Physiological measurement

Beat-to-beat detection of His-Purkinje system signals using adaptive filters

H. A. M. Al-Nashash **S. W. Kelly**

Medical Electronics Research Group, Electronic Engineering Laboratories, University of Kent at Canterbury, Canterbury, Kent CT2 7NT, UK

D. J. E. Taylor

Cardiac Department, Kent & Canterbury Hospital, Canterbury, Kent, UK

Abstract—*Noninvasive techniques to record the activation wave from the His-Purkinje system have so far depended largely on signal averaging. Although this approach produces representative signals, any beat-to-beat variations are removed by the averaging process. These beat-to-beat variations are important in the diagnosis of many heart abnormalities, particularly arrhythmias. The paper describes an experimental system which can detect His-Purkinje system electrograms at the body surface while preserving beat-to-beat variations. The system uses a number of different techniques, but an important feature is the use of an adaptive filter to reduce additive noise. The experimental system is not real-time, as the adaptive filtering is performed offline using software, but a real-time hardware implementation is quite feasible. The system's ability to detect beat-to-beat variations has produced a number of interesting results, which are discussed. These include a cyclic variation in a normal subject, believed to be due to the modulating effects of breathing, and signals from a patient suffering from second degree A-V block.*

Keywords—*Adaptive filters, Arrhythmias, His-Purkinje signals, Noninvasive techniques, Signal processing*

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1 Introduction

ELECTROCARDIOGRAPHY HAS become established as a powerful diagnostic tool using the heart's electrical activity as a measure of its muscular activity. However, many heart defects occur not in the myocardium but in the specialised conduction system. Normal atrioventricular conduction of electrical events depends first on the generation of an impulse in the sinus node. The sequential propagation of this impulse begins in the atria and spreads to the atrioventricular (A-V) node, from whence it travels down the bundle of His, the bundle branches, the Purkinje ramifications and finally to the ventricular myocardium (NETTER, 1969; FURNESS, 1975).

Because of the relatively small mass of the specialised conduction system, signals of the order of only a few microvolts are produced on the body surface. Hence an invasive measurement technique was introduced by SCHERLAG *et al.* in 1969. A catheter electrode was introduced into the heart via the femoral vein and positioned close to the His bundle to record the electrical activity of the conduction system of the heart. The waveforms produced were called the 'His bundle electrogram' (HBE) (SCHERLAG *et al.*, 1969). Analysis of the HBE is of great value in studying the electrophysiology of the heart, investigating the effects of drugs, localising A-V conduction defects in all degrees of A-V block and investigating the mechanisms of conduction arrhythmias (FURNESS, 1975; HAFT, 1973; ROSEN, 1971; SCHERLAG *et al.*, 1969).

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While this technique has provided a good understanding of the physiology and various pathologies of the specialised conduction system, its invasive nature imposes limitations on its clinical use. The principal problems are the high cost of carrying out such invasive tests, the risk incurred to the patient and the limited recording field of the electrode. These limitations have prompted various research studies into possible noninvasive techniques. BERBARI *et al.* (1973) and FLOWERS *et al.* (1974), working independently, reported the detection of potentials generated by the canine conduction system, using body surface electrodes, high amplification and signal averaging. In this way the desired signal, which is recurring but corrupted by random noise, is enhanced by the resulting improvement in signal-to-noise ratio (SNR). More recent work has reported techniques to improve the detection by concentrating on such areas as electrode positioning (BERBARI *et al.*, 1983; 1986; PEPER *et al.*, 1983; FLOWERS *et al.*, 1974), triggering (FURNESS *et al.*, 1975), bandwidth (BERBARI *et al.*, 1979), atrial interaction (PEPER *et al.*, 1982; 1985; ROS *et al.*, 1981) and beat-to-beat detection (FLOWERS *et al.*, 1981; 1982).

Only limited success has been achieved and there is still no accepted clinical tool, based on averaging, for the detection of the His-Purkinje System (HPS) signals. Furthermore the wide variety of recording parameters, such as bandwidth and electrode placement, has led to serious differences among reported results and waveforms. This is shown by Berbari (BERBARI *et al.*, 1979; 1983).

However, two studies (BERBARI *et al.*, 1981; BONES *et al.*, 1982*a*) have reported that the surface His-Purkinje system

waveforms resemble a ramp-shape waveform beginning at the time of His bundle activation and continuing until the earliest ventricular muscle activation. These results are corroborated by a first-order model used to predict the shape and morphology of the HPS signals at the body surface, given specific electrode placement. BERBARI *et al.* (1983) have shown that the spike-like signals reported by many investigators were due to the high-pass filtering of the ramp-like waveforms. This conclusion was supported by another study by PEPER *et al.* (1984).

The inability of averaging techniques to provide beat-to-beat information about the status of the conduction system, especially during the study of arrhythmias, is now the main obstacle to the acceptance of the technique as a clinical tool. A further remaining problem is that triggering instability can lead to an undesirable smoothing of the waveform (ROS *et al.*, 1981).

Our aim has been to develop a new, noninvasive technique which does not rely on averaging, is capable of detecting cycle-to-cycle variations and is suitable for routine clinical use.

Technically, the problem is essentially one of detecting an extremely small signal in the presence of noise. There are many types of noise corrupting the desired HPS signal, including muscle noise, electrode noise, mains interference and both noise and drift in the measuring system electronics. The measuring system can also cause contamination of the HPS waveform due to phase shifting of the atrial components into the HPS region if filters having nonlinear phase characteristics are used (BERBARI *et al.*, 1979; 1981). As these various noise sources have widely differing characteristics, no one technique is optimal for dealing with all of them. Thus our research uses a multistage system, although it will be seen that adaptive filters (implemented in software) play a major role in extracting the desired signal.

