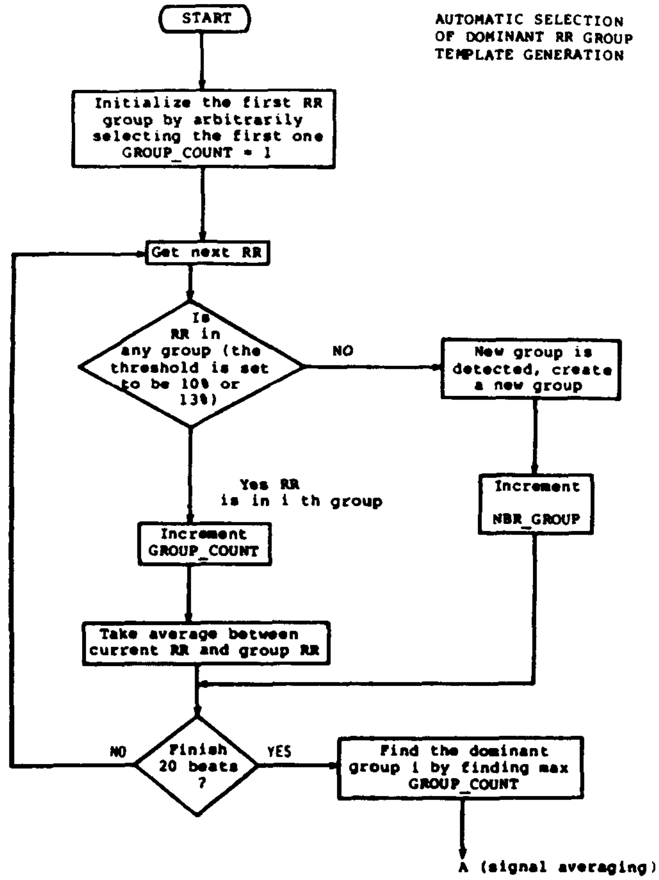


Fig. 2. Flow chart of automatic template generation. All beats in the group containing the dominant RR interval, G_d, are signal-averaged to produce the QRS template.

After the template has been created the following features are found: the peak of the R wave, the P–Q segment (baseline), and the QRS window size. These features are indicated in Fig. 4. Location of a baseline reference point is determined automatically from the template and is used for baseline normalization of subsequent beats. The baseline level is found within the P–Q segment by searching backward from the beginning of the Q-wave until a passage of three consecutive points is found to be in agreement with a rule of flatness. The flatness of a passage of three consecutive points within the P–Q segment is defined as an absolute signal difference pairwise less than or equal to one unit of digitized data. A flow chart depicting the algorithm is seen in Fig. 3. The window size for QRS waveform analysis is determined automatically and is based on recognition of the isoelectric portion of the P–Q segment and the major peak of the QRS. The QRS window is defined as twice the sum of $d1$, $d2$, and $d3$. In Fig. 4 is seen a schematic depicting the automatically determined points described above.

During processing, each incoming QRS complex in the exercise passage is subjected to a statistical comparison with the template. To provide for baseline normalization, a *weighted* correlation is computed to give an index representing normal or abnormal morphology. The correlation coefficient ρ between each beat and the template is computed using the formula

$$\rho = \frac{\sum_{i=1}^{W} X_i(Y_i - \delta)}{\left(\sum_{i=1}^{W} X_i^2 \sum_{i=1}^{W} (Y_i - \delta)^2\right)^{1/2}} \tag{2}$$

where

X_i denotes the ith sample point within the QRS window of the template
Y_i denotes the ith sample point within the QRS window of the current beat
W denotes the window size
δ is a baseline correction factor which is computed by subtracting the baseline voltage level of the current beat from the baseline voltage level of the template

This *weighted* correlation formula which corrects for baseline wander by simple subtraction of the current dc offset has been found superior to the correlation which assumes zero means in the processing of the exercise ECG. We have rejected the standard correlation formula

