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Monitoring of heart and respiratory rates by photoplethysmography using a digital filtering technique

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ABSTRACT

An apparatus for simultaneously monitoring heart and respiratory rates was developed using photoplethysmography (PPG) and digital filters, and compared with conventional methods. The PPG signal, which includes both heart and respiratory components, was measured at the earlobe with an original transmission mode photoplethysmographic device. A digital filtering technique was used to distinguish heart and respiratory signals from the PPG signal. The cut-off frequency of the respiratory signal filter was selected automatically depending on the heart rate. Using digital filtering techniques, heart and respiratory signals were separated at rest and during exercise. The digital signal processor was employed to realize an adaptive and real-time filtering. The heart rate was calculated by the zero-crossing method and the respiratory rate from the peak interval of the filtered signal. To evaluate the newly developed monitor, an ECG for heart rate and a transthoracic impedance plethysmogram for respiratory rate were monitored simultaneously. To obtain higher heart and respiratory rates, exercise was performed on an electrical bicycle ergometer. Heart and respiratory rates calculated by the new method compare to those obtained from ECG and the transthoracic impedance plethysmogram. The maximum error of heart and respiratory rates was 10 beats/min and 7 breaths/min, respectively. Copyright © 1996 Elsevier Science Ltd for IPEMB.

Keywords: Heart rate, respiratory rate, photoplethysmography, digital filter, digital signal processor, exercise

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INTRODUCTION

Photoplethysmography (PPG), which was developed by Hertzman[1], is a simple and useful method for measuring the pulsatile component of the heartbeat and evaluating the peripheral circulation. The PPG waveform contains two components[2]; one, attributable to the pulsatile component in the vessels, i.e. the arterial pulse, is caused by the heartbeat, and gives a rapidly alternating signal (AC component), and the other is due to the blood volume and its change in the skin and gives a steady signal that changes only slowly (DC component). Motion gives a large artefact in the DC component. In general, only the AC component can be distinguished using band-pass filtering. Although quantitative analysis in terms of blood flow is difficult, qualitative evaluations can be made, for example comparing the record from peripheral blood vessels before and after surgical vascular repair[3].

The PPG signal consists of not only the heartbeat but also a respiratory signal[4]. Using suitable filters, the two signals can be separated and heart and respiratory rates recorded simultaneously. Recently, methods for simultaneous monitoring of heart and respiratory rates at rest using optical fibres have been developed[5,6]. Continuous respiratory rate monitoring has been used for investigating the sleep apnoea syndrome, in sports medicine, and evaluation of stress using pulse oximetry[7], in thermistor devices, impedance plethysmography[8], inductance plethysmography, and capnography. However, these monitors have proved difficult to use, or of low reliability, and the technique of PPG may be an improvement.

When analogue filters with suitable cut-off frequencies are used, heart and respiratory signals can be distinguished in the PPG signal. However, this technique is difficult to use during exercise. At rest the respiratory rate is normally 10–20

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breaths/min. During exercise it can increase to approximately 45 breaths/min[9]. Because the heart rate at rest is 38–110 beats/min, this range includes the value for respiratory rate during exercise.

Using an analogue filter with a fixed cut-off frequency, low respiratory rate signals can be distinguished from the heart rate signal, but not high respiratory rate signals, and it is therefore difficult to obtain accurate respiratory rate measurements during exercise. However, using a digital signal processing technique, two methods can be used to distinguish heart and respiratory rates, frequency analysis and digital filtering. In the former method, peaks corresponding to the heart and respiratory rates were shown as a spectrum[5]. In clinical use, however, it was difficult to distinguish these harmonics and to use it as an on-line monitoring system. Continuous digital filtering of the PPG can be used to measure heart and respiratory rates simultaneously[6,10,11]. We have developed an adaptive digital-filtering technique between heart and respiratory waveforms in the PPG signal and for simultaneously monitoring heart and respiratory rates at rest and during exercise.

SUBJECTS, MATERIALS AND METHODS

Subjects

This studies were conducted on 11 healthy males (mean age 26 years, SD 4 years, range 22–34). The PPG probe was attached to the earlobe (*Figure 1*). Room temperature was maintained at 24 ± 2°C.