Adaptive filters allow statistically stationary or nonstationary signals to be enhanced against additive noise. Two or more input channels, containing correlated signal components but uncorrelated noise components, are required. The output is then a best least-mean-square estimate of the original signal (FERRARA and WIDROW, 1981). As adaptive filters are, to a certain extent, 'self-designing', a complete *a priori* knowledge of the statistics of the signals is not required. Furthermore, an optimal improvement in SNR is achieved for signals when spectra overlap the noise spectra (as is the case for HPS signals) and relatively simple computations are involved.

Adaptive filters have been used successfully in other biomedical applications. FERRARA and WIDROW (1982) reported their use in the enhancement of low-amplitude foetal electrograms obscured by the presence of excessive noise, including the maternal ECG.

2 Experimental system

The system for the noninvasive detection of the HPS signals comprises many sections, as shown in the block diagram of Fig. 1. These may be grouped into five main stages, which are discussed below.

2.1 *Patient preparation and electrode placement*

This is an important stage because there are three sources of noise which can degrade the desired signal. These are motion artefacts, impedance effects of the outermost layer of the skin and 50 Hz interference. To ensure a good recording, the following steps are necessary (HUHTA and WEBSTER, 1973; OTT, 1976; WEBSTER, 1984):

(i) vigorous rubbing with sandpaper to minimise the skin impedance and short-circuit the skin potentials
(ii) the use of silver/silver chloride electrodes to minimise the effects of motion artefacts
(iii) making leads as short as possible and of the same length
(iv) screening all cables and keeping them as close to the patient as possible
(v) eliminating all earth loops
(vi) earthing screens at one point only.

Two sets of bipolar electrodes were used to record the HPS signals. The V6R–V6L was chosen for both sets, with the positive terminals positioned at the V6L and the negative terminals positioned at the V6R. The two sets were longitudinally displaced by a few centimetres to ensure decorrelation of random muscular noise, for reasons that will be apparent later on. The main two reasons for choosing this position were:

(*a*) electrodes are less sensitive to chest wall movements, especially with female patients
(*b*) signals represent a global view of the HPS (BERBARI *et al.*, 1979).

2.2 *Preprocessing*

The Analog Devices AD624 low-noise instrumentation amplifier is used for the first stage. This has an input impedance of $10^9\,\Omega$, a common mode rejection ratio (CMRR) of 130 dB at a gain of 500 with 1 kΩ impedance imbalance and a noise voltage of $2{\cdot}5\,nV\,Hz^{-1/2}$ with respect to the input. This has the advantage that no external components are required, preventing additional noise

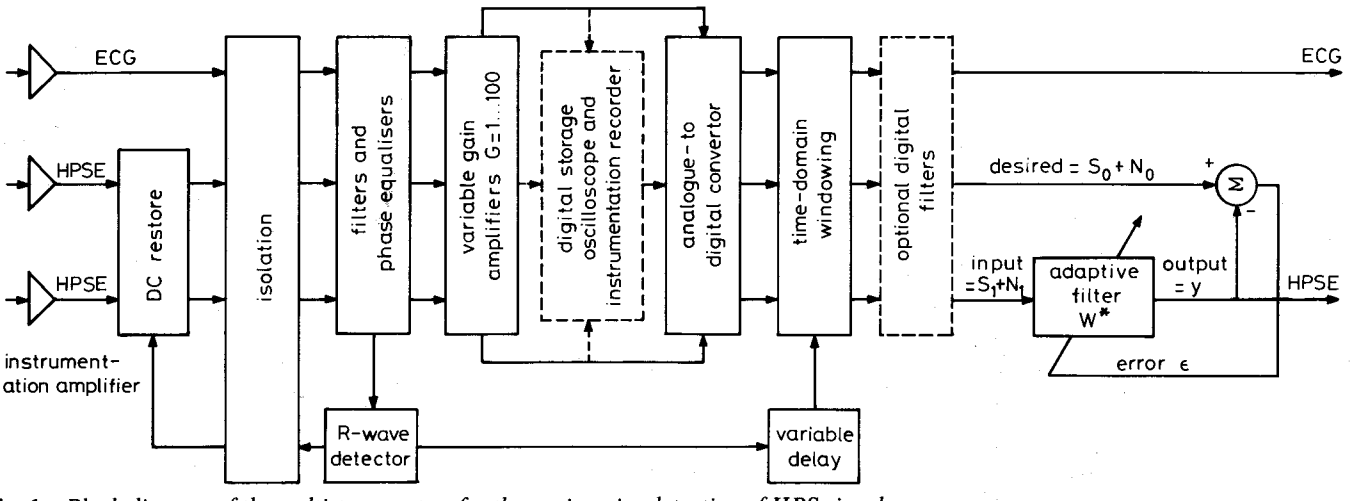

Fig. 1 *Block diagram of the multistage system for the noninvasive detection of HPS signals*

from being introduced by such components. The amplifier is battery powered to ensure minimum noise at this very important stage. Three of these amplifiers are used to provide parallel channels, two for the HPS signals and the third for the ECG, which is used as a reference timing signal and to obtain an R-wave trigger (the reasons for which will be apparent later on).

Because the HPS signals are extremely small, a high degree of amplification is required. However, DC shifts caused by electrode movement, drift and amplifier offset can be significantly greater than the signal, resulting in saturation. High-pass filtering was not considered to be a satisfactory solution to this problem, as it removes the low-frequency components of the signal. Hence, an automatic DC restore circuit has been designed, which reduces the DC shift to an acceptable level. This is achieved by low-pass filtering the signal at the output of the instrumentation amplifier and then sampling it at a point just before the P-wave. This results in a measure of the DC component which is stored (using a sample-and-hold) during the period of HPS activity and subtracted from the original signal using a differential amplifier.

An adjustable R-wave detector comprising a bandpass filter, comparator and non-retriggerable monostable provides a timing reference, from which the sample signal for the DC restore is derived. The R-wave detector also provides a signal for time gating in the interface stage (see later).

