ROBUST EAR LOCATED HEART RATE MONITOR

L. Rossini, R. Vetter, C. Verjus, P. Theurillat, P. Renevey, M. Bertschi and J. Krauss
Swiss Center for Electronics and Microtechnology, Jaquet-Droz 1, Neuchâtel, Switzerland

Keywords: Ear located heart rate monitor, Strapless heart rate monitor, Ear clip, Earphone, Portable audio players.

Abstract: We have developed a device for heart rate estimation with the optical sensing unit integrated in a classic media player earphone. The sensing principle is based on an optical infrared measurement directly on the ear lobe whereas the heart rate estimation is obtained utilizing robust model-based signal processing techniques valid even for quasi periodic activities such as running. Nevertheless, the remaining problem related to these strategies are statistically undetectable inter-beat during short term sporadic activities. In this paper, we present a novel robust inter-beat discarding method based on an activity related modeling of the expected heart rate dynamics that incorporates a simple cardiovascular model to reduce related inaccuracies in the heart rate estimation. A validation protocol has been designed and 9 subjects were asked to carry out their daily normal office activities for a time length ranging from 1 to 2 hours. A global absolute relative error mean of 0.9% between the estimated heart rate and a reference device and a sensitivity above 90% demonstrate encouraging performances of the proposed device.

1 INTRODUCTION

The advent of portable audio players is continuously increasing the popularity of exercising with music. Besides enjoying their favorite songs, people may find themselves often more motivated toward the achievement of a given training load (Elliott et al., 2005; Schie et al., 2008). However, to obtain a desired training effect such as weight-loss or improvement of the cardiovascular performance, the exercising person should observe and respect its personal heart rate target zones (Noakes, 2003). This requires supplementary equipment such as a heart rate monitor with its associated chest-strap ECG sensing. Often, due to the tightness of the strap, the heart rate monitor and chest-strap sensor may be perceived by users as oppressive and may consequently diminish the exhilarating and motivating feeling provided by music. In order to avoid this degradation, a device for heart rate estimation based on a sensing unit directly located in a classical audio player earphone has been presented recently in (Celka et al., 2004; Verjus et al., 2003). The sensing is based on an infrared measurement at the ear cartilage whereas the signal processing is directly performed in the audio player unit where auditory user feedback may be achieved. A critical aspect of this approach is the signal processing. Adaptive model-based enhancement of infrared signals is performed to obtain a robust heart rate estimation even for quasi periodic activities such as for example running. A remaining problem of this device is a reliable management of heart rate estimation during short term sporadic activities. Indeed, in this case, model based IR signal enhancement may not operate accurately due to the convergence time of the enhancement model (Haykin, 1991). Consequently, residual movement related artifacts might induce erroneous inter-beat detections that may not be detected with a statistical assessment (Vaseghi, 1996). In order to cope with this problem, we have developed a reliable discarding of erroneous inter-beat intervals based on an activity related modeling of the expected heart rate dynamics. The exploitation of previous biomedical knowledge allowed us to develop a simple cardiovascular model to reduce inaccuracies in the heart rate estimations.

Rossini L., Vetter R., Verjus C., Theurillat P., Renevey P., Bertschi M. and Krauss J. (2009).
ROBUST EAR LOCATED HEART RATE MONITOR.
In *Proceedings of the International Conference on Biomedical Electronics and Devices*, pages 214-219
DOI: 10.5220/0001541002140219

2 METHODS

2.1 Background

Optical probes for sensing biological tissue properties based on photoplestysmography (PPG) have been widely used over the past years for the estimation of cardiovascular parameters such as for example pulse oximetry and heart rate (Webster, 1997), (Tremper and Barker, 1989). Corruption of the PPG signal arises from the influences of ambient light and subject motion (Tremper and Barker, 1989) (Trivedi et al., 1997). Processing of ambient light artifacts is not critical since the influence can be measured using multiplexing techniques and an artifact free PPG signal can be restored using subtractive-type techniques (Trivedi et al., 1997). Various methods for improving the PPG technique during motion artifacts and low perfusion of the tissue have been designed (Coetzee and Elghazzawi, 2000). A very sound and robust approach of motion artifacts removal in PPG measured signals has been recently addressed (Celka et al., 2004). The parametric signal enhancement method exploits the information contained in a motion reference signal generated by a two-dimensional accelerometer in order to obtain a robust PPG heart rate estimation. Very reliable heart rate estimations have been obtained even under intense physical activity such as for instance running. However, because of the slow convergence of the enhancement algorithm, the heart rate estimation may become erroneous during sporadic, short, and transitory activities. The convergence time of the enhancement algorithm is in the range of the few seconds that are necessary to obtain sufficient accuracy of the enhancement parameters.