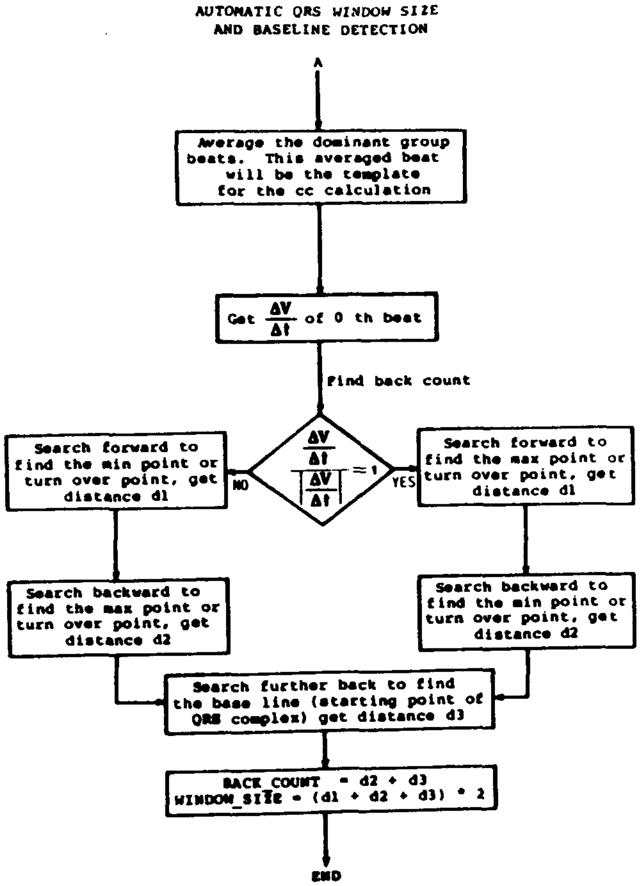
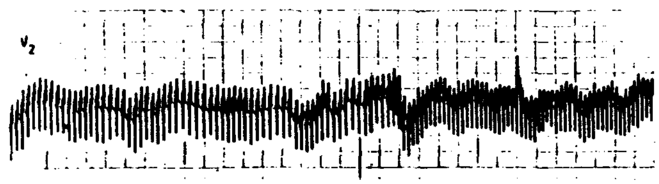


Fig. 3. Automatic baseline detection and *QRS* window size calculation. A search is made for the region of the *P-Q* segment which meets the criteria of flatness for recognition of the isoelectric line. The distance from this location to the major peak or valley of the complex defines one-half the *QRS* window size.



FORMULA ONE

$$\frac{\sum_{i=1}^{W} X_i (Y_i - \bar{Y})}{\left[\sum_{i=1}^{W} X_i^2 \sum_{i=1}^{W} (Y_i - \bar{Y})^2\right]^{\frac{1}{2}}}$$

#	RR	C.C.
0001	0872	+0.99784000
0002	0786	+0.99419800
0003	0834	+0.98620400
0004	0799	+0.99812600
0005	0772	+0.99764400
0006	0723	+0.99092800
0007	0695	+0.97762500
0008	0678	+0.99434300
0009	0626	+0.97472900
0010	0613	+0.98464500
0011	0614	+0.99221000
0012	0618	+0.98584500
0013	0603	+0.96246200
0014	0638	+0.99711500
0015	0652	+0.99254000
0016	0663	+0.99090200
0017	0697	+0.99742700
0018	0672	+0.98996100
0019	0760	+0.99116500
0020	0774	+0.98238800

FORMULA TWO

$$\frac{\sum_{i=1}^{W} X_i Y_i}{\left[\sum_{i=1}^{W} X_i^2 \sum_{i=1}^{W} Y_i^2\right]^{\frac{1}{2}}}$$