For purposes of patient safety, Analog Devices AD204 isolation amplifiers are used to isolate the battery-powered first stages (which are connected to the patient) from the subsequent mains-powered equipment. These devices provide an isolation of ± 1000 V peak and a CMRR of 160 dB at unity gain.

In addition to the minimisation of electrical noise by careful choice of components, much consideration has also been given to the reduction of possible 50 Hz interference. Thus the preprocessing section is fully screened, in common with the input leads. Good earthing techniques are also vitally important and separate signal and digital returns are used, both returning to a single point. Printed circuit boards employing earth planes are used throughout the preprocessing stage.

The whole of the preprocessing stage was tested for random noise and 50 Hz interference effects. After placing the whole stage in the area where clinical measurement would be made and short-circuiting the input leads, the random noise level was found to be $2{\cdot}5\,\mu V\,Hz^{-1/2}$ at a gain of 500 and over a bandwidth of 0·02–250 Hz. The 50 Hz component was $23{\cdot}8\,\mu V$ with respect to the output at the same gain.

2.3 *Signal conditioning*

The output HPS signals from the preprocessing stage are of the order of a millivolt with a DC offset which is of the same order as the signal amplitude and a bandwidth of 50 kHz. Although electrode noise, electronic noise and mains interference have been minimised, these signals will inevitably contain significant amounts of noise, primarily muscle noise. The function of the signal conditioning stage is to prepare the signals for subsequent digitisation or recording, as follows:

Low-pass filtering is employed for two reasons: first to avoid aliasing in the subsequent digitisation and secondly to reduce the high frequency noise. Although a linear phase response is required to prevent overlap between the P-wave and the HPS signals, the linear phase Bessel filter was rejected in favour of a maximally flat Butterworth design. The phase nonlinearity is corrected by the use of special phase equalisation circuits. The filters used are second order, with a cutoff frequency of 500 Hz. The phase equalisation is achieved by the use of special circuits which provide a phase delay as a function of frequency but a constant (unity) gain (VAN VALKENBURG, 1982).

Optional high-pass filters are also included to investigate their effect on the signal. These filters are also a second-order Butterworth design, with a cutoff frequency of 0·1 Hz.

Variable gain amplifiers are used to raise the amplitudes of the signal to a level suitable for digitisation or recording (a few volts). The gains are continuously variable over the range 1 to 1000, making the total system gain range (preprocessing + signal conditioning) 500 to 500 000.

Finally, a DC control is provided to remove any DC offset introduced in the signal conditioning stage.

2.4 *Interface*

Two options are available at this stage. Following the conditioning stage, data can either be stored on an instrumentation tape recorder (TEAC MR-30) and subsequently played back for further processing, or data can be directly fed to a three-input analogue-to-digital convertor (A/D) to be stored using a microcomputer for further offline processing. Because of the automatic control of the DC level, it is possible to achieve very low offset. Hence, signals of several volts amplitude can be obtained without saturation. This means that high resolution digitisation is not necessary. The A/D has an 8-bit resolution with a range of 5 V. The sampling rate is 1·5 kHz for the HPS signals, whereas 250 Hz is considered sufficient for the ECG.

To achieve the optimum conversion accuracy, the A/D convertors are constructed using PCBs incorporating earth planes and powered by batteries. Opto-isolators prevent noise from the microcomputer's digital circuitry affecting the signals.

It is evident that to record the entire continuous signal is a waste of memory and processing time. Hence time gating is used to store the P-R interval only. This is achieved by software using a cyclic array. Thus on the arrival of the R-wave trigger, the last 256 samples only are transferred to the main array. The last 256 samples are equivalent to a period of approximately 170 ms, which contains the H-wave under most conditions.

2.5 *Computer processing*

A number of programs have been written and developed which enable the signals to be analysed, processed or displayed. A menu is provided for the user, allowing access to any of the following programs:

1 input new data
2 display waveform
3 produce numeric listing
4 perform frequency analysis (BRIGHAM, 1974)
5 perform stationarity check (PAPOULIS, 1984)
6 perform low-pass and high-pass filtering (RABINER and GOLD, 1975; WILLIAMS, 1986)
7 compute statistics (PAPOULIS, 1984)
8 perform adaptive signal enhancement.

After storing data using option 1, the normal mode of operation is to perform adaptive signal enhancement using option 8 to reduce the random noise.

Quasiperiodic signals such as the HPS signal corrupted by additive random noise (muscle noise) can be recovered and enhanced by using adaptive filters (WIDROW, 1966; WIDROW *et al.*, 1975; WIDROW and YELDERMAN, 1983;

WIDROW and STEARNS, 1985). As shown in Fig. 1, two inputs are normally required. The first input contains the signal S_0 and additive nosie N_0. The second input contains a correlated signal S_1 and additive noise N_1. N_0 and N_1 must not be correlated with each other or with either signal. This can be achieved in two different ways:

(*a*) By locating the electrodes detecting S_0 and S_1 a few centimetres apart, thus ensuring that N_0 and N_1 are decorrelated. (S_0 and S_1 need not be of the same waveshape (FERRARA and WIDROW, 1981).) This method is not always reliable as the degree of decorrelation may not be known.

(*b*) $S_0 + N_0$ can be obtained by delaying $S_1 + N_1$, so that only one signal is required. Provided that the delay is sufficiently long, S_0 will still be correlated with S_1 but N_0 will be uncorrelated with N_1. A delay equal to half the length of the adaptive filter has the extra advantage of obtaining the performance that would be obtained if the adaptive filter was not causal (WIDROW *et al.*, 1975). Also the method is easier and of lower cost as one of the analogue channels is no longer required.

Both of the above techniques have been investigated, but the latter method is now used exclusively.

The adaptive filter adjusts its own impulse response iteratively through a least mean squares algorithm developed by WIDROW (1966) based on the method of steepest descent, which responds to the error signal ε (the difference between the output y and the desired response d). After convergence, the power of the error ε is minimised and the output y is then a best least mean square estimate of $d = S_0 + N_0$ (WIDROW *et al.*, 1975). y is then the least mean square estimate of S_0, as N_0 is uncorrelated with $x = S_1 + N_1$.