2.2 Sensor and Processing Device

In this paper we propose a fully integrated heart rate measurement device with sensing located at the ear and providing reliable estimates of the heart rate that are robust against short sporadic or transitory activities. This system is based on infrared optical measurement of the sub-cutaneous blood flow by transillumination, together with an integrated two-dimensional accelerometer (see Figure 1). The chosen optical wavelength is 875 nm. The emitter is a light emitting diode (LED) and the receiver is a photodetector (PD). The light wave is sent trough the ear cartilage and penetrates the skin and blood vessels to finally reach the PD. The PD transforms the received light intensity $I(t)$ into a current that is then transformed into a voltage. Subsequently, as shown in Figure 2, a compensation of the ambient light is performed using a subtractive multiplexing technique (Trivedi et al., 1997).

Figure 1: Ear located sensor device with portable processing unit.

The resulting signal $IR(t)$ together with the two analog acceleration signals $a_1(t)$ and $a_2(t)$ are then conditioned through amplification and a 2nd order Butterworth bandpass filter between 0.5 Hz and 3.5 Hz. Finally, an analog-to-digital (ADC) conversion is performed on these signals at a sampling frequency of 20 Hz. Signals are processed in the portable unit and the resulting heart rate estimations are displayed to the user. A direct auditory user feedback through the sensing earphone is an attractive option for a future development.

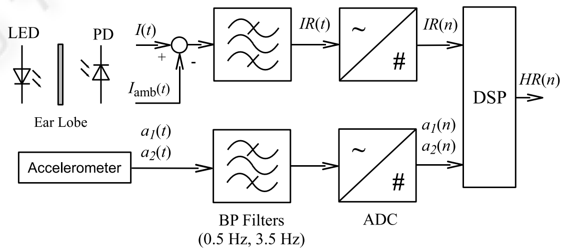

Figure 2: The optical concept and electronic signal conditioning of the proposed ear located heart rate estimation device.

2.3 Physical Principle

The principle of the proposed method resides in injecting an optical infrared (IR) signal at the surface of the body tissue and measuring the resulting optical signal. This signal propagates through the ear tissue where it is subject to modifications due to reflection, refraction, scattering and absorption. The resulting signal is captured by one or multiple optical sensors distributed on the earphone. For the near IR wavelength, the light propagation into the tissue is primarily governed by scattering and absorption (Cheong et al., 1990). The *Beer-Lambert* equation is generally used to describe the phenomenon of light absorption in biological tissue (Coetzee and Elghazzawi, 2000) relating the injected light I_i to the output light I_o. However, motion artifacts affect the components of the Beer-Lambert equation. Under this conditions, the received intensity can be written in terms of the major contributions:

$$I_o(t) = I_i(t) \cdot \gamma_{\text{tissue}} \cdot \gamma_{\text{pulse}}(t) \cdot \gamma_{\text{motion}}(t) \quad (1)$$

where γ_{tissue} is the static attenuation due to the tissue, $\gamma_{pulse}(t)$ is the pulsatile component due to variations in the sub-cutaneous blood flow, and $\gamma_{motion}(t)$ is the contribution due to dynamic changes of the tissue induced by movements of the head. The contribution of $\gamma_{pulse}(t)$ in equation 1 is of pivotal interest for the heart rate estimation. On the other hand, the time-invariant term γ_{tissue} is of no interest and can therefore be removed using low-pass filtering. Adaptive signal processing enhancement techniques of the signal $I_o(t)$ to cope with long term harmonic contribution of $\gamma_{motion}(t)$ have been previously successfully presented in (Celka et al., 2004). The technique developed during the present activity addresses the problem of the heart rate estimation during short and sporadic activities.

