A SOFTWARE TRIGGER FOR INTRACARDIAC WAVEFORM DETECTION WITH AUTOMATIC THRESHOLD ADJUSTMENT

Robert MacDonald, Janice Jenkins, Robert Arzbaecher*, Robert Throne

University of Michigan, Ann Arbor, MI
* Illinois Institute of Technology, Chicago, IL

ABSTRACT

A system has been designed for on-line multichannel analysis of intracardiac signals. The system performs automatic waveform detection and interval measurement during cardiac electrophysiologic studies. The system can detect atrial depolarizations (A) on the high right atrial channel (HRA), His bundle depolarizations (H) on the His bundle electrogram (HBE), and ventricular depolarizations (V) on the signal from the right ventricular apex (RVA). In addition, the waveform recognition algorithm recognizes and ignores pacing artifacts. Capture or non-capture is reported during pacing protocols, and the A–A, A–H, H–V, and V–V intervals are presented in real-time.

BACKGROUND

Cardiac electrophysiology studies are an important tool in the management of patients with arrhythmias. These studies involve the insertion of electrode catheters into the viens of the subject. The catheter tips are advanced to specific locations within the heart from which the activation sequence of the heart can be recorded. The heart can be paced through these catheters to determine the patients susceptibility to severe sinus bradycardia or to symptomatic heart block. Pacing can be used to provoke tachycardias and the efficacy of different drugs in suppressing these tachycardias can be evaluated.

During the electrophysiology study, signals are recorded through electrophysiology amplifiers on photographic or strip chart paper. Detailed manual measurement of intervals between waveforms is delayed until after the study, a process requiring 2 to 4 hours [1]. Immediate, automated interval measurement would allow the electrophysiologist to use information gathered during the study to direct its progress, saving time during the study and reducing the possibility that a follow-up study may be required.

Free floating intracardiac catheters encounter many problems which do not arise with QRS detectors on a fixed surface lead. These include waveform size and shape variability, varying contributions from extraneous signals (i.e. V waveforms on HRA), and multiphasic waveforms. For effective interval measurement of intracardiac waveforms, these problems must be dealt in real-time. Previous techniques have worked only in restricted cases or not in real-time [2][3].

METHODS AND MATERIALS

Hardware

The hardware platform for the data acquisition and waveform recognition algorithms was a Dell System 325 computer. This computer includes a 25 MHz 80386 processor, a high speed 90 MB hard disk drive, and a 720 by 350 pixel monochrome graphics display. Additional hardware includes a high speed waveform display board from DATAQ Instruments, Inc., and a Labmaster DMA data acquisition board from Scientific Solutions, Inc.

Program Organization

Four channels of data are acquired in real-time: three intracardiac and one surface. The data is doubled buffered; while one buffer is being filled, the other is being stored to disk. The input buffer is examined by the recognition algorithms to determine the timing of cardiac events. The time of occurrence and type of each event is stored to disk in the order of occurrence.

For testing purposes, the program is designed so that data can either be acquired via the A/D converter, or read from disk. In either case, the reading of data is synchronized by the real-time clock on the Labmaster DMA so that real-time operation can be verified.

Atrial and Ventricular Event Recognition

Our algorithm for recognition of atrial and ventricular events is designed to allow for variable waveform size, multiphasic waveforms, baseline wander, and the existence of waveforms representing events other than the event of interest. This flexibility is accomplished by using a multi-stage triggering algorithm incorporating bandpass filtering, an automatically adjustable threshold, and a blanking interval for suppression of multiple triggers on a single waveform. A block diagram of the algorithm is given in figure 1.

The first stage of the trigger is a bandpass filter. The filter is designed to suppress baseline wander, high frequency noise, and pacing artifacts, and to enhance the depolarization waveform relative to signals originating in regions distant from the recording site. A good passband to accomplish these goals was found to be 20–60 Hz, which passes the largest components of the waveforms and does not significantly affect waveform timing. The filter used is a 2 pole digital bandpass filter. The digital filter is derived using a