Once the adaptive filter has converged, it represents an unconstrained Wiener solution to stationary stochastic inputs (WIDROW *et al.*, 1975). Under this steady-state condition, we can describe the effect of the filter on the signal and noise spectral power densities by means of the z-domain, unconstrained, optimal Wiener transfer function:

$$W^*(z) = \frac{\text{cross power spectrum between the input signal and desired response}}{\text{power density spectrum of the input signal}}$$

From this equation, it can be shown that the frequency domain output is given by

$$|y(\omega)| = \left|\frac{\rho_{in}(\omega)}{1 + \rho_{in}(\omega)}\right| \cdot |x(\omega)|$$

where ρ_{in} is the signal-to-noise spectral power density of d and $z = e^{j\omega}$.

The above formula shows that when the input power spectral density ρ_{in} is large, the transfer function increases the gain so as to pass most of the signal power in an optimum fashion. If the ρ_{in} is low, the filter adjusts its own impulse response in such a way that it suppresses these signal components leading to a small signal at the output. However, in regions where the signal and noise exhibit spectral overlap, a tradeoff takes place where signal and noise components are affected, again in an optimum manner.

The following brief description of the least mean square (LMS) adaptive filter is based on a paper by WIDROW *et al.* (1975). For a more complete description the reader is advised to refer to that work and others, such as WIDROW (1966), WIDROW *et al.* (1975) and WIDROW and STEARNS (1985).

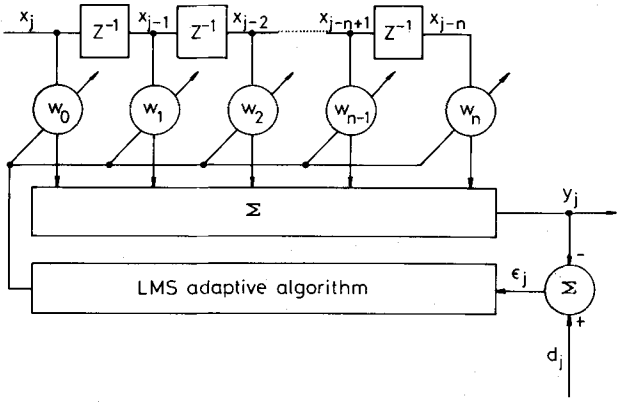

Fig. 2 *Adaptive linear combiner*

The fundamental component of all adaptive systems is the adaptive linear combiner as shown in Fig. 2.

The input signal vector $\boldsymbol{X}_j$ is defined as

$$\boldsymbol{X}_j = [x_j \quad x_{j-1} \quad \ldots \quad x_{j-n}]^T \tag{1}$$

The weighting coefficients are adjustable and the weight vector $\boldsymbol{W}$ is

$$\boldsymbol{W} = [W_0 \quad W_1 \quad \ldots \quad W_n]^T \tag{2}$$

The output y_j is equal to the inner product of $\boldsymbol{X}_j$ and $\boldsymbol{W}$

$$y_j = \boldsymbol{X}_j^T \boldsymbol{W} = \boldsymbol{W}^T \boldsymbol{X}_j \tag{3}$$

Assume that the input signal and the desired response (an externally supplied input, sometimes called the 'training signal') are stationary ergodic processes.

Then the error ε_j is defined as the difference between the desired response d_j and the actual response y_j

$$\varepsilon_j = d_j - \boldsymbol{X}_j^T \boldsymbol{W} = d_j - \boldsymbol{W}^T \boldsymbol{X}_j \tag{4}$$

The square of this error

$$\varepsilon_j^2 = d_j^2 - 2d_j \boldsymbol{X}_j^T \boldsymbol{W} + \boldsymbol{W}^T \boldsymbol{X}_j \boldsymbol{X}_j^T \boldsymbol{W} \tag{5}$$

The mean square error (MSE) is defined by

$$\xi = E[\varepsilon_j^2]$$

$$\xi = E[d_j^2] - 2E[d_j \boldsymbol{X}_j^T]\boldsymbol{W} + \boldsymbol{W}^T E[\boldsymbol{X}_j \boldsymbol{X}_j^T]\boldsymbol{W} \tag{6}$$

The correlation matrix at discrete time j, $\boldsymbol{R}_j$, is defined by

$$\boldsymbol{R} = E[\boldsymbol{X}_j \boldsymbol{X}_j^T] \tag{7}$$

and the cross-correlation vector $\boldsymbol{P}$ is defined by

$$\boldsymbol{P} = E[d_j \boldsymbol{X}_j] \tag{8}$$

Therefore eqn. 6 becomes

$$\xi = E[d_j^2] - 2\boldsymbol{P}^T \boldsymbol{W} + \boldsymbol{W}^T \boldsymbol{R} \boldsymbol{W} \tag{9}$$

It can be seen from eqn. 9 that, with stationary ergodic inputs, the MSE performance function is a quadratic function of weights—a paraboloidal 'bowl' that has a unique minimum point for

$$\boldsymbol{W} = \boldsymbol{W}^* = \boldsymbol{R}^{-1} \boldsymbol{P} \tag{10}$$

Eqn. 10 is a matrix form of the Wiener-Hopf equation.

The exact least mean square solution is complex from the computational point of view because it needs extensive calculations of the correlation matrices, matrix inversion, ... etc. The least mean square (LMS) algorithm of Widrow and Hoff, based on the method of steepest descent, provides a practical method of finding a close approximation solution to eqn. 10 as shown below.