Apparatus

Figure 1 shows the experimental set-up. The system comprised analogue and digital sections. The analogue part consisted of an original transmission mode photoplethysmographic probe, a current-to-voltage converter, amplifiers, and analogue filters. A LED (wavelength 880 nm, CN305, Stanley Electric Co., Japan) was used as a light source and the transmitted light was detected by a photodiode (S1087–01, Hamamatsu Photonics K.K., Japan). A current-to-voltage converter and an amplifier were made by an operational amplifier (TL072CP, Texas Instruments). In this study, it was assumed that the maximum heart rate was 200 beats/min and minimum respiratory rate was 6 breaths/min. To attenuate higher frequencies than the maximum heart rate and lower ones than the minimum respiratory rate, analogue filters were used. According to assumption the cut-off frequencies of the low-pass and high-pass filters were selected as 5 Hz and 0.1 Hz, respectively. The analogue filters were chosen for the following three reasons: (1) the filter of Butterworth type is low in the pass-band ripple; (2) higher frequency noises could be filtered using fourth-order in low-pass filter; and (3) lower frequency noises could not be filtered using fourth-order, but eighth-order were filtered enough in high-pass filter. The analogue low-pass filter (SR-4BL1, NF Electronic Instruments, Japan) and high-pass filter (SR-4BH1 × 2, NF) used in this study were unity gain active filters (fourth- and eighth-order Butterworth type, respectively). The digital portion of the apparatus consisted of a personal computer

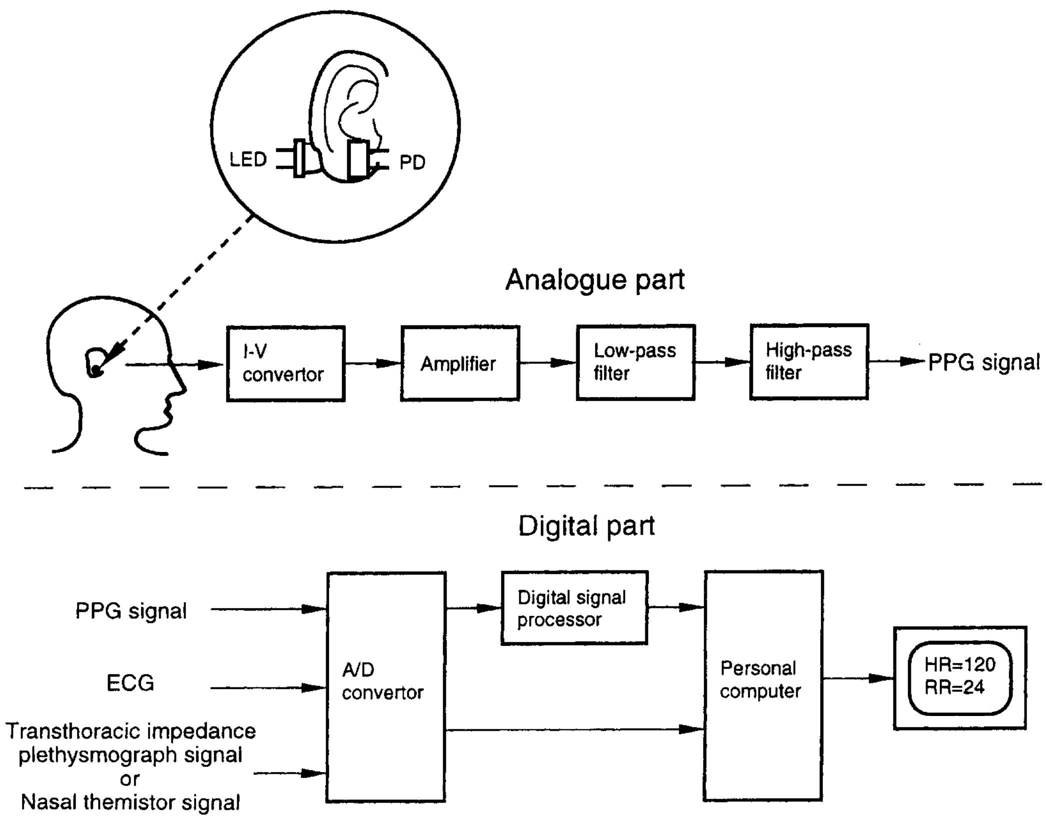

Figure 1 Block diagram of PPG using digital filters

(16 MHz 386SX, PC-9801NS/E, NEC, Japan), a 12-bit A/D converter board (Analog-pro1, Canopus Co., Japan), and a 16-bit fixed point digital signal processor (DSP) board (ADSP-2101–50, 12.5 MIPS, Analog devices, DSP98–2101BPC, Micro Science Co., Japan). The PPG signal was sampled at 100/s.