#	RR	C.C.
0001	0872	+0.80367500
0002	0786	+0.97544700
0003	0834	+0.98360500
0004	0799	+0.98076400
0005	0772	+0.99685800
0006	0723	+0.98868300
0007	0695	+0.97355100
0008	0678	+0.95771400
0009	0626	+0.75396900
0010	0613	+0.79697700
0011	0614	+0.90825600
0012	0618	+0.96175600
0013	0603	+0.78865300
0014	0638	+0.72182800
0015	0652	+0.71752500
0016	0663	+0.79196900
0017	0697	+0.96009300
0018	0672	+0.98230100
0019	0760	+0.98218200
0020	0774	+0.97716200

Fig. 5. Comparison of two methods of waveform analysis with and without baseline correction. The ECG passage shown is recorded at a chart speed of 2 mm/s to demonstrate the presence of major baseline drift. A correlation coefficient computed by formula 2 on the right degrades dramatically in presence of baseline changes. (See beats 1, 9, 10, 13-16.) The same passage processed with formula 1 on left corrects for baseline and correctly classifies all normal beats.

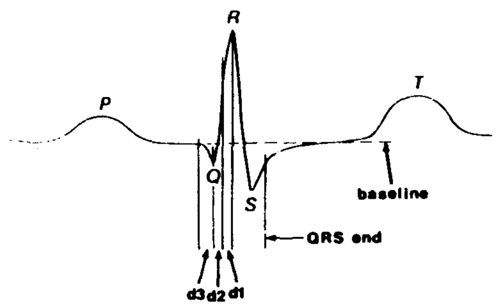


Fig. 4. Schematic of electrocardiographic points of interest which are automatically detected by computer where $d1$ is the interval from peak R to the trigger location, $d2$ is the interval from the trigger location to the trough of Q, and $d3$ is the interval from the trough of Q to the end of baseline or isoelectric line.

in which the actual means are subtracted because of the time-consuming computation involved. In Fig. 5, we show a typical exercise ECG and the correlation results by two different methods. The chart speed in Fig. 5 was set to 2 mm/s to demonstrate the presence of baseline wander. It can be seen that beats number 1, 9, 10, and 13-16 reveal a low correlation in the simple product moment method (formula two) but correlate well in our modified method (formula one). By examining the original ECG strip chart played at a faster speed, one can confirm that the beats indicated above are all normally shaped beats.

After the correlation coefficient has been computed, a threshold is applied to separate *QRS* waveforms into one of two types, normal and abnormal. This plus the size of the corresponding *RR* interval determines the classification of the beat. Table I shows the single beat classification scheme. The computed correlation coefficient (ρ) is considered *normal* if it is above or equal to the predefined threshold which is determined empirically from the data. The default value for the threshold is $\rho = 0.85$. The current *RR* interval, denoted by RR_c, is considered normal if it falls into the normal *RR* range which was defined to be $[RR_n(1 - f), RR_n(1 + f)] \equiv [RR_L, RR_H]$, where RR_n is a running average updated with each *normal* beat, and f is the computed tolerance. Tolerance is predefined as 15 percent of the normal *RR* interval unless this value is overridden by an operator entry. If RR_c is less than RR_L it is called a *short RR* interval. If RR_c is greater than RR_H but less than $2 * RR_H$, it is called a *long RR* interval. Each possible combination of the outcome of ρ and the associated RR_c interval is assigned a beat code according to the rules in Table I. A normal beat is defined as one with

TABLE I
SINGLE BEAT CLASSIFICATION

Code	Type	Criteria
1	Normal	($\rho \geq$ th) and (RRc = N)
2	Premature Ventricular Beat(PVB)	($\rho <$ th) and (RRc = S)
3	Premature Atrial Beat (PAB)	($\rho \geq$ th) and (RRc = S)
4	Nonconducted Beat (N.COND)	(RRc = 2 * RRn)
5	Ventricular Escape Beat(VEB)	($\rho <$ th) and (RRc = L)
6	Long RR interval (LNG.RR)	($\rho \geq$ th) and (RRc = L)
7	Compensatory pause (CMP.P.)	($\rho \geq$ th) and (RRc = L) and (RRc + RRc' = 2 * N) and (previous beat is a PVB)
8	Aberrant beat (ABE.B)	($\rho <$ th) and (RRc = N)
9	None	(If code is not 1 ~ 8)

th = threshold for the correlation coefficient
ρ = the correlation coefficient of current beat
RRc = the current RR interval
RRn = the normal RR interval
RRc' = the previous RR interval
S = denotes a short RR interval range
L = denotes a long RR interval range
N = denotes the normal RR range