According to the steepest descent method

$$\boldsymbol{W}_{j+1} = \boldsymbol{W}_j - \mu \nabla_j \tag{11}$$

where

$\boldsymbol{W}_{j+1}$ is the next weight vector
$\boldsymbol{W}_j$ is the present weight vector
μ is a factor which controls the stability and the rate of convergence
∇_j is the true gradient of the jth iteration

By differentiating eqn. 6 with respect to $\boldsymbol{W}_j$ we obtain

$$\nabla_j = -2E[d_j \boldsymbol{X}_j] + 2E[\boldsymbol{X}_j \boldsymbol{X}_j^T]\boldsymbol{W} \tag{12}$$

The LMS algorithm uses an estimate of the gradient in which the expected values of the random variables are replaced by the instantaneous values:

$$\hat{\nabla} = -2d_j \boldsymbol{X}_j + 2\boldsymbol{X}_j \boldsymbol{X}_j^T \boldsymbol{W}_j \tag{13}$$

i.e. $\hat{\nabla} = -2\varepsilon_j \boldsymbol{X}_j$

Therefore eqn. 11 becomes

$$\boldsymbol{W}_{j+1} = \boldsymbol{W}_j + 2\mu\varepsilon_j \boldsymbol{X}_j \tag{14}$$

It is clear that this algorithm does not require explicit measurements of correlation functions, nor does it involve matrix inversion.

When the parameter μ is greater than zero but less than the reciprocal of the largest eigenvalue λ_{max} of the matrix $\boldsymbol{R}$, the weight values generated by the LMS algorithm will converge to the optimum Wiener weight vector

$$\frac{1}{\lambda_{max}} > \mu > 0 \tag{15}$$

As the eigenvalues of matrix $\boldsymbol{R}$ are usually unknown, we generally use the total input signal power to the filter, $tr[\boldsymbol{R}]$, in place of λ_{max}.

Therefore eqn. 15 becomes:

$$\frac{1}{tr[\boldsymbol{R}]} > \mu > 0 \tag{16}$$

Usually, μ is chosen to be in order of one-tenth of the upper bound given (WIDROW and STEARNS, 1985).

As the weight vector converges to the Wiener solution, the mean square error decreases as a sum of exponentials to the minimum value. The time constants of these exponentials are expressed as a number of iterations of the adaptive process rather than in units of time. A general expression for the time constant is

$$\tau_{MSE} = \frac{n+1}{4\mu tr[\boldsymbol{R}]} \tag{17}$$

Noisy gradient estimates of the MSE are obtained after convergence. The excess MSE divided by the LMS error of the Wiener solution is termed the misadjustment M. An approximate misadjustment is defined as

$$M = \mu tr[\boldsymbol{R}] \tag{18}$$

The minimum number of weights n_{min} is related to the signal bandwidth and to the required frequency resolution of the filter:

$$n_{min} = \frac{2 \times \text{total signal bandwidth}}{\text{frequency resolution of the filter}} \tag{19}$$

Eqns. 16–19 form the basis of the design of adaptive filters. Although they were derived on the basis of certain assumptions (such as the range of eigenvalues being small and consistent), experience has shown that they can be used as a guide in most cases, even when the eigenvalues vary (WIDROW and STEARNS, 1985). From the equations we can see that, because μ is directly proportional to M but inversely proportional to τ_{MSE}, there is a tradeoff between the misadjustment M and the speed of adaptation τ_{MSE}.

For the adaptive filter used in the HPSE system, the minimum number of weights was calculated to be 17. However, increasing the number of weights results in an improved impulse response approximation. Empirically, it was found that increasing the number of weights beyond 40 resulted in very little observable improvement. In order to achieve a low misadjustment and good stability, a value of 0·0005 was selected for μ. While the system allows easy adjustment of these parameters, the above values have proved to be satisfactory in all cases reported.

3 Results

Before attempting to process HPS signals using the adaptive filter, a number of tests were conducted to verify correct operation. These included spectral analysis, a statistical stationarity check and the processing of signals with known shape, power and statistics using the adaptive signal enhancer.

3.1 *Performance check*

The performance of the adaptive signal enhancer has been investigated, using simulated waveforms. Fig. 3*a* shows the simulated periodic signal S_1, which provides the input to the filter. Uncorrelated white noise N_1, of equal power to the signal was also simulated and added to S_1. The result, $S_1 + N_1$, is shown in Fig. 3*b*. The 'desired' input signal, $S_0 + N_0$, was obtained by delaying $S_1 + N_1$ by a period equivalent to half the length of the filter. The output y of the 32-weight adaptive filter, which represents the least mean square estimate of S_0, is shown in Fig. 3*c*. It can be seen that, after the filter has converged, an optimum estimate of S_0 is achieved.

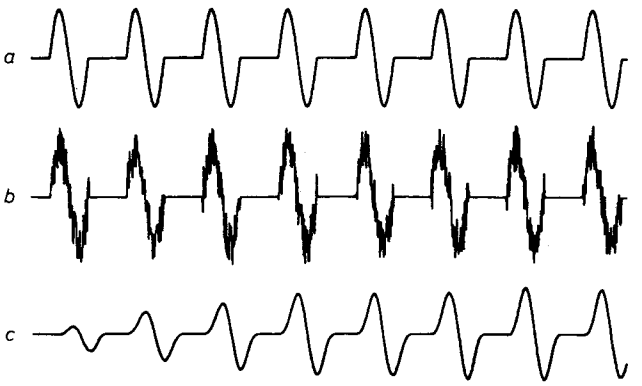

Fig. 3 *Example illustrating the performance of an adaptive filter during adaptation. (a) Simulated periodic signal; (b) same signal with white noise added; (c) output of adaptive filter. $n = 32$; $\mu = 0{\cdot}0005$*

3.2 *Spectral analysis*

Two experiments have been conducted to test the spectrum of His-Purkinje system electrograms (HPSEs).

(*a*) After averaging 128 beats of a normal subject's HPSE, the averaged waveform was subjected to filters with six different cutoff frequencies. From Fig. 4, it can be seen that the ramp feature of the HPSE is preserved down to $f_c = 51$ Hz. At $f_c = 25$ Hz, the ramp starts to become distorted. This indicates that most of the spectral power lies below 100 Hz.