Digital filters

Digital filters are divided into two types; infinite impulse response (IIR) and finite impulse response (FIR) filters[12]. Input and output signals to both types of filter are related by the convolution sum. The current output sample, $y(n)$, is given by equation (1) for the IIR and by equation (2) for the FIR filter:

$$y(n) = \left\{ \sum_{k=0}^{N} a_k x(n-k) \right\} - \left\{ \sum_{k=1}^{M} b_k y(n-k) \right\} \tag{1}$$

$$y(n) = \sum_{k=0}^{N-1} a_k x(n-k) \tag{2}$$

where a_k and b_k are the coefficients of the filters, and N and M are the values of a_k and b_k, respectively. In equation (1), $y(n)$ is a function of past outputs as well as present and past input samples; the IIR is therefore a feedback system. In the FIR equation $y(n)$ is a function only of past and present input values. FIR requires larger N for sharp cut-off filters than IIR. Thus for a given amplitude response specification, more processing time and storage are required for FIR implementation. However, we chose the FIR for the following reasons: (1) FIR filters can be constructed to have a strictly linear phase response, so that no phase distortion is introduced into the signal by the filter. The phase responses of IIR filters are nonlinear; (2) FIR filters realized nonrecursively, that is by direct evaluation of equation (2), are always stable. The stability of IIR filters cannot be guaranteed; (3) the effects of using a limited number of bits to implement filters such as round-off noise and coefficient quantization errors are much less severe in FIR than in IIR; and (4) DSP can calculate rapidly enough for real-time monitoring of heart and respiratory rates (for example, when N = 1024, the calculation time using this DSP was 165 μs).

To obtain suitable coefficients of the FIR, the window method[12] was achieved. We could obtain coefficients by evaluating the inverse Fourier transform of an ideal frequency characteristics. Then, white-noise was filtered by FIR filter using determined coefficients, and the filtered signal was evaluated again by Fourier transform for checking the characteristics of FIR filter. We obtained suitable specifications of the filters, as shown in *Table 1*. *Figure 2* shows the characteristics of the filters for separating heart- and respiration-related waves. The heart-related signal was band-pass filtered, and a low-pass filter was used for respiration. To obtain enough attenuation level (over 100 dB), each filter has 1024 coefficients. These coefficients, which were stored as a file, were entered into the DSP board from the PC's hard disk by a control program before starting a recording program. The heart-related wave was distinguished using filter no. 1, while the respiration-related wave was distinguished by filters nos. 2–4. The appropriate cut-off frequencies of the filters were chosen empirically.

Depending on the heart rate, three low-pass filters were selected automatically. To obtain stable and smooth filtering, hysteresis inclusion was required. Reference heart rates for the filter-election logic are shown in *Figure 3*. These reference heart rates were determined empirically.

Evaluation of heart and respiratory rates

The PPG signal was simultaneously recorded with a telemeter ECG monitor (Dyna scope DS-3100, Fukuda Denshi Co., Japan) for the heart rate and a transthoracic impedance plethysmograph (TR-601T, Nihon Kohden Co., Japan) for the respiratory rate. Although transthoracic impedance plethysmography has some problems such as moving artefact during measurement, difficult-to-evaluate qualitative values and attachment of electrodes, we chose it as a reference method because this technique was an unconstrained and relatively suitable monitor during exercise using the bicycle ergometer if the electrodes were securely attached. The heart rate was calculated from the filtered heart waveform of the PPG using the zero-crossing method. The respiratory rate was calculated from the respiratory waveform of the PPG using the peak detection method. The peak interval of the transthoracic impedance plethysmogram was counted to obtain a reference respiratory rate. In our monitor, heart and respiratory rates were calculated as the median value of ten and five sequential counts, respectively. The respiratory rate obtained from the transthoracic impedance plethysmogram was also calculated as the median value of five sequential counts. The heart and respiratory rates displayed on the PC obtained from PPG and references were stored every 1 s (2 s, except for two subjects) on the RAM disk of the PC.

Experimental procedure

To obtain higher heart and respiratory rates, exercise was performed on an electrical bicycle ergometer (RE800 Ergo system, Rodby Elektronick AB, Sweden). The subjects were studied in a sitting position on the bicycle ergometer and were allowed to rest for at least 10 min before experimental recording. The total recording time was 20 min. During the first 3 min, subjects were allowed to rest in a sitting position, and exercise was then performed for 10 min. The ramp load was applied increasing by 10 W per min from 30 to 130 W. The pedalling rate was maintained at 55 rpm. After exercise, the subject rested, sitting on the bicycle for the remaining 7 min.

