CARDIAC BEAT DETECTOR
A Novel Analogue Circuitry for the First Heart Sound Discrimination

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Abstract: Cardiac beat detector, which is an analogue circuitry installed in a novel non-invasive system for measuring heart rate in mice by using a piezoelectric transducer (PZT) sensor, performs an critical role in detecting the first heart sound (S1) in heart sounds. The PZT sensor detects heartbeat vibration and converts it to an electrical signal, namely the heart sounds. The measurement in intervals of S1s in the heart sounds is required to calculate heart rate, however, it is not simple because a S1 is a vibrating signal and has multiple peaks, which fluctuate in interval and in magnitude. In addition, respiration sound noise, which has frequency components similar with that of S1, makes S1 detection difficult and complex. The cardiac beat detector made it possible to overcome these problems by transforming multi-peaked S1 signal into a quasi-digital pulse. This technique is also available for the use in humans. Thus, the cardiac beat detector would contribute to the progress in the non-invasive heart rate measurement when it is installed in various, novel phonocardiogram-based equipments for the use in the fields of clinical and basic science in medicine.

1 INTRODUCTION

In experiments using small animals such as mice, a clip ECG electrode is often used for ECG recording (Yamada et al., 2001). However, investigators often encounter the problem with ECG signal deterioration or instability during long recording due to the hairy limbs and drying up of electrolytic paste between the limbs and the clip electrodes particularly in small animals. Moreover, there is an undeniable possibility that the pain induced by the electrode attachment might activate the sensory neurons and influence on the physiological state even in an anesthetized animal (Sato, 2007).

To overcome these problems, we recently developed a non-invasive cardiorespiratory monitor system for small animals using a piezoelectric transducer (PZT) sensor, which converts cardiac beats into an electrical signal when a small animal was simply placed on it (Sato et al., 2006; US patent 7174854). The PZT cardiorespiratory monitor enables stable and long measurement of heart rate of sleeping or anesthetized animals. Only placing an animal on the PZT sensor is required to monitor heart rate, and therefore, it gives no pain to animals. To calculate the heart rate, it is required to detect the first heart sound (S1) in a heartbeat signal detected by the PZT sensor. However, it is not simple to detect S1 constantly by distinguishing from noises of a frequency range similar to that of S1 because a S1 is composed of multi-peaked vibrating signal (Rangayyan and Lehner, 1988) and its magnitude decreases in anesthetized animals and humans (Manecke et al., 1999). A cardiac beat detector, which is made of a custom-designed analogue circuitry for S1 detection, was strikingly effective for detecting S1 and the second heart sound (S2) and for computing heart rate with a simple microprocessor algorithm.

2 METHOD

2.1 PZT Cardiorespiratory Monitor

The PZT cardiorespiratory monitor system consists of a PZT sensor device and a main unit, which contains two band-pass filters, a cardiac beat detector, a breathing movement detector, microprocessors and a temperature controller for the PZT sensor device. The sensor device consists of a disk-shaped thin PZT placed in a hole cut in a copper plate and covered by 0.5 mm-thick insulating sheets, which all were mounted on an electronic controlled heater (Sato et al., 2006) (Fig. 1).

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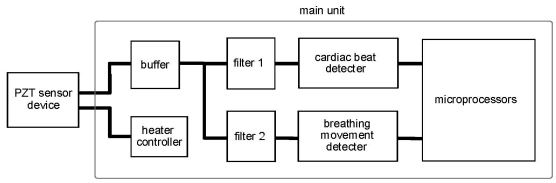

Figure 1: Block diagram of PZT cardiorespiratory monitor.

2.2 Signal Separation by Filters

Heart sound and breathing movement signals were separated from the PZT output signal by filters with pass band of 280-1000 Hz and 0.4-2.6 Hz for heart sounds and breathing movements, respectively, as shown in Fig. 2.

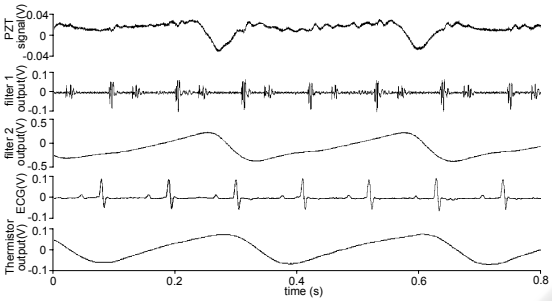

Figure 2: PZT output signal, Heart sounds, breathing movement signal, ECG and thermistor airflow sensor.