(*b*) After adaptive signal enhancing of HPSE's, one beat was selected and its fast Fourier transform (FFT) computed. Fig. 5 shows that most of the power lies in the low part of the spectrum (below 100 Hz).

The above two experiments indicate that the desired

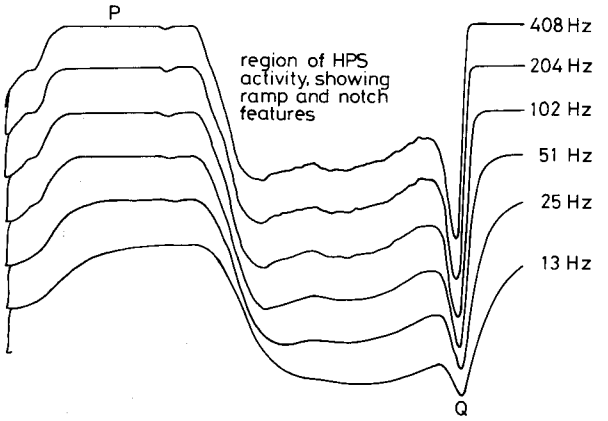

Fig. 4 *HPS recordings produced by averaging 128 consecutive beats, subjected to low-pass filtering at various different cutoff frequencies*

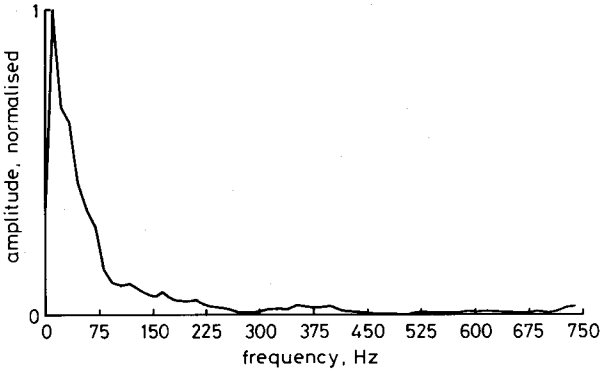

Fig. 5 *FFT of a 128-sample subwindow (from the end of the P-wave to the start of the Q-wave) taken from a single beat of the HPS signal*

power lies below the 100 Hz region. WOOLLONS *et al.* (1982) have reported that most power of the HPSE lies below 100 Hz. Berbari reported that 78 per cent of the total power lies below 50 Hz (BERBARI *et al.*, 1979).

3.3 *Stationarity check*

The adaptive filter converges to the Wiener solution under the condition that the input signal and the desired response are statistically stationary (WIDROW *et al.*, 1975). Once the adaptive filter has been optimised and has 'learnt' the statistics of the signals, it is able to track any slow variations in statistics (i.e. nonstationarity). Stochastic signals are described as 'wide-sense stationary' if only some of their statistical properties are unaffected by a shift in the time origin (PAPOULIS, 1984). These include such measures as the mean, variance and autocorrelation function (ACF).

To test the stationarity of the HPSE, a sequence of 10 beats was selected at random. The mean and variance of each beat were then computed and plotted against the beat numbers. The first beat of the selected portion was then shifted one beat at a time and the ACF was computed. The resultant ACF was then plotted against the shift number. The same test was repeated three times on different randomly selected sequences. Visual comparisons suggest that, in most cases, the HPS signals are in the wide-sense stationary.

3.4 *His-Purkinje system electrograms*

A total of 15 subjects have been studied to date, comprising 12 normal and 3 pathological cases. Satisfactory HPSEs have been obtained from 10 of the normal cases and from 2 of the pathological cases. It was noticed that subjects who did not produce satisfactory results tended to be grossly overweight. Spatial averaging might help to extract the desired signals in cases such as this, where the signal-to-noise ratio is low. However, for the majority of cases, successful results are obtained without spatial averaging. Results from three such cases are presented below, as examples.

Fig. 6 shows 132 ms subwindows of waveforms obtained from a normal 28-year-old male subject (case 1). The top trace is the ECG showing the end of the P-wave, P-R

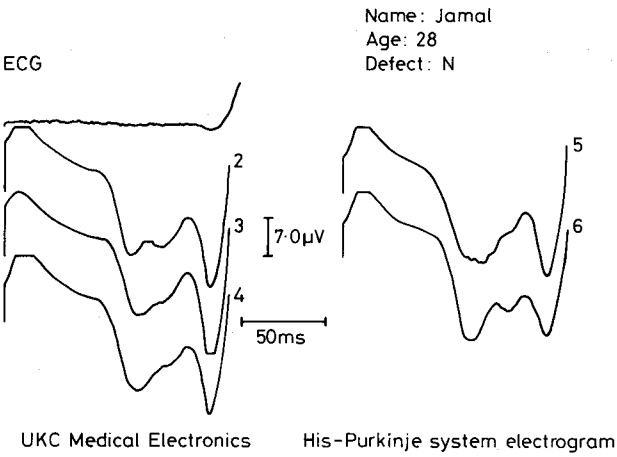

Fig. 6 *HPS recordings from a normal 28-year-old male (case 1), showing details from the end of the P-wave to the onset of the QRS complex*

segment and the beginning of the QRS complex. Below the ECG signal are five consecutive HPSEs which have been processed using the adaptive signal enhancer having 40 weights and a stability factor μ of 0·0005.