TABLE II
CRITERIA FOR ECG CONTEXTUAL DIAGNOSIS

Type	RR interval	Consecutive n	Pattern
1. Couplet	-	n = 2	PVB
2. Triplet	-	n = 3	PVB
3. Bigeminy	-	n = 4	{N,PVB,N,PVB}
4. Trigeminy	-	n = 6	{N,N,PVB,N,N,PVB}
5. Ventricular rhythm	RR $\geq$ 600	n $\geq$ 4	PVB
6. Salvo	RR < 600	4 $\geq$ n $\geq$ 7	PVB
7. Run	400 < RR < 600	n $\geq$ 8	PVB
8. Ventricular tachycardia	400 > RR	n $\geq$ 8	PVB
9. Supraventricular tachycardia	400 > RR	n $\geq$ 8	PAB
10. Bradycardia	RR > 1500	n $\geq$ 8	-
11. Asystole	RR $\geq$ 4000	-	-

RR = RR interval in msec
N = Normal beat or compensatory pause
n = Number of consecutive beats

normal ρ and normal *RR* interval. All remaining combinations are called *abnormal* and are further classified into seven different categories. An algorithm for contextual diagnosis of underlying rhythm was developed based on patterns of sequences of the individual beat codes. Particular strings of successive abnormal beat codes reflect specific arrhythmias and the criteria for each diagnostic statement are shown in Table II. These criteria define simple diagnoses such as couplet, triplet, bigeminy, and trigeminy as well as more complex patterns such as ventricular tachycardia.

Since heart rate usually increases with exercise, automatic updating of the normal RR_n interval is done continuously during the entire ECG passage. This running RR_n is the average of the previous 16 normal *RR* intervals. Abnormals are excluded. During the rhythm diagnosis stage, the single beat diagnosis is reported after each beat is processed and a contextual diagnosis is given if an arrhythmia is present.

Further analysis of the ECG is done for recognition of ischemic events by recognition of *ST* segment changes which occur during exercise. The system implements a signal averaging technique which selectively includes only normal beats and rejects any that are not. The signal-averaging algorithm uses the single beat classification as its basis and allows inclusion of only beats with a code of 1 and accumulates 16 such beats into a summed beat. The interval of the summed beat is 320 ms and contains *P-QRS-T* complexes. *ST* measurements are made on this summed beat which is subsequently averaged for reconstruction purposes. This is repeated with each successive group of 16 normal beats. The technique of computing *ST* measurements on the summed beat (before averaging) results in an effective resolution of 2.44 μV in the scaled *ST* measurements. Division by 16 to compute the averaged values is done in floating point, thus retaining the precision inherent in the initial calculations.

Quantities measured are *ST* depression or elevation, and *ST* slope. Both measurements are made at a location within the *ST* segment determined by the following formula [12].

$$ST - \text{POINT} = R + 64 \text{ ms} + \max\left(4, \frac{200 - HR}{16}\right) \times 4 \text{ ms} \quad (3)$$

or

$$ST - \text{POINT} = S + 44 \text{ ms} + \max\left(4, \frac{200 - HR}{16}\right) \times 4 \text{ ms} \quad (4)$$

where

R denotes the peak of *R* in the case of a beat with primarily positive deflection
S denotes the trough of *S* in a beat with primarily negative deflection
HR denotes the current heart rate as computed from the interval of the current RR_n.