However, it was found that the presence of respiration sound noise, which was produced by airflow in airway, disturbs the detection of S1 (S2) when the magnitude of S1 (S2) declined in anesthetized mice (Fig. 3; upper trace). In addition, airway secretion produced marked, large-amplitude respiratory noise (Fig. 3; lower trace). These respiration-related noises were hardly possible to be removed by a filter because the frequency component of them were similar to that of S1; the period of vibrating signal of S1 (Ts) and respiration sound noise (Tr) was ranged between 1.4 and 4.0 ms (average = 2.4 ms, n = 50) and between 3.5 and 7.7 ms (average = 5.4 ms, n = 50), respectively. The frequency components of both S1 and respiration sound noise fluctuated, and therefore, they ranged widely and overlapped each other.

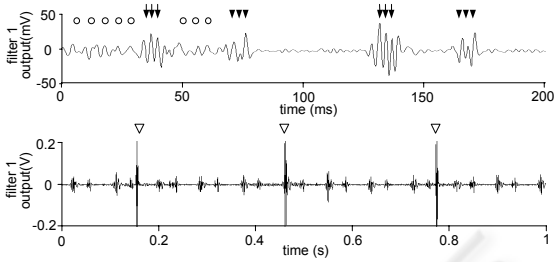

Figure 3: Representative traces of respiration sound noise (open circles; upper trace) and large-amplitude respiratory noise (open triangles; lower trace). Arrows and arrowheads indicate S1 and S2, respectively.

2.3 Cardiac Beat Detector

Since it was difficult to accomplish effective removal of the respiration sound noise from heart sounds by a filter because they have similar frequency components and fluctuate, we have developed a novel cardiac beat detector circuit, which consists of two diode detectors connected to a differential amplifier, an AC amplifier, and a hysteresis comparator (Fig. 4a). This circuit has three functions; (1) S1 emphasizing, (2) transforming S1 into a quasi-digital pulse and (3) automatic threshold controlling (Sato et al., 2006).

2.3.1 S1 Emphasizing Function

As described above, frequency components of S1 are at slightly higher range than those of respiration sound noise, although the both components fluctuate and overlap in part. To overcome the fluctuation, the cardiac beat detector was designed to emphasize always a higher frequency sounds over relatively lower frequency sounds. The S1 emphasizing function is produced by the combination of two diode detectors, which work as envelop detectors, and a differential amplifier (Fig.

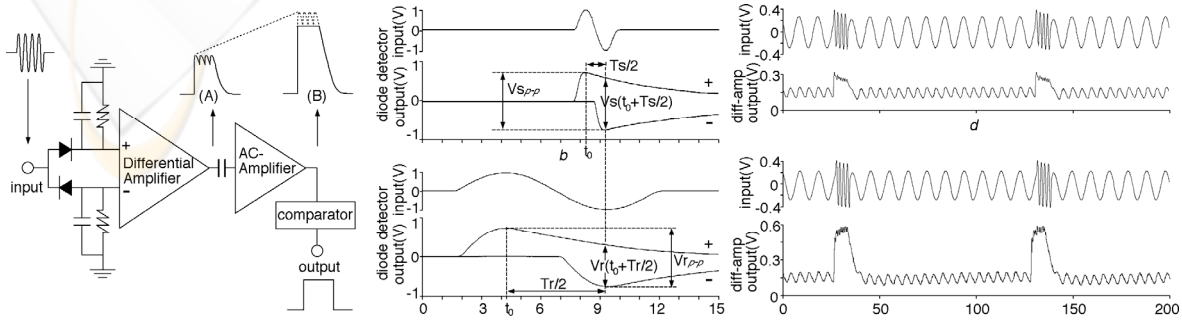

Figure 4: Function of the cardiac beat detector.

4a).