Fig. 7 shows the results of averaging 128 consecutive unprocessed beats, using the waveform processing facility

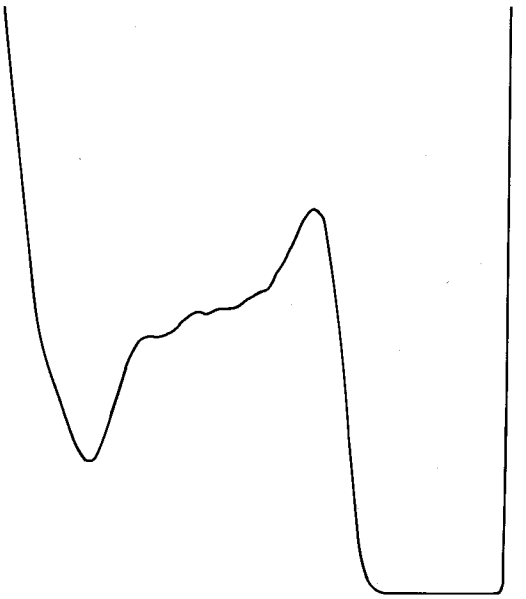

Fig. 7 *HPS recording produced by averaging 128 consecutive beats. Results are for case 1 and are presented for comparison with the adaptively enhanced beat shown in Fig. 6*

of a Gould digital storage oscilloscope, type 125. This trace is plotted on a different scale, but is included to demonstrate the similarity between averaged HPSEs and those produced using our system. Fig. 8 shows similar waveforms obtained from a normal 22-year-old male (case 2).

These results show that there is a ramp feature containing a high-frequency notch between the end of the P-wave and the onset of the QRS complex. Similar ramps have been reported by others (BERBARI *et al.*, 1979; BONES *et al.*, 1982*b*; PEPER *et al.*, 1984) and it is generally thought that they represent electrical activity from the His-Purkinje

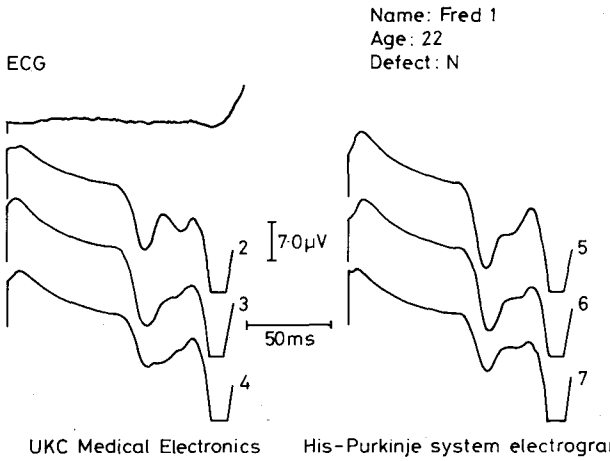

Fig. 8 *HPS recording from a normal 22-year-old male subject (case 2), illustrating variations due to the effects of breathing*

system. However, we intend to investigate this relationship further, by comparison with invasive measurements.

During our investigations we noticed that, while some patients exhibited a fairly consistent HPSE (as in case 1), others produced results which varied cyclicly. Case 2 is an example of the latter category. We believe that these cyclic variations are due to effects of breathing, which could affect the signal in two ways:

(i) As the subject breathes, the heart moves up and rotates slightly. Thus the heart's position with respect to the electrodes changes, which could alter the vector signal arriving at the electrodes.
(ii) Activity of the chest wall muscles during breathing could have the effect of modulating the signal.

To verify this theory, we repeated our measurements on case 2 with the subject holding his breath. These results are presented in Fig. 9, which shows that the cyclic variations are greatly reduced. The problem of the effects of breathing on the HPSE is still being investigated.

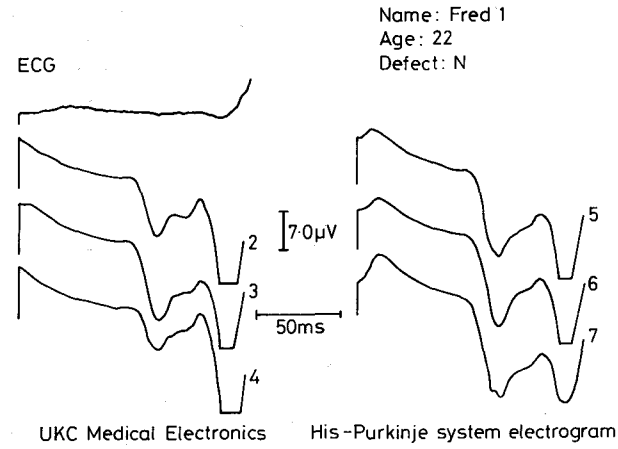

Fig. 9 *HPS recordings from case 2 taken while the subject was holding his breath*

Figs. 10 and 11 show junctional ectopics from a 53-year-old male patient with second degree A-V block (case 3). There are a few noteworthy points about this case:

(i) The ramp feature is absent during the junctional ectopic beats, but the notches remain.
(ii) The ramp and notch features are both present during every normal conduction (indicated by the presence of a P-wave).
(iii) As the P-R segment timing changes from beat-to-beat, the ramp notch timing also changes, with reference to the Q-wave.

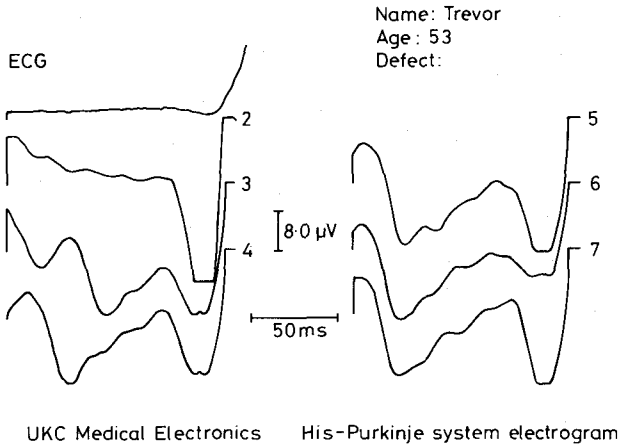

Fig. 10 *HPS recordings from a 53-year-old male patient with type I A-V block and junctional ectopics (case 3)*