0276-6574/90/0000/0167$01.00 © 1990 IEEE

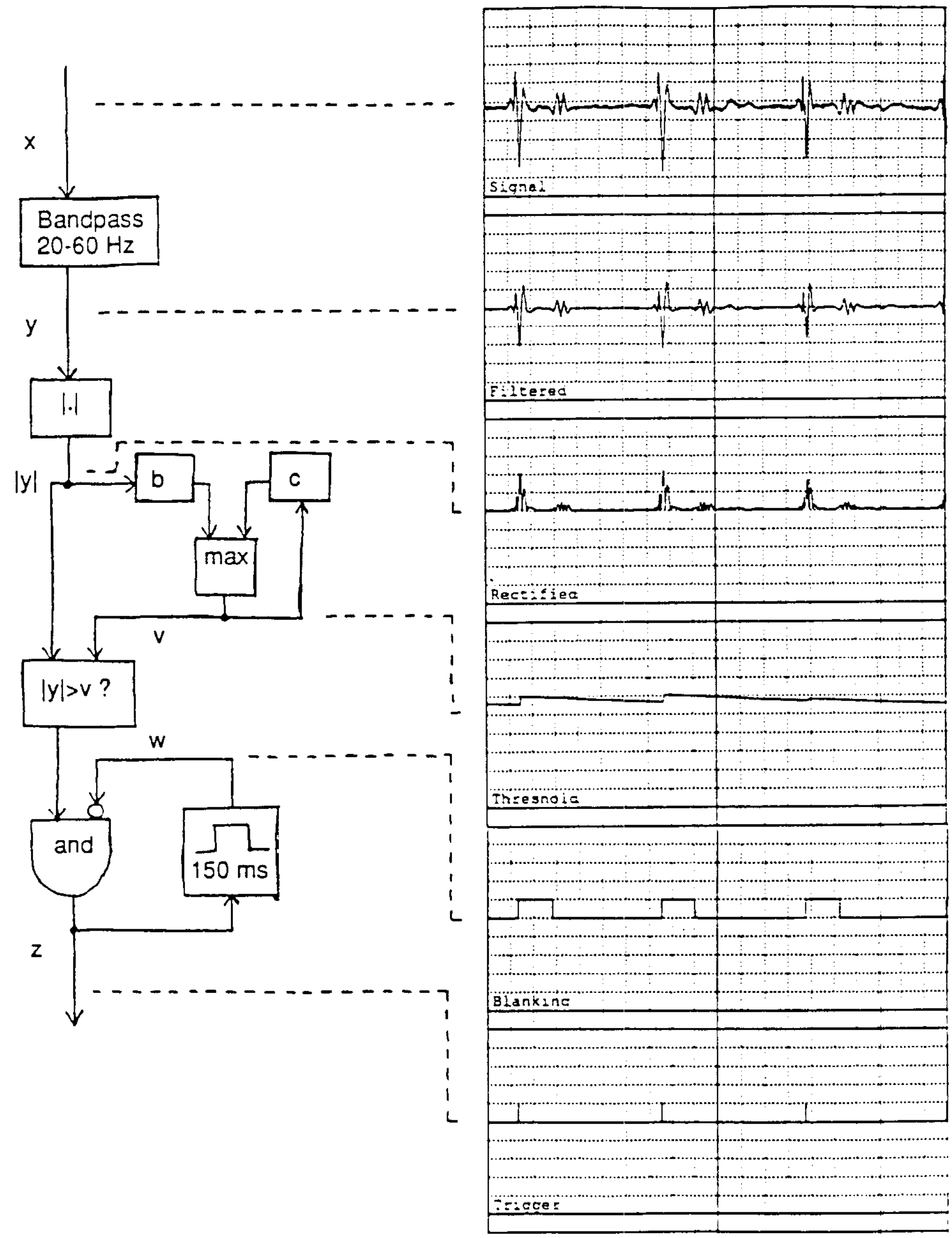

Figure 1: Block diagram of the waveform detection algorithm. The output at each stage of the algorithm is shown in the tracing on the right.

	Protocol	Patients	Waveforms	Error			
				FP	FN	± 10 ms	± 20 ms
1.	Atrial fixed rate pacing	6	1260	1	7	12	2
2.	Atrial single extrastimulus	7	992	1	3	54	0
3.	Atrial extrastimulus in paced rhythm	6	1346	2	13	22	6
4.	Ventricular fixed rate pacing	4	553	0	1	13	2
5.	Ventricular single extrastimulus	4	812	13	9	24	5
	Overall	8	4963	17	33	125	15
				0.3%	0.7%	2.5%	0.3%

Table 1: Triggering algorithm performance during selected electrophysiologic protocols.

bilinear transform of the analog filter prototype

$$H(s) = \frac{1}{(1 + s/w_L)(1 + w_H/s)}$$

giving

$$H(z) = \frac{a_0(1 - z^{-2})}{1 + a_1 z^{-1} + a_2 z^{-2}}$$

where

$$\begin{aligned} a_0 &= \frac{w_L}{(w_L + 1)(w_H + 1)} \\ a_1 &= \frac{w_L - 1}{w_L + 1} + \frac{w_H - 1}{w_H + 1} \\ a_2 &= \left(\frac{w_L - 1}{w_L + 1}\right)\left(\frac{w_H - 1}{w_H + 1}\right) \end{aligned}$$

and

$$w_L = \tan \pi f_L / f_S, \quad w_H = \tan \pi f_H / f_S$$

The resulting difference equation is

$$y_i = a_0(x_i - x_{i-2}) - a_1 y_{i-1} - a_2 y_{i-2}$$

Each of the coefficients, a_i, is scaled so the the final real-time filter requires only 3 integer multiplies and 3 integer additions.