Table 1 Specifications of the obtained filters

Filter no.	Type	Pass-band (Hz)	Transition width (Hz)	Pass-band ripple (dB)	Stop-band attenuation (dB)
1	band-pass	1.50–2.30	1.00	0.35	138
2	low-pass	0–0.30	0.35	0.40	80
3	low-pass	0–0.40	0.60	0.18	90
4	low-pass	0–0.55	1.45	0.07	118

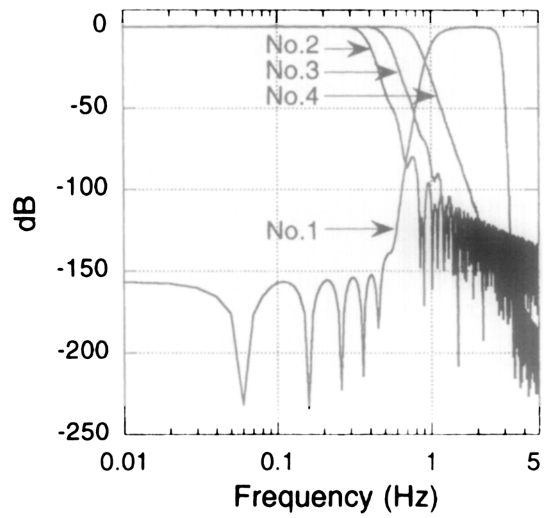

Figure 2 Frequency response of heart rate filter (no. 1) and respiratory rate filters (nos. 2–4)

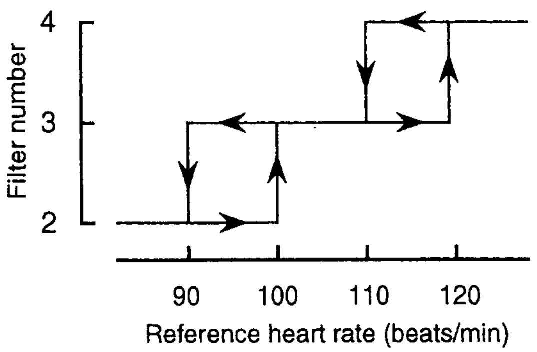

Figure 3 Reference heart rate for selecting the respiratory filters

RESULTS

PPG recordings

Figure 4 shows a typical example of analysed heart signals in a subject aged 32 years: (*a*) shows the data at rest and (*b*) the data at high heart rate, when performing the ergometer exercise. The upper recording shows the data obtained from the filtered PPG signal and the lower one shows the ECG. The intervals of the PPG peaks agreed well with those obtained from the R waves of the ECG. There was a time lag between the PPG peaks and the R waves of the ECG. This consisted of the delay of the pulse wave transmitted to the earlobe and the time delay of the analogue and digital filters.

Figure 5 shows a typical example of analysed respiratory signals of the same subject: (*a*) shows the data at rest and (*b*) the data at high respiratory rate during exercise. The upper recording shows the data obtained by the filtered PPG signal and the lower one shows the transthoracic impedance plethysmograph. The PPG signal was filtered by filter no. 2 in *Figure 5a* and by filter no. 4 in *Figure 5b*. The intervals of the PPG peaks agreed well with those obtained by plethysmography at rest and during exercise. During exercise, the respiratory signal of the PPG began to decrease and included some artefact.

Comparison of heart rate

The heart rate derived from our system was compared with that obtained by the ECG monitor. *Figure 6* shows a typical example of dynamic variation in heart rate at rest, and during and after exercise. Because of the median value procedure, heart rates calculated from the PPG signal were obtained approximately 10 s after the start of the recording. Heart rates obtained from PPG (solid line) agreed well with those obtained by ECG (dashed line) at rest and during exercise in this subject. The maximum error of heart rate was 10 beats/min.

Figure 7 shows the relationship between heart rates obtained with PPG and ECG, and includes all subjects recorded at rest, during exercise, and after exercise, omitting the initial 30 s of the recording.

Comparison of respiratory rate

Figure 8 shows a typical example of dynamic variation in respiratory rate at rest and during exercise. Because of the median value processing, the respiratory rate was evaluated approximately 60 s after the start of the recording in this subject. Because of the hysteresis in the filter selection procedure the heart rate for changing from filter no. 2 to 3 was 100 beats/min, while that for changing from no. 3 to 2 was 90 beats/min. When changing filters from no. 3 to 4 and no. 4 to 3, heart rates were 120 and 110 beats/min, respectively (see *Figures 6* and *8*).

Each plot in *Figure 9* shows the relationship between respiratory rate obtained by PPG and that obtained with the transthoracic impedance plethysmograph in all subjects recorded at rest, during exercise, and after exercise, omitting the initial 90 s from the recordings. The standard deviation increased gradually with the increase in respiratory rate. The maximum error of respiratory rate was 7 breaths/min.