The two-diode detectors produce positive and negative envelopes with ripples when a sine wave is input. The voltage of the positive envelope during the declining phase (V(t)) is determined by a time constant RC as

$$V(t) = V_p\, e^{-t/RC} \quad (1)$$

where R and C are the resistance and the capacitance of the diode detectors and t is the time after a time of positive peak in the input and V_p is the voltage of the peak.

Output voltage difference between the two diode detectors at t = T/2 (V(T/2)) is calculated as

$$V(T/2) = V_p\,(1 + e^{-T/2RC}) \quad (2)$$

where T is the period of the input signal. Therefore, the higher input signal frequency, the larger voltage difference the diode detectors output. In fact, output voltage difference for an input sine wave of higher frequency ($Vs(t_0+Ts/2)$; Fig. 4b) is larger than that of lower frequency ($Vr(t_0+Tr/2)$; Fig. 4c).

This voltage difference appears equally in the differential amplifier output. Responses of the diode detectors to an input of a synthesized waveform, which consisted of alternating 4 cycles of a 500Hz sine wave (artificial S1) and 10 cycles of a 100Hz sine wave (artificial respiration sound), are shown in Fig. 4d and e. The artificial S1 is enhanced as compared to the artificial respiration sound when amplitudes of both inputs are almost the same (Fig. 4d), while the artificial S1 is largely enhanced when it is slightly larger than the artificial respiration sound in input signal (Fig. 4e). Fig. 4d demonstrates that the diode detectors have the S1 emphasizing function, while Fig. 4e shows an additional contribution of a rectifying property of diode, which abruptly reduces its resistance to the signal that exceeds about 0.3V, to the S1 emphasizing function.

Fig. 5 shows an example of quasi-digital pulses output from the differential amplifier when a real filtered heart sound signal was input ((A); Fig. 4a). The amplitude ratio of the S1 signal (filled circle) to the respiration sound noise (open circles) was enhanced from 3-fold in the input (broken lines; upper trace) to 10-fold in the output (lower trace) (Fig. 5). In addition, the cardiac beat detector combines the multi peaks of S1 into a quasi-digital pulse, which is helpful for the comparator to detect S1 easily. The unique combination of these effects enabled the emphasizing of S1 of higher frequency over the respiration sound noise of lower frequency, thus enabling a great improvement in the S/N ratio of the quasi-digital pulse.

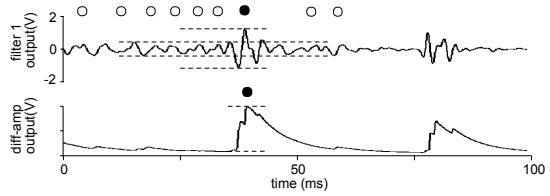

Figure 5: S1 emphasizing function of cardiac beat detector.

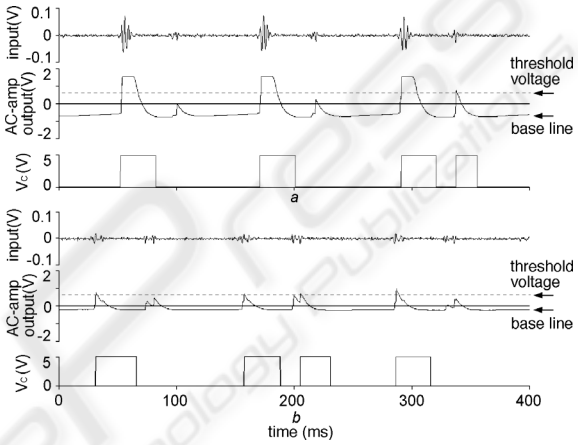

Figure 6: Automatic threshold adjustment controlled by the cardiac beat detector.

2.3.2 Automatic Threshold Controlling Function

The quasi-digital pulse (Fig. 5; lower trace) output from the differential amplifier was fed into the AC amplifier. The AC amplifier lowers the baseline of the differential amplifier output (quasi-digital pulse) to the negative direction when the magnitude of the quasi-digital pulse becomes larger (Fig. 6a), while the baseline approaches 0V when the pulse height declines (Fig. 6b). These responses of the AC amplifier to the change in magnitude of the quasi-digital pulse act as an automatic threshold control, which help comparator to detect the S1 signal with a higher sensitivity (Fig. 6).