2.4 Proposed Algorithm

2.4.1 Concept

The concept of the proposed ear located heart rate estimation device is shown in Figure 3. It mainly consists of an enhancement of the motion corrupted IR signals, an inter-beat interval (IBI) extraction on the enhanced IR signal, a discarding of unreliable IBIs and a final estimation of the most likely heart rate through histogram clustering. The enhancement method that receives one part of its parameters from the activity analysis has been presented in detail in (Renevey et al., 2001). The histogram clustering is based on a standard method of statistical signal processing (Gersho and Gray, 1992). The new features that we introduce address the optimal sensor placement strategy, the IR pulse-amplitude reliability assessment, the activity analysis and the estimation of a prior of heart rate using a simple model of the cardiovascular dynamics. This prior estimation of heart rate and its associated confidence intervals together with the reliability index of the IR-pulse-amplitude are then used to discard unreliable IBIs.

2.4.2 IR-Pulse-Amplitude Reliability

The reliability index of IR-pulse-amplitude describes the likelihood that a given IBI has been extracted from an IR signal with an amplitude that corresponds to the expected amplitude of pulse contributions. It is obtained by evaluating the probability that the instantaneous IR amplitude has been generated by a process with mean $\mu_{IR,ref}$ and standard deviation $\sigma_{IR,ref}$, which are continuously re-estimated during periods without any movement activity.

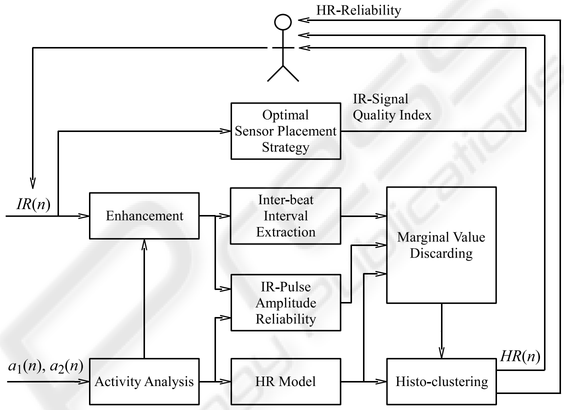

Figure 3: The proposed heart rate estimation algorithm based on motion artifact removal using accelerometer signals and discarding of unreliable inter-beat intervals using activity related cardiovascular modeling.

2.4.3 Optimal Sensor Placement Strategy

Due to peripheral vasoconstriction and under-optimal sensor placement, IR pulse contribution may be very low. Therefore, we introduce an optimal sensor placement procedure that evolves during the initialization phase of the device:

- The signal amplitude is fed back to the user by means of a quality index.
- As long as the quality index is below a given threshold, the user is asked to adjust the sensor placement to improve the signal quality.

We observed that this strategy leads to a sufficient quality of the IR pulse contribution. In about 2 % of the cases only, the subject was unable to properly adjust the device in an appropriate position. This procedure could also be processed fully automatically if a multi-sensor approach is used.

2.4.4 Activity based Model Heart Rate Model

The proposed ear located sensor is based on IR signals and as such is prone to movement related artifacts. During transitory periods when the enhancement method cannot converge, these artifacts may induce erroneous IBI detections. To discard such erroneous detections we improve the statistical robustness of the proposed heart rate estimator by introducing a modeling of the expected heart rate based on accelerometer measurements. The study and development of cardiovascular models has attracted a wide spreading interest in the scientific community (see (Le et al., 2008) and references therein). In our application, we are particularly interested in the description of the heart rate dynamics with respect to a given input activity. Such an approach has been recently proposed for cycling (Le et al., 2008). In particular, the instantaneous heart rate is modeled as a weighted function of its past value and the power generated by the cyclist that is estimated using commercial devices. Notice, however, that for activities like running and walking the generated power output is not obviously measurable. Nevertheless, during running and walking the exerciser's power output is approximately related to his speed and a method for speed estimation using core body located accelerometers has been recently presented (Vetter et al., 2008). Even though core body accelerometer measurements are not identical to ear located accelerometer measurement due to slight head movements during each stride cycle, we may assume that they are strongly related. We exploit the method presented in (Vetter et al., 2008) to obtain a rough measurement of the power output of a runner $P(n)$ as a function of the maximal eigenvalue of stride-wise speed variations.