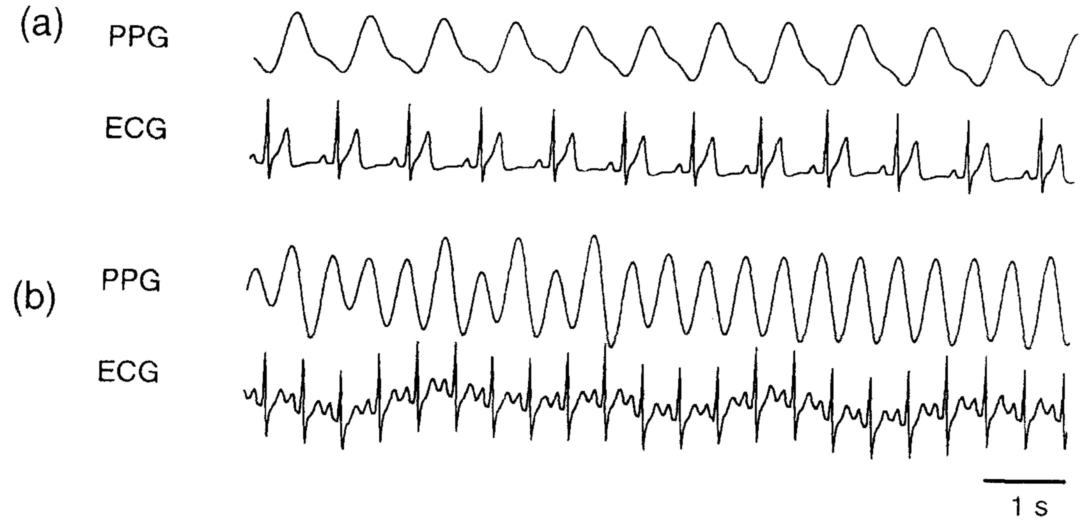

Figure 4 Heart rate signals measured by PPG (upper line) and ECG (lower line): (a) at rest; (b) at high heart rate

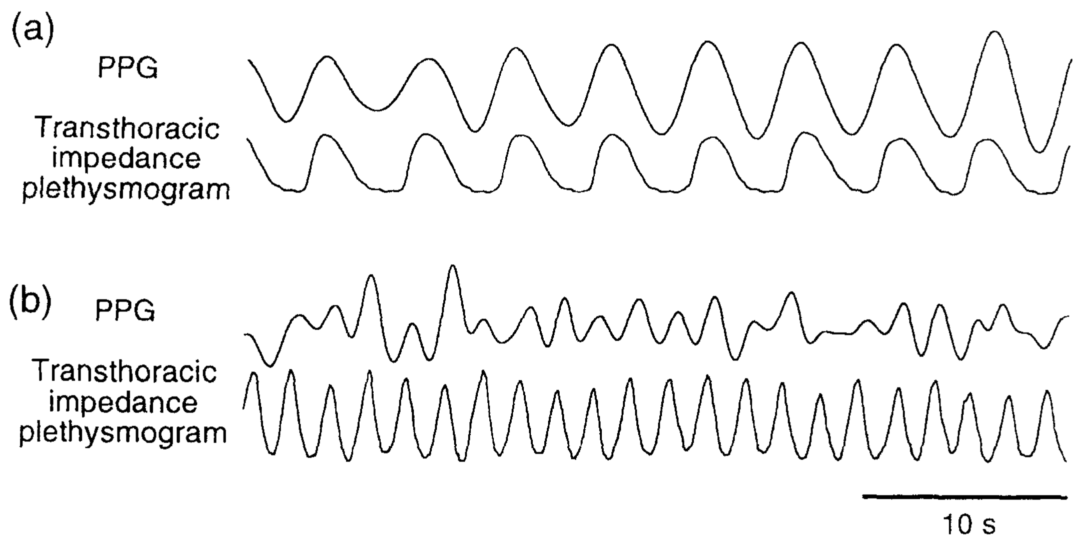

Figure 5 Respiratory signals measured by PPG (upper line) and transthoracic impedance plethysmography (lower line): (a) at rest; (b) at high respiratory rate

