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A Wearable ECG-HR Detector and Its Application to Automatic Assist-Mode Selection of an Electrically Assisted Bicycle

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***Abstract*—Recently, an electrically assisted bicycle has been widely used in daily life and becomes very popular. The user selects the stepwise assist-mode to determine the assistive torque for pleasurable running. From the viewpoint of improvement of health by exercise, the electrically assisted bicycle can be an exercise machine like a treadmill. The heart rate (HR) is regarded as an indication of exercise load. This paper presents an automatic assist-mode selection system based on the HR of the bicycle user. The HR is obtained from the R-waves measured by the proposed wearable electrocardiograph on the user. The mode-selection system is simply implemented by a personal computer, USB-connected interface, and some electronic switching circuits. The running experiments confirm that the proposed assist-mode selection method has practicability.**

I. Introduction

A bicycle is a vehicle with a very long history and is regarded as one of the ideal vehicles from the viewpoint of energy saving. Many kinds of bicycles have been widely used for various purposes: shopping, commuting, sporting, child's toy, and others. Recently, electrically assisted bicycles have been introduced and have become popular in the world. The technical evaluation of electrically assisted bicycles has been reported [1]. In 2008, the annual production of electrically assisted bicycles exceeded the total annual production of motor bikes (<50cc) in Japan.

The electrically assisted bicycle has a lot of advantages with respect to gradeability, heavy goods carrying, and long-distance traveling. In particular, the bicycle is useful for senior citizens. A few types of electrically assisted bicycles are capable of braking in regenerative mode. The usual electrically-assisted bicycle has a function of discontinuous or stepwise adjustment of assisted torque versus the user's torque ratio, i.e. the assistance level is manually selected by the user.

From the viewpoint of improvement of health by exercise, the electrically assisted bicycle can be used as an exercise machine like a treadmill and a cycle ergometer. The HR is regarded as an indication of exercise load [2]. Servo-controllers of HR using a treadmill or a cycle ergometer were reported in [3, 4]. In these researches, the HR of the user was obtained from the measured electrocardiogram (ECG) and was controlled at a constant by means of the speed control of the treadmill or the work rate control of the ergometer.

We have been studying a wearable capacitive electrocardiograph and its application to electrically assisted bicycles [5-7]. Unlike an indoor treadmill/ergometer system, several research problems are pointed out: noise from the outdoor circumstance, wireless data transmission from the user to the bicycle, etc. Our research takes these problems into account.

When an exercise machine is used for health exercise, the load strength of the exercise is so adjusted that it should stay in the range of 50-70 % of the maximum strength that the user personally has. In other words, the HR of the electrically assisted bicycle user should be kept within that range during cycling.

The purposes of the paper are (1) to examine the proposed wearable ECG-HR circuit on the bicycle in running condition, (2) to apply the wearable ECG-HR circuit to the assist-mode controller that adjusts the bicycle user's HR within the predetermined range, and (3) to verify the practicability of the proposed system by experiment.

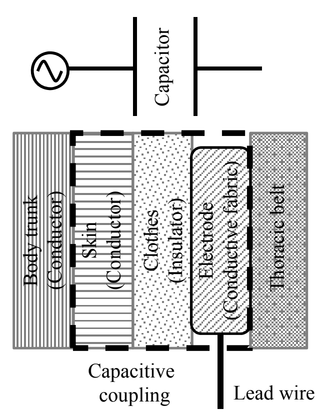

Fig. 1. A model of the capacitive electrode coupled to the skin with the clothes.

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II. WEARABLE ECG-HR DETECTOR

The method of the wearable ECG-HR detector is based on the measurement of the ac component of the active potentials evoked by the heart. The measurement method employs the capacitive coupling between the skin, the clothes, and the electrode, as shown in Fig. 1 [5].

As the conductivity of the clothes is not important, ordinary clothes are available. Conductive rubber or fabric, for example, can be used as an electrode instead of a metal electrode. The clothes work as an insulator and have a low relative-permittivity value. This may result in a low capacitance between the skin and electrode. A high input-impedance amplifier, buffer must be utilized in spite of a low signal-to-noise ratio (SNR), and the body motion artifact of the user. Therefore, a method for reducing the noise and artifact is indispensable.