The heart rate model estimation $HR_{\text{model}}(n)$ at sample n is a combination of the HR value before the exercise $HR_{\text{rest}}(n)$ and a heart rate increase $\Delta HR(n)$

$$HR_{\text{model}}(n) = HR_{\text{rest}}(n) + \Delta HR(n) \qquad (2)$$

The dynamic behavior of the heart rate variation under physical effort is described by

$$\Delta HR(n) = \alpha \Delta HR(n-1) + (1-\alpha) HR_{\text{ATR}}(n) \qquad (3)$$

where α determines the temporal behavior of our model. Specifically, we use α_{act} when the activity is increasing or stationary and α_{recov} during a recovery phase. For this model, the evolution of the heart rate is a function of both its past values and the innovation term $HR_{\text{ATR}}(n)$. The innovation term combines the power output of an aerobic effort $P(n)$ with the difference between the cardiac anaerobic threshold HR_{AT} and the heart rate at rest $HR_{\text{rest}}(n)$.

$$HR_{\text{ATR}}(n) = \alpha_P \left[HR_{\text{AT}} - HR_{\text{rest}}(n) \right] P(n) \qquad (4)$$

The main goal of our model is to obtain a rough estimation of the heart rate in order to discard non-plausible instantaneous estimations. In this sense, to the various unknown parameters α_P, α_{act}, α_{recov} and HR_{AT} of our model, we apply commonly utilized sport physiological values (Noakes, 2003). These values are then updated during the use of the device so as to correspond to a user's specific profile.

2.4.5 HR Reliability

Sporadic motion artifacts augment the number of discarded marginal values, reducing the reliability of the estimated HR. For this reason, together with the estimated HR, the user is provided with a HR reliability index defined as

$$HR_{\text{rel}}(n) = \frac{N_{IBI,\Delta T}(n)}{\Delta T \cdot \mu HR_{\Delta T}(n)} \qquad (5)$$

where ΔT is the window length multiple of the sampling time, $N_{IBI,\Delta T}(n)$ and $\mu HR_{\Delta T}(n)$ are respectively the number of valid IBIs and the expected mean HR at the sample n over the time span ΔT.

3 RESULTS

In order to validate the developed estimation technique, nine subjects were requested to follow an adequate experimental protocol. Subjects were asked to carry out their daily normal office work for a total time length comprised between 1 to 2 hours and to carry out 2 to 3 high intensity activities lasting approximately 10 to 20 seconds. In addition to the developed ear located heart rate device, subjects were equipped with a commercial chest-strap based heart rate monitor (POLAR, RS800) employed as a reference.

To illustrate the performances of the proposed algorithm, we present quantitative and qualitative results comparing the heart rate estimated by the proposed method to the POLAR reference. The most commonly applied quantitative assessments are based on the Mean Absolute Relative Error (MARE) and Mean Absolute Error (MAE):

$$\text{MARE} = 100 \frac{1}{N} \sum_{n=1}^{N} \frac{\left\| \hat{HR}(n) - HR_{\text{polar}}(n) \right\|}{HR_{\text{polar}}(n)} \qquad (6)$$

$$\text{MAE} = \frac{1}{N} \sum_{n=1}^{N} \left\| \hat{HR}(n) - HR_{\text{polar}}(n) \right\| \qquad (7)$$

Furthermore, we compute a mean reliability index (R_{HR}) associated to each measure. R_{HR} is obtained by adding all the HR reliabilities greater than 0.5 and normalizing by the measurement length N, where HR reliabilities have been previously defined in equation 5.

$$R_{HR} = 100\frac{1}{N}\sum_{n=1}^{N}(HR_{\mathrm{rel}}(n) > 0.5) \qquad (8)$$

The promising performances relative to the entire database are presented in Table 1. Indeed, for the whole database, the mean μ and standard deviation σ associated to the MARE are 0.9 and 0.6 respectively. It is important to observe that the objective of this validation is to demonstrate the ability of our device to accurately estimate the heart rate under baseline resting conditions without concentrating on heart rate variabilities (HRVs) (of the European Society of Cardiology et al., 1996). For this reason, we filtered the heart rate provided by the POLAR to eliminate HRVs above 0.04 Hz. Therefore, HRVs associated to the sympathetic and parasympathetic nervous system are not retained in the validation.