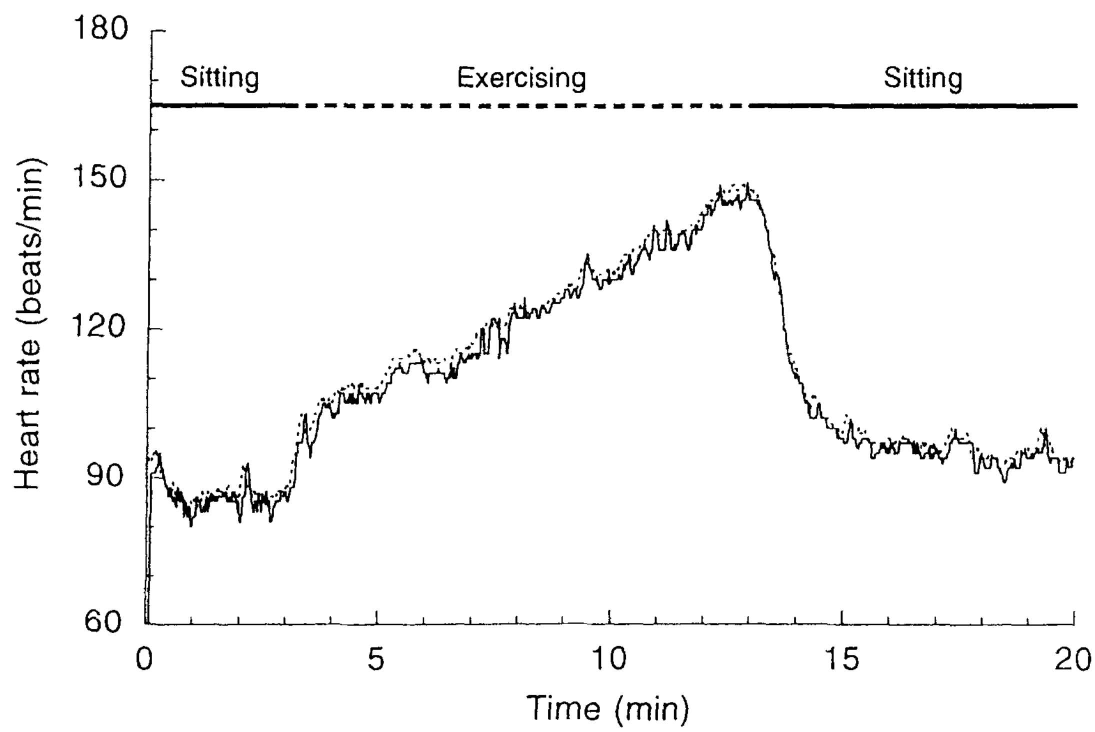

Figure 6 Time course of heart rate obtained by PPG waveform (solid line) and ECG (dashed line)

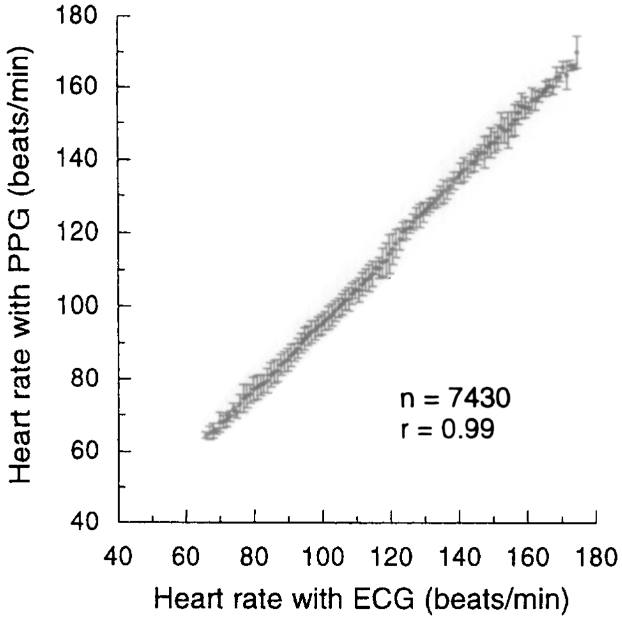

Figure 7 Relationship between heart rate obtained from PPG and ECG. Points and bars indicate mean and standard deviation of heart rate by PPG

The relationship between heart and respiratory rates

Figure 10 shows a typical example of the relationship between heart and respiratory rates at rest, during exercise, and after exercise, omitting the initial 30 s of the recording in the same record as shown in *Figures 6* and *8*. The relationship between heart and respiratory rates showed a hysteresis of counterclockwise rotation in five subjects (include *Figure 10*), clockwise rotation in four subjects, and showed no hysteresis in two subjects. The size and the rotary direction of hysteresis depended on the subject.