A. Implementation of Capacitive Electrodes

The part of the electrodes is composed of a commercial thoracic belt and three sheets of conductive fabrics. The conductive cloths are flexible and have a low contact resistance; carbon-coated fabrics (Kitagawa Kogyo, CSTK) were employed. As shown in Fig. 2, the capacitive coupling forms on the part of the skin, the underwear and the electrodes at the inside of the thoracic belt. To reduce the effect of the baseline wander caused by the body motion, the two active electrodes, each of which is rectangular (60mm x 40mm), are set on the skin parts of the left and right costae. The reference rectangular electrode (60mm x 40mm) is placed on the epigastric fossa. All the electrodes are attached with a fine adjustment according to the personal body size.

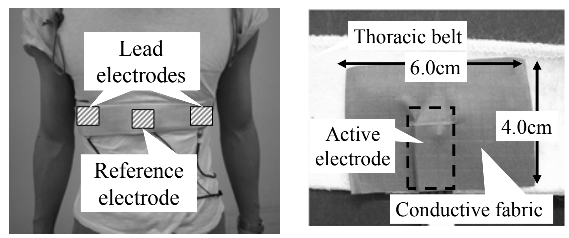
At the costae two active electrodes are used and usual operational-amplifier buffer circuits are installed in the conductive cloths. The use of active electrodes instead of passive ones can reduce the effects caused by the disturbance noises and the movement of the body. In addition, the active electrodes are not subjected to the input impedance decrease caused by the stray capacitance around the leads of the electrodes. The input buffer's input impedance is a repeat at the level of the electrodes. The integrated circuits (ICs) employed in the buffers have a rated input-impedance value of 1 [tera-Ohm] - 1 [pico-Farad].

B. The Implemented ECG-HR Circuit

The block diagram of the implemented ECG detector is shown in Fig. 3. The circuit is composed of two buffers for the active electrodes, an instrumentation amplifier, a notch-filter, a high-pass filter, a low-pass filter, and a pulse signal generator. In addition, a Driven-Ground-Plane (DRL) circuit was introduced to reduce the common-mode noise such as ham noise, and to improve SNR. The notch filter is used to eliminate 50-Hz noise. The high-pass filter is a 20-Hz multiple-feedback type with the forth-order Bessel characteristics. The low-pass filter is a 40-Hz multiple- feedback type second-order Bessel characteristics. The purpose of this circuit is to obtain HR from the ECG signal, and then the circuit simply has a narrow frequency range in comparison to the general ECG frequency range of 0.05-100 Hz. The total gain of the circuit devices is 3000.

The HR for the operation of the electrically assisted bicycle is obtained from the R-waves involved in the ECG. A pulse signal synchronized to the R-wave is generated by a comparator and a one-shot multivibrator in the pulse signal generator.

It is necessary to separate electrically the user from the bicycle to avoid wiring trouble when getting on and off. The wireless data transmittance is simply made by means of a wireless mouse for a personal computer (PC) in this research. The output voltage of the one-shot multivibrator is transferred to short-circuit electronically the click contacts of the wireless mouse. The USB mouse receiver is installed on the PC that works as a controller on the electrically assisted bicycle.



(a) (b)

Fig. 2. Images of the developed wearable electrode unit. (a) configuration of the wearable electrode unit and (b) side view of the electrode attached to the thoracic belt.

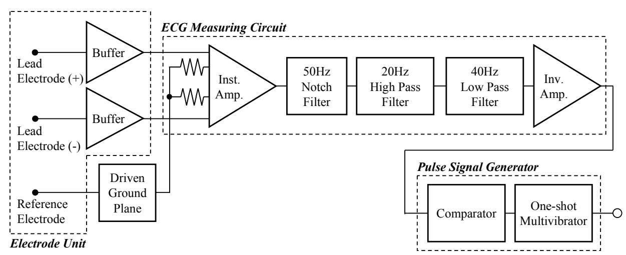


Fig. 3. Block diagram of the developed ECG-HR detector.