Table 1: Mean Absolute Relative error (MARE), Mean Absolute Error (MAE), and reliability for 9 subjects in baseline conditions with intermittent sporadic activities.

Subject	MARE [%]	MAE [bpm]	R_{HR} [%]
1	0.4	0.3	98
2	0.3	0.6	95
3	1.2	1.0	92
4	1.2	0.8	94
5	0.6	0.8	94
6	1.4	1.0	90
7	0.7	0.5	99
8	2.8	2.4	93
9	1.6	0.9	93
μ	0.9	1.1	94
σ	0.6	0.8	2.8

Although the MARE and MAE provide an information about the average estimation performance of the developed device when compared to the POLAR RS800, they both only partially describe the impression perceived by the user. Indeed, the estimation performance may be excellent for long periods despite being erroneous for some short time intervals. Therefore, to take into account this temporal variation, we also apply an analysis of the specific threshold sensitivity, namely, an assessment of the performance as the percentage of time where the absolute value of the error is lower than a given threshold κ. The specific threshold sensitivity of the algorithm is presented in Table 2. The percentage of time where the error in heart rate estimation is lower than the indicated threshold κ highlights that low sensitivities are obtained for a threshold of 1% and 1bpm respectively. However, since the accuracy of the reference heart rate monitor is about $\pm 1\%$, the analysis may not be of high statistical relevance. A validation with such an accuracy should be performed using medical reference devices. In contrast, sensitivities for a threshold of 3% and 5% are above or equal to 90%.

Table 2: Percentage of time where the error in heart rate estimation is lower than the indicated threshold κ .

	κ					
	%			bpm		
Subject	1	3	5	1	3	5
1	92	99	100	95	100	100
2	90	95	97	93	97	99
3	56	88	96	62	90	97
4	68	89	95	76	92	97
5	74	94	98	82	97	99
6	48	85	94	62	91	97
7	77	98	100	84	99	100
8	33	68	83	46	80	91
9	62	83	92	72	92	97
μ	81	90	95	75	93	97
σ	19	10	5	16	6	3

Figure 4 illustrates qualitative performances for the first subject in the database. One can observe the high accuracy of the proposed method. Notice that at the beginning of the recording, the IR signal quality was insufficient to provide a valid heart rate estimation because the user was unable to properly adjust the ear located sensor. However, once the sensor placement strategy is successfully achieved, heart rate estimation converged as confirmed by a MARE of 0.4% and relative heart rate estimation error lower than 1bpm for 99% of the time (see subject number 1 of Table 1 and Table 2).

In Figure 5 we depict the result relative to subject 8. A closer visual inspection on the data highlights that there are mainly three segments where the results of the POLAR and of our device diverged. To illustrate, we observe two segments at the beginning of the recording where the heart rate estimated by POLAR is about 170 bpm despite weak accelerometer signals corroborating that the subject was in a resting position. This unlikely estimation may be the result of an incorrect manipulation of the chest strap reference device such as for instance an insufficient humidification. Finally, for the third segment at about 1300 seconds where we notice a difference between the two heart rate estimates there is an insufficient reliability $HR_{\mathrm{rel}(n)}$ of the heart rate provided by our device.

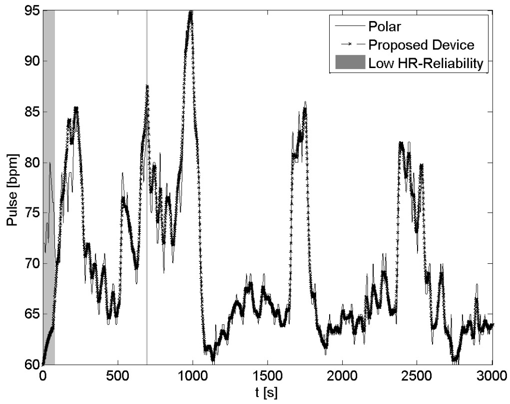

Figure 4: Heart rate estimation for subject 1 in baseline condition with sporadic intermittent activities using the proposed algorithm.