DISCUSSION

The apparatus for monitoring simultaneous heart and respiratory rates was developed using PPG and digital filtering techniques with DSP, and compared with conventional methods. Heart rates showed good agreement between our monitor and the ECG monitor, while the respiratory rate also showed agreement between this new method and transthoracic impedance plethysmography. However, there were certain problems encountered in measuring respiratory rate with the new monitor.

- The analogue filters were used to attenuate a higher frequency than the heart rate during exercise and a lower one than the respiratory rate at rest. The sampling theory of A/D conversion and stop-band of digital filters had a limited attenuation level. High- and low-frequency noises were not attenuated using digital filters only.
- The amplitude of the PPG signal varied between subjects. In this study, the amplifier gain was adjusted for individual subjects. The heart signal of the PPG increased with the increase in blood flow caused by exercise. Therefore, in clinical use the monitor system might require an auto-gain control circuit.
- During exercise, the amplitude of the synchronous respiratory PPG signal was smaller than the signal at rest. The motion artefact was large in the PPG signal during exercise. Therefore, it was difficult to obtain the synchronous respiratory PPG signal clearly at a high respiratory rate (see *Figure 5*).

For these reasons, the error increased with the respiratory rate. In order to distinguish the respir-

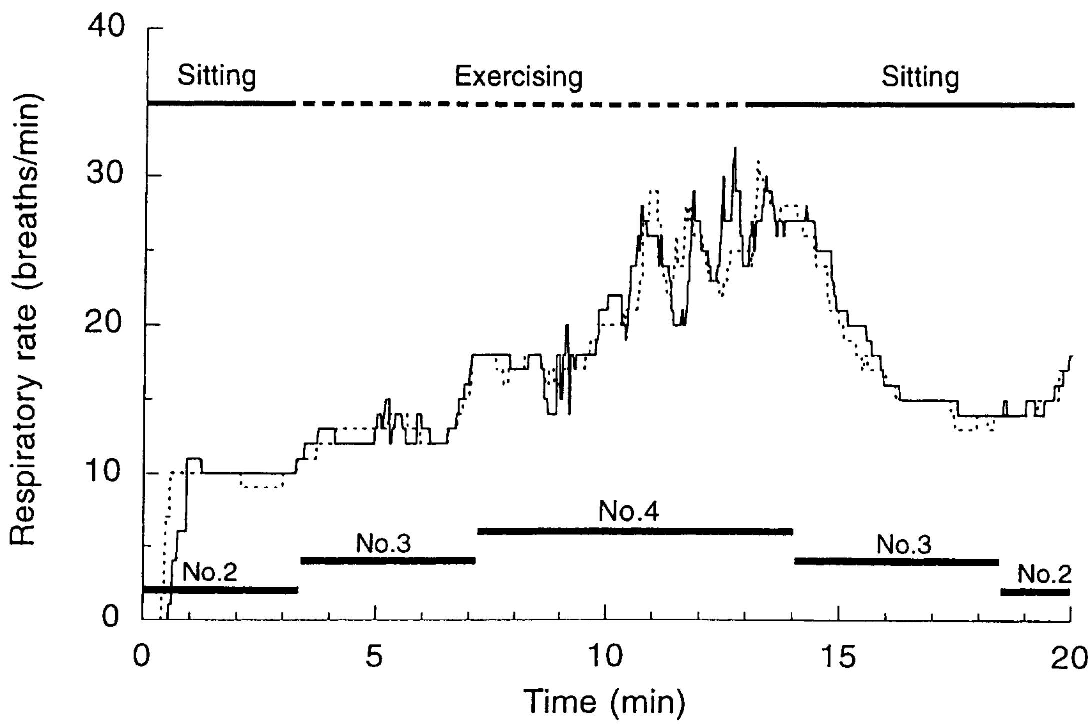

Figure 8 Time course of respiratory rate obtained by PPG waveform (solid line) and transthoracic impedance plethysmography (dashed line)