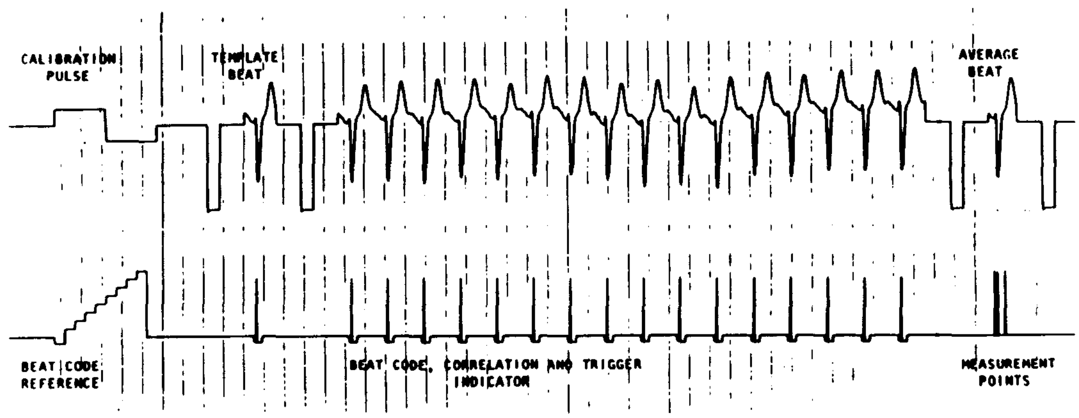


Fig. 6. Digital-to-analog reconstruction of a passage of 16 normal beats showing template and average beat (channel 1). On the lower channel (left) is seen the beat code reference. The spike depicts the location of the *QRS* trigger, the pulse beneath the spike indicates the beat code (in this case normal = 1) and the width of the pulse represents the *QRS* window size. On the right (lower channel) the three spikes show the location of baseline point, trigger, and *ST* measurement point on the averaged beat.

This effectively allows the *ST* measurement point to be modified at heart rate changes.

The *ST* level is a relative voltage difference between the measurement location and the isoelectric level. This isoelectric level is automatically determined from the amplitude of the *P–Q* segment of the current averaged beat. The algorithm used to find the isoelectric line and *R* wave in the averaged beat is identical to that used to find the baseline in the template algorithm (Fig. 3). *ST* slope is determined from a window of five data points centered on the *ST* level measurement location. This five-point window represents a 16 ms region within the *ST* segment.

An on-line display of the beat-by-beat ECG analysis appears on the monitor during the exercise test. The two most current beats are shown, the single beat code and diagnosis are displayed, a contextual diagnosis appears if warranted, *ST* level and slopes are reported and heart rate is given. At the completion of the test a report is printed which contains all of these items plus the correlation coefficient and *RR* interval for each beat. A statistical summary of rhythm appears at the end of the report.

An alternate channel can be selected for subsequent offline processing since all data are retained on hard disk and can be archived on floppy disk. Editing on a beat-by-beat basis is possible for verification of beat classifications. For example, by specifying the beat code for a premature ventricular beat, the operator can examine each instance of a beat so coded, and edit the classification if in error.

Results

To evaluate the arrhythmia diagnostic capability of the software, we processed 13 passages of abnormal rhythm from our tape library acquired earlier from patients in coronary intensive care [9]. These recordings exhibited a variety of complex arrhythmias and contained a large number of abnormal beats. Passages of 1–5 min in length were computer analyzed. There were a total of 2741 beats processed (1005 normal, 1736 abnormal) and of these 2697 (98.4 percent) were correctly diagnosed by computer. Ten of the 44 beats misdiagnosed by computer were due to trigger failure. Since the trigger is a hardware feature of the gamma camera, these errors cannot be corrected by software. The remainder were due to incorrect waveform analysis, particularly fusion beats which were difficult to separate reliably from normals.