III. THE ASSIST-MODE SELECTOR

The PC on the electrically assisted bicycle has USB interfaces for the input and output signals, as shown in Fig. 4. One of the inputs is for the signal from the pulse signal generator and the other is for the bicycle speed. A few outputs for the controller of the bicycle are prepared; the number of the outputs (channels) is equal to the number of the assist-mode buttons of the bicycle controller. In general, electrically assisted bicycles have two or three modes of torque assistance.

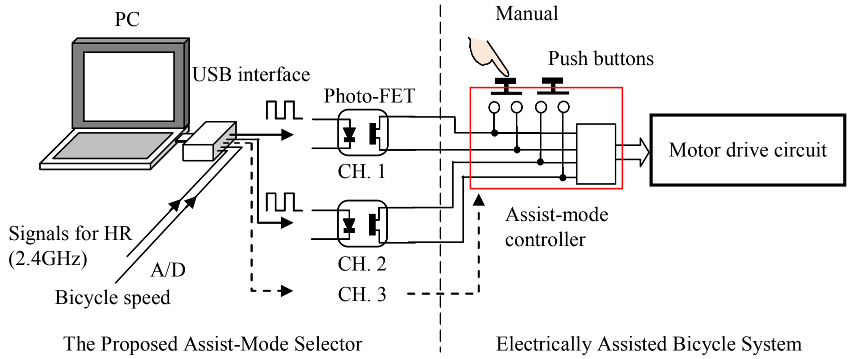


Fig.4. The proposed assist-mode selector.

A. Calculation of HR

The HR of the bicycle user is calculated as follows. The on/off signals synchronized with the R-waves in the ECG are inputted to the PC by way of the wireless mouse and USB connection (2.4 GHz). The PC counts the pulses in terms of the number of the mouse clicks. The HR is calculated every several counts (e.g. five counts), i.e. the HR is a reciprocal of the total time, T, of m intervals. Furthermore, the average value is obtained from this present HR value and the previous HR value to reduce unexpected fluctuations in the HR, as in the following equation:

$$h_N = \frac{1}{2}\left(\frac{60m}{T} + h_{N-1}\right) \tag{1}$$

where h_N, [bmp] and $h_{N\text{-}1}$ [bpm] are the present HR and the previous HR, respectively.

B. Assist-Mode Selector

The HR of the user on the electrically assisted bicycle is adjusted so as to be within a predetermined range. The selector automatically determines an appropriate assist-mode. If the HR is larger than the given threshold, the assist-mode is selected as a more strengthened mode. If the HR is smaller than the threshold, the assist-mode is switched to a weaker assist-mode. This adjustment makes the user get continually a proper assist-mode, in other words, a proper exercise load.

Fig. 5 shows a switching scheme of the assist-modes with respect to the HR. In this case, the electrically assisted bicycle is assumed to have three assist-modes: "Low," "Medium," and "High". In addition, the no-assist condition is expressed as "Zero". Thresholds h_1, h_2, and h_3 are placed with some hysteresis band to avoid chattering between the two adjacent modes. In Fig. 4, the circuits for push-buttons of the controller on the bicycle are represented. The signals from the PC short-circuit the contacts of the push-buttons by the photo-FETs. As the manual control is also available in this connection, the user can select the assist-mode as he/she likes.

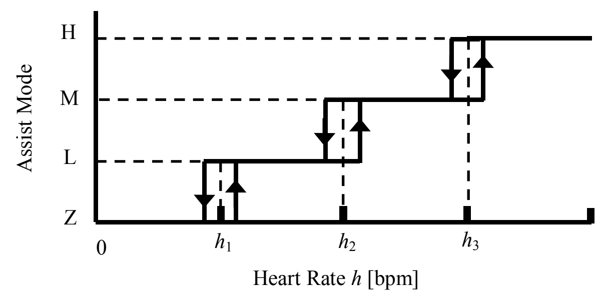


Fig. 5. Switching schemes of assist-mode with hysteresis.