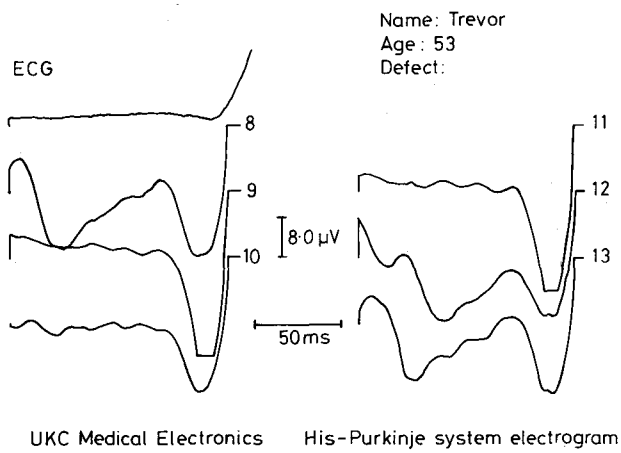

Fig. 11 *Further consecutive HPS signals from case 3*

(iv) Similar features are noticed every time the junctional beats stop.

While it is possible to display some of these features using specialised techniques, such as selective averaging of beats which have been classified by arrhythmia detection (BERBARI *et al.*, 1986), such techniques presuppose the result as they require an *a priori* knowledge of how particular HPSE features are related to arrhythmias. For diagnostic purposes it is desirable to detect HPSE features without such knowledge. Furthermore, some features, such as the gradual prolongation of the ramp feature seen in Fig. 10, are very difficult to categorise until after the signal has been processed. This supports the need for an alternative technique for recording HPS electrograms.

4 Discussion

A multistage system for the detection of beat-to-beat changes in signals from the His-Purkinje system at the body surface has been described. In the introduction, it was concluded that, because there is more than one type of noise involved, there is no optimal single solution to the problem. However, some of these types of noise can be suppressed without the need for sophisticated signal-processing techniques. Careful skin preparation, thorough screening, good circuit construction and the choice of low-noise electronic components have proved to be a major step in the improvement of SNR. Even the atrial interaction with the desired signal, due to phase problems generated by high-pass filtering, can be avoided by the use of automatic DC subtraction as described earlier on.

However, electromyographic potentials remained as the major source of noise corrupting the desired signals. This

study has demonstrated that there is a promising alternative to signal averaging for the reduction of this noise. The use of adaptive signal enhancement of periodic signals against additive random uncorrelated noise has been described. The results shown in Figs. 6 and 7 (case 1) demonstrate the similarity between the adaptively enhanced signals and averaged waveforms. However, problems such as the modulating effects of breathing can lead to variations from beat-to-beat. As these signals are from a normal healthy young man who has never experienced cardiac problems, we would expect a similar pattern of events for every beat. Because breathing is to a certain extent uncorrelated with the wave of activation, signal averaging will minimise the problem but will also smooth any important beat-to-beat variations in the HPSE. The delay used in the adaptive filter is capable of decorrelating random muscular noises but, because of the low-frequency mechanism of breathing, this delay is not sufficient to ensure decorrelation of breathing effects.

In cases where the signal-to-noise ratio is very poor, due to a large amount of muscular noise or where the patient is overweight, spatial averaging could be beneficial as suggested by MEHRA *et al.* (1983). As the 'desired' and 'input' components of the signal need not be of the same shape, signals recorded from multiple pairs of electrodes located at different sites on the chest could be averaged. An improvement in signal-to-noise ratio should then result. The averaged lead could be used as the input to the adaptive filter, while the desired signal would again be obtained from the V6R-V6L position.

There are of course limitations to adaptive filtering. These include the filtering of nonstationary signals, such as when there is a momentary increase in muscular activation. In such cases signal enhancement will be suboptimal. Also, the filter weight and convergence factor need to be carefully selected because of the instability which could result from the wrong choice of stability factor. The offline processing of the signals would be a fundamental limitation to use in long-term monitoring. However, once this technique has been accepted and proved to be useful as a clinical tool, LSI processing could be used to perform the filtering in real time.

The results which we have obtained show good agreement with those reported by Berbari and others and also with the results of first-order models derived by Berbari and Bones; i.e. they exhibit the same ramp and notch features, thought to represent HPS activity.

The technique which has been developed should be equally capable of beat-to-beat detection of late potentials, as these signals have similar characteristics to the HPS potentials. Both signals have amplitudes of a few microvolts and the bandwidth of late potential signals (100–300 Hz) (BRIEHARDT *et al.*, 1981; BERBARI, 1983) lies within the system's capability (0–500 Hz). The difference in timing necessitates repositioning of the time domain window, but this can be achieved with minor software modification. We are currently investigating this possibility.

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Authors' biographies

Hasan Al-Nashash (26) was born in Jordan and obtained his B.Sc degree in Electronic Engineering with Medical Electronics in 1984. He then undertook a short project in conjunction with the Medical Physics Department at the Kent and Canterbury Hospital prior to registering for Ph.D. research at the University of Kent. He is currently engaged in research in noninvasive electrocardiography.

Stephen Kelly (36) M.Sc., C.Eng., MIEE, MIERE is a lecturer in Electronic Engineering at the University of Kent. Prior to this appointment (in 1983) he worked in industry for 11 years, achieving chartered status in 1977 and obtaining his M.Sc. degree, in Computers and Cybernetics, in 1981. He is currently involved in medical electronics research, primarily in the fields of cardiology and audiology.

Dr D. J. E. Taylor MB, BS, MRCS, FRCP is Consultant Cardiologist at the Kent and Canterbury Hospital. He graduated from the Westminster Medical School and London University in 1958, obtained his MRCP in 1962 and FRCP in 1975. As Consultant Cardiologist, he runs a noninvasive cardiology laboratory and gives consultant opinion on cardiology for a large area of Kent. He also has an active research programme in all aspects of noninvasive cardiology.