2.4 Heart Rate Calculation by Microprocessor Program

The cardiac beat detector improved the incidence of S1 detection by removing the influence of respiration sound noises, however, large-amplitude respiratory noises, which were elicited by an airflow

in airway with airway secretion, still remained and induced errors in S1-S1 interval detection for heart rate calculation. Discrimination between S1 and S2 is also required for the heart rate calculation. I made a microprocessor program to overcome these problems. The major algorithms adopted in the program are; (1) to calculate the correct HR by selecting four S1-S1 intervals of less error from eight consecutive intervals and (2) to set a non-detection period of 75 ms after a S1 (or S2) for the discrimination of S1 from S2 (Sato et al., 2006).

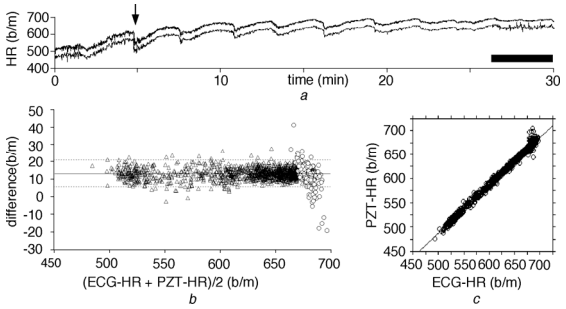

Figure 7: Comparison between the heart rate calculated by the PZT system and ECG for 30 min. Output of the PZT system (lower trace) and the heart rate calculated from the R-R intervals in ECG (upper trace) (a), their difference plot (b) and cross-correlogram (c). Black bar in (a) indicates the duration where large-amplitude respiratory noise appeared. Lower trace in (a) is lowered to show almost complete agreement between the two traces.

3 RESULT

Heart rate output from D/A converter in the PZT system (PZT-HR) and that calculated from ECG reading (ECG-HR) averaged over every 1 s were compared using 6 anesthetized adult C57BL/6 mice.

The PZT-HR and the ECG-HR were highly correlated (Fig. 7a). Difference plot between them also showed good correlation (Fig. 7b) even during the period when large-amplitude respiratory noise appeared (open circles; Fig. 7b). The difference plot demonstrated the highly reliable detection of HR by the PZT system; 96.2% (1,729/1,798) of total points fell within ±2 SD of the mean value. The PZT-HR also closely followed a rapid decrease in HR at a rate of 33 b/m/s (arrow in Fig. 7a). Cross-correlation coefficient between PZT-HR and ECG-HR was 0.995±0.003 (mean±SD, n = 6; Table 1, Fig. 7c).

Table 1: Correlation between PZT-HR and ECG-HR.

mouse	*r*	difference (%)
1	0.995	1.9±0.5
2	0.990	1.8±1.7
3	0.992	2.9±1.6
4	0.997	1.3±0.6
5	0.999	2.4±1.3
6	0.997	2.1±0.6

4 DISCUSSIONS

Since the high-frequency component of S1 is comprised of multi peaks of vibrating signal, the program code for heart rate calculation would be a complex one in the case without the use of the cardiac beat detector although recent developments in digital signal processing of the phonocardiogram have been reported (Durand and Pibarot, 1995; Wang et al, 2001). All intervals between peaks of S1 and S2 in addition to respiration sound noises, which are all composed of multi-peaked signal and fluctuate in interval and/or in magnitude, should be measured quickly and the initial point of the S1 should be properly identified almost instantaneously during each heart cycle of less than 100 ms. In contrast to such considerably complex digital signal processing, making the quasi-digital pulse from vibrating S1 signal with enhancing S/N ratio using the cardiac beat detector ensures the easier digital conversion of the S1 signal for the heart rate calculation.

In conclusion, the present study demonstrated that the cardiac beat detector has a performance suitable for the non-invasive detection of S1 in the heart sounds of small animals. It should be noted that the cardiac beat detector is available not only for anesthetized small animals but also unanesthetized animals and humans at sleep or rest. Indeed, the PZT system can be applied to unanesthetized newborn mice (Sato et al., 2007), human infants (Sato et al., 2006) or bedridden patients after some alteration to the sensor construction. As the cardiac beat detector greatly reduces the program code for S1 detection, it would help us to create novel phonocardiogram-based equipments for a wide range of fields in clinical and basic sciences in medicine.

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