IV. Experiment and Discussion

A. Evaluation of Wearable ECG-HR Detector

In the experiment the developed wearable ECG-HR detector was examined at three conditions: (1) resting on an electrically assisted bicycle, (2) pedaling the bicycle on a flat road, and (3) pedaling the bicycle on a slope. Ten volunteers were requested to perform each experiment with the detector for 2 minutes. As a reference signal the NASA-lead ECG was wirelessly measured with a commercial electrocardiograph. According to the results the detector operated stably; mean detection rates of R-wave at the experimental conditions were (1) 99%, (2) 97% and (3) 96%, respectively.

B. Evaluation of Assist-Mode Selector

In the experiment we used an electrically assisted bicycle, Yamaha PAS (PZ26LS) and attached the above-mentioned controller on it. The speed of the bicycle is measured by counting the pulses generated by the six magnets on the spokes of the rear wheel and magnetic sensor on the frame. A frequency-to-voltage conversion technique was used to obtain a dc voltage proportional to the speed. The speed voltage was transferred to PC by an A/D converter and USB interface.

The asphalt-paved test course is a circuit of 330m long (224m

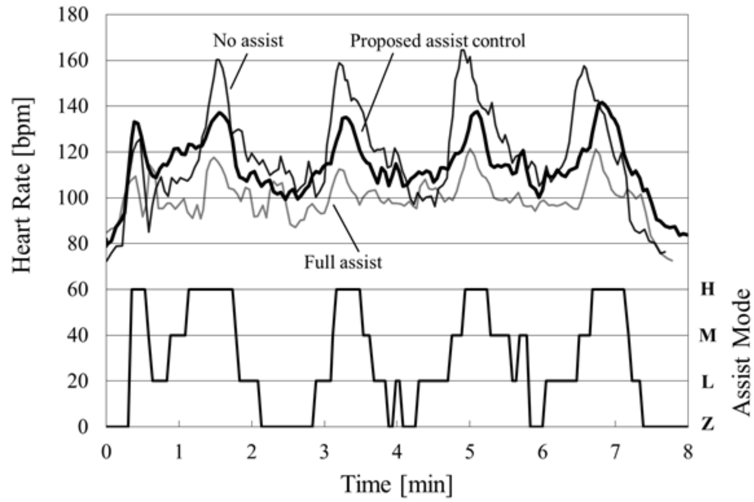


Fig. 6. Experimental results.

flat road + 106m slopes). The user runs up the 53m slope and then U-turned the same slope.

Fig. 6 shows the experimental results. The participant (male, age: 22) was in the normal health condition. He was requested to go round the test course 4 times at a speed of about 12 km/h. Some speed fluctuations were acceptable.

The bicycle running test was carried out for the three assist-modes: no-assist running, full-assist running, and running with the proposed control.

Each result in Fig. 6 shows that the HR of the participant synchronized with the road up/down conditions. At no-assist the HR rose sharply up to 160 bpm. At full-assist the HR was kept at a low value, and was not largely affected by the slopes. At the proposed assist-mode control the HR was in-between. The peak values of the HR caused by the slopes were decreased in comparison with that of the no-assist case. The peak-to-peak value of the HR at the proposed control was within the range of 100-141 bpm. The switching thresholds were set at h_1=108 bpm, h_2=113, h_3=120, and hysteresis band=4 (=2x2) bpm. In the experiments, these thresholds and the hysteresis bands were determined by trial and error. An appropriate method for predetermining the thresholds should be suggested for the next stage of the research that focuses on exercise training to show the clinical validity of the electrically assisted bicycle under consideration. The medical data of the user, such as weight, age, and heart rate at rest are important and will be taken into account.

V. Conclusion

We have discussed the wearable HR detection system and its application to an automatic selection method of the assist-modes for an electrically assisted bicycle. Running experiments were carried out. The proposed method is applicable to an electrically assisted bicycle with a push-button type controller. By using the appropriate thresholds set in the controller, the HR of the bicycle user is adjusted within a desirable percentage range of his/her maximum HR.

Developing a simpler method for predetermining such thresholds is a problem to be solved in the future. In addition, the derivative of the HR can be used for more sophisticated control.

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