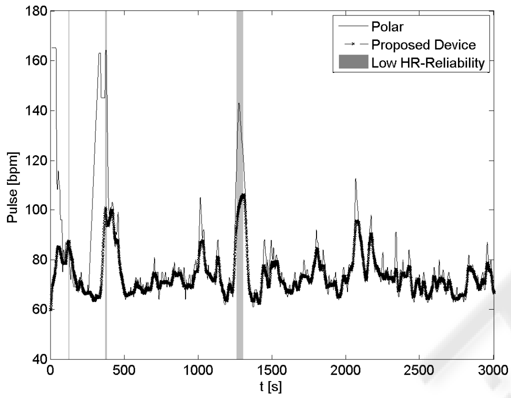

Figure 5: Heart rate estimation for subject 8 in baseline condition with sporadic intermittent activities.

4 CONCLUSIONS

In this paper we have presented an innovative approach for heart rate estimation using infrared optical measurements of the tissue at the ear lobe. The sensing device integrated in a classical audio player earphone improves the user's comfort with respect to commonly used chest-strap based heart rate monitors. The challenge connected to statistically undetectable inter-beat during short term sporadic activities has been approached using a novel robust inter-beat discarding method. The latter incorporates a simple cardiovascular dynamic model to reduce related inaccuracies in the heart rate estimation. Nine subjects were asked to participate in the validation protocol of the proposed device. The validation resulted in positive performances with an absolute relative error mean of 0.9% and a threshold sensitivity above 90% relative to the chest-strap heart reference monitor.

REFERENCES

Celka, P., Verjus, C., Vetter, R., Renevey, P., and Neuman, V. (2004). Motion resistant earphone locatedinfrared based heart rate measurement device. In *IASTED Conference on Biomedical Engineering (BioMED 2004)*, pages 582–585.

Cheong, W.-F., Prahl, S., and Welch, A. (1990). A review of the optical properties of bilogical tissues. *IEEE Journal Quantum Electronic*, 26:2166–2185.

Coetzee, F. M. and Elghazzawi, Z. (2000). Noise-resistant oximetry using a synthetic reference signal. *IEEE Trans. on Biomedical Engineering*, 47(8):1018–1026.

Elliott, D., Carr, S., and Omre, D. (2005). The effect of motivational music on sub-maximal exercise. *European Journal of Sport Medicine*, 5:97–106.

Gersho, A. and Gray, R. (1992). *Vector Quantization and Signal Compression*. Kluwer Academic Publishers.

Haykin, S. (1991). *Adaptive Filter Theory*. Prentice Hall.

Le, A., Jaitner, T., Tobias, F., and Litz, L. (2008). A dynamic heart rate prediction model for training optimization in cycling. In *The Engineering of Sport(ISEA2008)*, volume 1, pages 425–433. Springer.

Noakes, T. (2003). *Lore of Running*. Ed. Champaign.

of the European Society of Cardiology, T. F., the North American Society of Pacing, and Electrophysiology (1996). Heart rate variability - standards of measurement, physiological interpretation, and clinical use. *Circ.*, 93:1043–1063.

Renevey, P., Vetter, R., Krauss, J., Celka, P., and Depeursinge, Y. (2001). Wrist-located pulse detectionusing ir signals, activity and nonlinear artifact cancellation. In *IEEE EMBS Conference*.

Schie, N., A.Stuart, Becker, P., and Rogers, G. (2008). Effect of music on sub-maximal cycling. *South African Journal of Sport Medicine*, 20:28–31.

Tremper, K. and Barker, S. (1989). Pulse oximetry. *Anesthesiology*, 70:98–108.

Trivedi, N., Ghouri, A., Shah, N., Lai, E., and Barker, S. (1997). Effect of motion, ambient light, and hypoperfusion on pulse oximeter function. *Journal of Clinical Anesthesia*, 9:179–183.

Vaseghi, S. (1996). *Advanced signal processing and digital noise reduction*. Wiley Teubner.

Verjus, C., Vetter, R., Celka, P., and Renevey, P. (2003). Equipement portable destine a la mesure et/ou la surveillance de la frequence cardiaque. In *European Patent EP 1 374 763, US Patent 7175601*.

Vetter, R., Onillon, E., and Bertschi, M. (2008). Estimation of a runner's speed based on chest-belt integrated inertial sensors. In *The Engineering of Sport(ISEA2008)*, volume 1, pages 151–159. Springer.

Webster, J. (1997). *Design of pulse oximeters*. IoP Publishing.