25 passages of 1–5 min in length from recordings of 15 subjects during exercise stress tests were processed to test the *ST* measurement algorithm. All but one of our exercise passages had only beats with normal *QRS* morphology. Three patients had significant *ST* segment changes which appeared during their exercise procedure. Measurements of these changes, both level and slope, were computed and reported by the computer system. Following the computer processing, selected segments from each case study were replayed on a strip chart recorder via digital-to-analog (D/A) conversion for validation purposes (Fig. 6). The top channel contains the 16 successive normal beats with embedded abnormal beats (when present) followed by a display of the averaged beat for that segment. The D/A strip also contains an additional channel which depicts the location of the trigger and a beat code indicator. The pulse seen on channel 2 indicates a code of one (normal) if it is depressed from the baseline. Codes of 2 through 8 are displayed by an incremental change of height of the pulse. The width of the pulse reflects the size of the *QRS* window. The trigger location is superimposed on the beat code pulse.

48 exemplary sets containing the beats comprising each 16 beat average with computer diagnosis were randomly selected from the processed studies for cardiologist evaluation. Fig. 7 shows a passage of 18 beats in which a premature ventricular beat and compensatory pause appear. These abnormal beats are excluded from the summed beat on which *ST* measurements are made. *ST* level and

a22/11

#	RR	BEAT CODE	DIAGNOSIS	LEVEL	SLOPE	CORR COEF	RUN RR
162	578	1 NORMAL				+0.98	590
163	573	1 NORMAL				+0.99	589
164	571	1 NORMAL				+0.99	587
165	580	1 NORMAL				+0.98	585
166	572	1 NORMAL				+0.99	584
167	573	1 NORMAL				+0.98	582
168	579	1 NORMAL				+0.98	581
169	467	2 P.V.B.				-0.49	581
170	700	7 CMP.P.				+0.99	581
171	578	1 NORMAL				+0.98	580
172	577	1 NORMAL				+0.99	579
173	572	1 NORMAL				+0.99	578
174	575	1 NORMAL				+0.99	577
175	571	1 NORMAL				+0.98	577
176	564	1 NORMAL				+0.98	575
177	569	1 NORMAL				+0.98	574
178	570	1 NORMAL				+0.99	574
179	566	1 NORMAL		-0.44	0	+0.98	573

Fig. 7. Results of computer analysis of exercise electrocardiogram. Each beat shown on the top channel of the strip chart is classified with a beat code. The listing above reports the beat number, *RR* interval (in ms), beat code and diagnosis, correlation coefficient, and the running *RR* average. Beat number 169 is a premature ventricular beat followed by a beat (170) with compensatory pause. After 16 normal beats are accumulated into an averaged beat, the *ST* measurements of level in millimeters and slope in millivolts/second are computed and reported on the last line of the listing. The format of the lower channel is the same as described in Fig. 6. The eighth and ninth (beats 169 and 170) trigger indicators appear on pulses which reflect beat codes other than 1.

slope are reported upon the acquisition of 16 normals. Fig. 8 shows a passage of all normal beats followed by the reconstructed averaged beat. A diagnostically significant *ST* depression of 0.2 mV is reported as well a negative slope of 2 mV/s.

The original raw data, the computer report, and D/A passages of digitized and averaged beats were submitted to a cardiologist (the fifth author) for determination of agreement between the computer and the observer. The raw data were read by the cardiologist in blinded fashion and *ST* levels and slopes determined in the following manner. The cardiologist selected complexes with stable *ST* segments, read *ST* values from at least three such complexes, and averaged the values. A magnifying glass was used, and a transparent overlay was placed over these complexes to determine slope. *ST* measurements within 0.05 mV of the cardiologist's measurement were considered to be in agreement. The cardiologist and computer were in agreement in 46 of 48 (95.8 percent) of *ST* level measurements and in 44 of 48 (91.7 percent) agreement in *ST* slope measurements in a comparison with the raw data. The cardiologist repeated his measurements on the digitized data reproduced by D/A converter. On these measurements the cardiologist and computer were in agreement in 97.9 percent of *ST* level measurements and 89.6 percent of *ST* slope measurements.