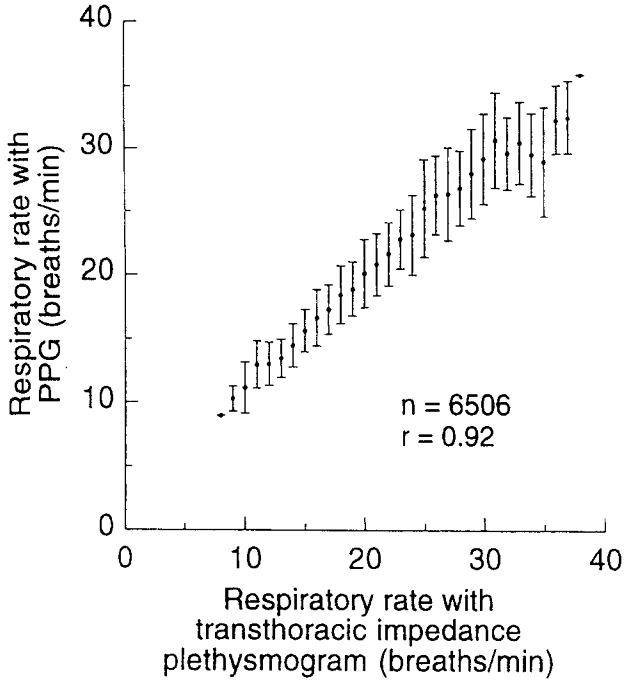

Figure 9 Relationship between respiratory rate obtained from PPG and transthoracic impedance plethysmography. The points and bars indicate the mean and standard deviation of respiratory rate obtained by PPG

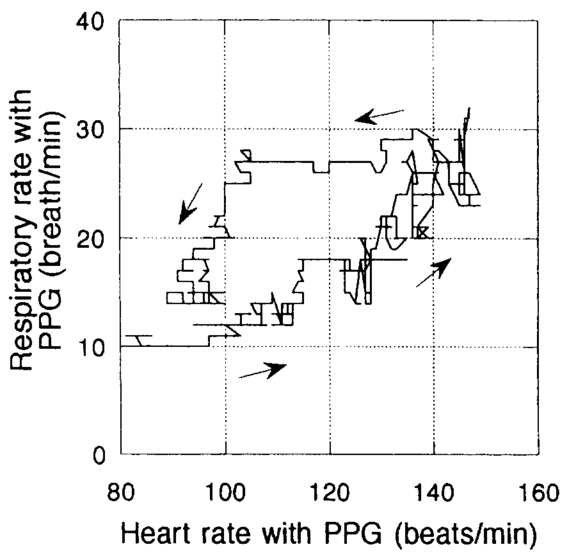

Figure 10 Relationship between heart and respiratory rates at rest, during exercise, and after exercise in the same record as shown in *Figures 6* and *8*

atory wave more precisely, a narrow band-pass filter is required. In addition the digital filter requires a large N in equation (2) and accurate calculation. In this study, the PPG signal was filtered by the 16-bit fixed point DSP. This DSP cannot have a greater coefficient set of the FIR digital filters because of the limitation of RAM area. If a 32-bit floating point DSP were used in digital filtering, the error in heart and respiratory rates would be decreased, even during exercise.

Because of the hysteresis in the automatic filter selection procedure, filter selection was stable and smooth filtering was achieved. Heart rate was different when changing from filter no. 2 to 3 and 3 to 2, and from filter no. 3 to 4 and 4 to 3 (see *Figures 6* and *8*).

It is known that work on the bicycle ergometer with a pedalling frequency of, for example, 50 rpm usually gives a respiratory frequency related to the pedalling frequency. In other words, because the respiratory rate is related to $1/n$ pedalling frequency ($n = 1, 2, 3, \ldots$), respiratory rate often increases stepwise from 12.5 to 16.6, 25.0, up to 50 breaths/min at maximal work[9]. In this study, as pedalling frequency was 55 rpm, respiratory rate increased stepwise from 13 to 18, and 27 breaths/min during exercise at time 4–6, 7–10, and 11–14 min, respectively (see *Figure 8*). Both respiratory and heart rates decreased continuously after exercise.

Both the maximum error and standard deviation of respiratory rate increased with increased exercise, because the signal amplitude was too small and the motion artefact was too large to evaluate the respiratory rate accurately.

The probe of the monitor was similar to that of a pulse oximeter. If the photoplethysmographic probe included two wavelength of LED signals, it could measure arterial oxygen saturation at rest and during exercise. Simultaneous monitoring of respiration rate and oxygen saturation could be used to measure respiratory function clinically. In this study, the synchronous respiratory waveform was filtered from the PPG signal using DSP. Using digital filtering techniques, the synchronous respiratory waveform can be distinguished from a signal which includes synchronous respiratory signals such as the ECG and the blood pressure waveform.

The relationship between heart and respiratory rates showed hysteresis. If this hysteresis indicates a physiological parameter, the development of this new system could be useful for monitoring heart and respiratory rates.

In conclusion, using digital filtering techniques with DSP, the heart and respiratory rates were measured at rest and during exercise with one photoplethysmographic probe. The development of this new system has uncovered some potential problems, but it has demonstrated that it is to possible to monitor simultaneously and on-line heart and respiratory rates.

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