The second stage of the trigger is a threshold comparison of the rectified filtered signal with an automatically adjusting threshold. The threshold is adjusted by two mechanisms. Whenever the threshold falls below a certain fraction, b, of the rectified signal, $|y|$, the threshold is reset to the level, $b|y|$. The second mechanism is an exponential decay of the threshold over time. The decay is accomplished by reducing the threshold, v, to an amount proportional to the threshold level, cv. The constant, c, is determined from the decay half-life, t_d, by

$$c = 2^{-1/(t_d f_S)}$$

The threshold update function in its entirety is then

$$v_i = \max(b|y_i|, cv_{i-1})$$

The two threshold update mechanisms work together to control the sensitivity of the trigger and its flexibility in recognizing waveforms of varying amplitudes. The decay half-life, t_d (and therefore c), controls the rapidity of amplitude variation which can be accommodated — the larger the t_d, the smaller the expected rate of variation. The fraction, b, in combination with t_d, controls the short term sensitivity to rapid changes in amplitude. A large b will result in the ignoring of relative large ventricular artifacts at the same time that a samll t_d accommodates large changes in the atrial waveform. We have found that on the atrial channel $t_d = 1$ s and $b = 0.5$ represent a good compromise between flexibility and over-sensitivity. On the ventricular channel, the lack of atrial artifacts allows the use of $b = 0.4$ for increased sensitivity to change.

The third and final stage of the trigger is the post-triggering blanking interval. This is represented in figure 1 as a pulse of fixed duration which is initiated by the threshold stage output and which inhibits that output. It is implemented in software by a counter which is set to the number of samples to blank at each trgger. It is decremented once per time step and inhibits all further detection until it reaches 0, at which time the decrementing ceases. The purpose of the blanking interval is to prevent multiple triggers on a single multiphasic waveform. The interval we have used is 150 ms.

Stimulus Recognition

A necessary feature of any intracardiac waveform recognition algorithm is the ability to recognise stimulus artifacts and to differentiate them from normal depolarizations. A modification to the triggering algorithm, in which the bandpass filter is replaced with a 400 Hz highpass filter and the exponential threshold decay is removed, allows detection of stimulus spikes alone. The location of the stimulus spikes can then be used to inhibit triggering during the interval over which the stimulus occurs.

His Bundle Depolarization Recognition

Recognition of the His bundle depolarization introduces problems not encountered in the recognition of other cardiac events. Far from being recognizable by its size, the H waveform can be one of the smaller waveforms of the His bundle electrogram. Filtering alleviates this problen only slightly. Fortunately, the time of occurrence of non-H waveforms can be inferred from the presence of corrosponding waveforms on other leads. We use this additional information to define a window, following the A and preceding the V, during which recognition of the H is allowed. The structure of the algorithm differs from the basic waveform detection algorithm only in that the output of the filter stage is set to 0 outside of the defined window.

Validation of Measurements

The waveform times of occurrence measured by the triggering algorithm were validated by verifying the correctness of the corrosponding interval measurements. The following protocol was established for validation of the computer measurement of intervals. Analog representations were produced on a strip chart recorder of a single (atrial or ventricular) channel and of a simultaneous trigger channel produced by the computer recognition of waveforms. Paper speed was 100 mm/s or 125 mm/s depending upon the available equipment. At 100 mm/s, 1 mm is equivalent to 10 ms, and at 125 mm/s, 1 mm is equivalent to 8 ms. Visual accuracy was considered to be approximately 0.5 mm giving a human resolution of 4–5 ms.

The overreader, using calipers, chose the first fast deflection of significant amplitude of each pair of sequential depolarizations and aligned the tips of the calipers with these locations. The interval spanned by the two trigger signals was examined for a match (within 10 ms) or a mismatch. In the case of a mismatch, the size of the error was entered onto the strip and later tabulated for each passage under analysis. Failure to recognize waveforms (false negatives, FN) and recognition of waveforms when none were present (false positives, FP) were also charted. A FP or FN was considered to constitute *two* errors in interval measurement. (In the case of a FP, there are two *wrong* intervals produced in place of a correct one; in the case of a FN, there is one wrong interval produced in place of two *correct* ones.) The, the number of FP and FN was doubled and summed with the remaining timing errors.