Discussion

The addition of arrhythmia analysis to a computer exercise system provides an improved diagnostic tool in an automated system. The reliability of this *ST* analysis system depends upon the ECG rhythm preprocessing which rejects artifact and arrhythmias. The classification of individual beats allows selective signal averaging which includes only beats with normal intervals and morphology. Measurements derived from the stress test which are indicative of coronary artery disease, e.g., *ST* segment level and slope, are made on a more accurate averaged beat because no abnormal beats are included.

Baseline wander is a major problem in exercise ECG. Skin resistance changes due to perspiration, soft tissue movement, and respiratory effects cause slow drift which makes accurate reading extremely difficult. The *weighted* correlation coefficient solves the baseline wander problem of the exercise ECG but, more important, it requires little computation time. This makes it possible to implement the system in real time without the addition of hardware. Computer processing of a patient who achieved a maximal heart rate of 173 was done without error; thus, fast heart rates are not a problem. The system nicely handles noise and artifact. If a false trigger occurs, the beat is classified abnormal and thus not summed into the averaged beat. Subsequent editing can reclassify it as needed.

#	RR	BEAT CODE	DIAGNOSIS	LEVEL	SLOPE	CORR COEF	RUN RR
522	612	1 NORMAL				+0.98	609
523	612	1 NORMAL				+0.97	610
524	609	1 NORMAL				+0.99	610
525	608	1 NORMAL				+0.99	610
526	607	1 NORMAL				+0.99	610
527	611	1 NORMAL				+0.98	610
528	606	1 NORMAL				+0.99	610
529	607	1 NORMAL				+0.98	610
530	607	1 NORMAL				+0.97	610
531	605	1 NORMAL				+0.98	610
532	597	1 NORMAL				+0.99	609
533	604	1 NORMAL				+0.99	608
534	604	1 NORMAL				+0.95	607
535	598	1 NORMAL				+0.99	607
536	605	1 NORMAL				+0.99	606
537	608	1 NORMAL		-2.00	-2.08	+0.98	606

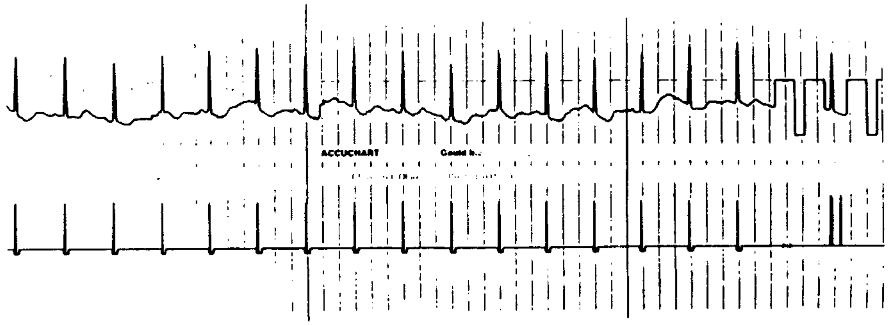


Fig. 8. Results of computer analysis of exercise electrocardiogram. This figure is ordered as in Fig. 7, but the averaged beat is seen on the right-hand side of the tracing. An ST depression of 2 mm (-0.2 mV) is reported on the last line with a negative ST slope of 2.08 mV/s.

The ECG analysis program was designed for incorporation into a gamma camera for use in exercise radionuclide ventriculography. The addition of ECG diagnosis to this device will provide two improvements: 1) control of image acquisition is done by the software in which abnormal beat recognition provides bad beat rejection during imaging resulting in an improved image, and 2) the combined diagnostic capability of the image parameters (ejection fraction and regional myocardial wall motion) and ECG parameters (QRS shape, ST level and slope) is expected to improve sensitivity and specificity in the detection of myocardial ischemia.

Information about the availability of the software can be obtained from the authors.

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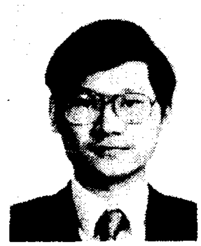

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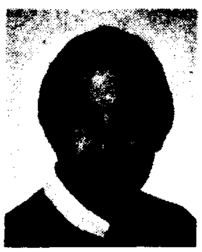

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