When a stimulus artifact was present, the caliper measurement was made to the first major deflection falling at least 30 ms after the stimulus artifact. This allows for the signal to return to baseline and corresponds to the stimulus blanking interval used by the computer algorithm.

RESULTS

Test Population

From our data base of over 50 patient recordings, 8 patients were chosen at random for evaluation of the triggering algorithm. A total of 28 passages representing a mix of 5 electrophysiology protocols were selected for detailed analysis. Data passages represented 2936 s of single channel recordings for a total of 4963 waveforms submitted for manual expert review. These patients constituted a random sampling of our data base. Passages were selected to include a majority of electrophysiology protocols within an individual patient procedure whenever possible. Table 1 summarizes our results.

Results in Stimulation Protocols

Protocol 1: Atrial Fixed Rate Pacing. The overall error rate in this protocol was 2.2%. Of this error, 1% constituted inaccurate measurement, and the remainder was due to 1 FP and 7 FN.

Protocol 2: Atrial Single Extrastimulus. The overall error rate in this protocol was 6.3%. The error rate was heavily weighted by a single patient who exhibited a very labile atrial waveform morphology causing the triggering algorithm to trigger alternately on different peaks of the waveform. The error rate without this patient was 2.2%.

Protocol 3: Atrial Extrastimulus in Paced Rhythm. The overall error rate in this protocol was 3.9%. Again, the bulk of the error for this protocol is attributable to a single patient.

Protocol 4: Ventricular Fixed Rate Pacing. The overall error rate in this protocol was 2.7% and included 1 FN.

Protocol 5: Ventricular Single Extrastimulus. The overall error rate in this protocol was 8.4%.

Summary of Results

In summary, 28 passages with a total of 2936 s of data from 8 patients were manually overread. These passages contained 4963 waveforms for which computer interval measurements were compared with those of a human expert. There were 13 instances of false recognition of waveforms for a FP rate of 0.3%. There were 37 instances of failure to recognize waveforms for a FN rate of 0.7%. The numbers of FP and FN (50) were doubled since each represented 2 erroneous interval measurements. Of waveforms correctly found, 125/4963 (2.5%) had measurement errors exceeding 10 ms; 17/4963 (0.3%) had measurement errors exceeding 20 ms. The total error count was 225/4963 for an overall error rate of 4.5%.

Results for Recognition of His Bundle Depolarization

The algorithm for recognition of His bundle depolarization was tested for false positives and false negatives only. Six passages totaling 263 seconds were processed. The passages were selected from sinus rhythm and atrial stimulation protocols. The passages were selected in which accurate visual identification of the His waveform was possible for the major part of each passage. Errors totaled 9 FP and 9 FN.

DISCUSSION

The triggering algorithm has been demonstrated to perform extremely well in event detection. The algorithm has a very low false positive and false negative rate (0.3% and 0.8% respectively). In cases of correct waveform recognition the algorithm measured 95.5% of the intervals to within 10 ms. In a small number of patients whose waveform morphology changed from beat to beat the trigger location within the waveform also varied, sometimes alternating between different peaks of a multiphasic waveform. Even in these cases the errors were within 20 ms.

Detection and interval measurement are performed online on three simultaneously acquired intracardiac signals (HRA, HBE, and RVA) on a fast personal computer. Immediate reporting of A–A, A–H, H–V, and V–V intervals is provided. Concurrent recognition of pacing artifacts and distinction these artifacts from waveforms of interest (A, H, and V) allows reporting of capture or non-capture. Such a feature yields data from conduction curves during atrial extrastimulus protocols performed for A–V conduction studies.

REFERENCES

1. Ross DL, Farre J, Bar FWHM, et al: Comprehensive electrophysiologic studies in the investigation of documented or suspected tachycardias — time, staff, problems, and cost. *Circulation* 61, pp. 1010–1016, 1980.
2. Antolini R, Kirchner M, Mongera A, et al: On-line measurement during invasive cardiac electrophysiology testing. *PACE* 11, pp. 33–36, 1988.
3. Gillette PC, Garson A, Zinner A, Kugler JD: Continuous computer automated measurement of electrophysiologic data during cardiac catheterization. *Computers in Cardiology*, pp.391–393, 1980.

This work was partially supported by NIH grant 1R43HL41406 and NSF grant EET-835125.