CHAPTER

2.3

Application of Optical Heart Rate Monitoring

Mathieu Lemay, Mattia Bertschi, Josep Sola, Philippe Renevey, Jakub Parak, and Ilkka Korhonen

Tampere University of Technology, Tampere, Finland

1. INTRODUCTION

Since the introduction of portable devices in 1957 by Dr. Norman Holter, heart rate (HR) monitors have been extensively used in clinical practices as a diagnostic and prognostic tool, mainly for cardiovascular diseases. In parallel, the use of HR monitoring devices by sport physiologists to analyze the response of the body to exercise or training stress and to evaluate the training level of athletes has gained popularity. With the increasing interest of consumers to health and wellness monitoring, an explosion of commercialized HR monitors based on various technologies has been observed during the last decade. This chapter will give an overview of the different techniques actually known to monitor HR, focusing in particular on the so-called photoplethysmographic (PPG) technique. The history of the PPG technique, its basics, including measurement principles, measurement sites, quality factors and its applications to sport, fitness, daily life, and healthcare, will be addressed in respective sections.

The heart is the muscle in charge of pumping the blood throughout the blood vessels, namely the arteries and the veins carrying oxygenated and deoxygenated blood, respectively. The regulation of the rate at which the heart is pumping is mainly controlled by the autonomic nervous system. This system comprises sympathetic and parasympathetic (or vagal) functions, which are partially complementary. The sympathetic nervous system is responsible for increasing HR, constricting blood vessels, and other "stress" responses, while the parasympathetic system promotes maintenance of the body at rest (e.g., slowing

down HR) [1]. The autonomic modulation of HR can be altered by health status, training and over-training conditions [2], mental stress and other anxiety forms, cardiac disorders, as well as other critical illnesses or injuries [3]. Therefore, the analysis of HR and its variability (HRV) provide quantitative information on the modulation of cardiac parasympathetic and sympathetic nerve inputs, and consequently constitutes one of the major tools to evaluate the health conditions of a subject, to improve the training and recovery of athletes, or on the medical side, to diagnose important diseases such as autonomous neuropathy, cardiac arrhythmia, or infarction, etc.

There are different techniques to non-invasively monitor HR. The following section summarizes the most popular ones; namely, the techniques based on bio-potential, electric-acoustic, ultrasound, and bio-electrical measurements. The novel, but nonetheless promising, approach based on PPG is finally introduced.

Electrocardiography (ECG) is a bio-potential technique aiming at monitoring the electrical activity of the heart, and constitutes the gold-standard technique to monitor HR [4]. In short, the contraction/relaxation cycle of heart cells is associated with periods of electrical depolarization (increasing the potential) and repolarization (decreasing the potential), respectively. These periods induce local electrical dipoles, which generate surface potentials. ECG monitors the resulting surface potentials observed on specific thorax locations. The delay between the depolarization/repolarization of different regions of the heart produces the typical PQRST waves that characterize the ECG signal of a healthy subject (Figure 1).

The HR is habitually expressed in beats per minute (bpm). However, its value is derived from consecutive heartbeat intervals expressed in milliseconds and measured on ECG signals. On ECG signals, heartbeat interval is defined as the time delay between two consecutive heartbeats. It is usually measured as the time delay between two R-wave

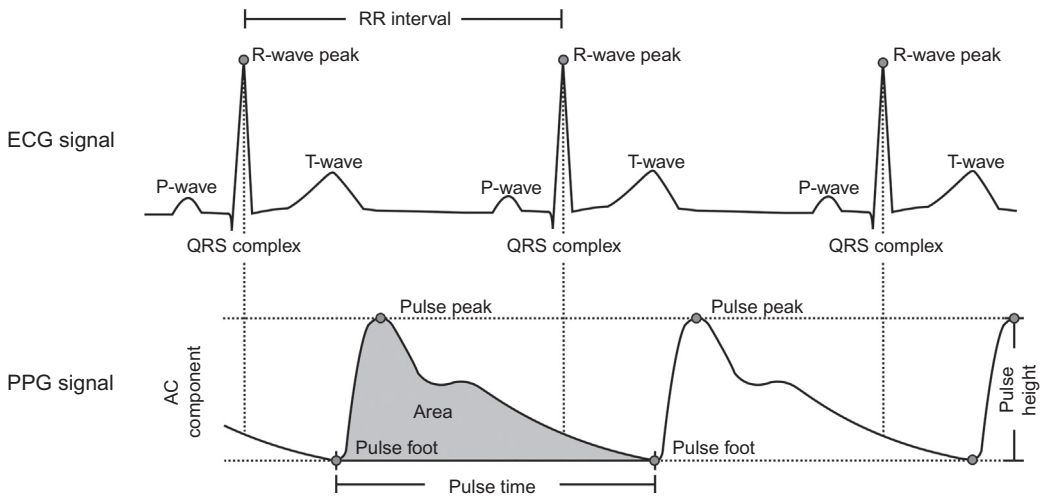

FIGURE 1 Typical synchronized electrocardiogram (ECG) and photoplethysmographic (PPG) waveforms and their respective components.

peaks and commonly referred as RR intervals (see Figure 1). These R-wave peaks, which denote the early depolarization of the ventricles (lower heart chambers), are correlated with the heart contraction [5]. Conventionally, these time series of heartbeat intervals are converted into bpm as follows: $(1/RR\ interval\ values \times 60)$. If desired, these heartbeat interval series are analyzed in the time- and frequency-domains to extract HRV features correlated to the influence of the parasympathetic and sympathetic nervous systems [6]. Physiological complex indicators such as stress level, physical recovery and sleep quality can also be derived from these HRV features [7].

ECG signals are classically acquired by placing silver/silver chloride electrodes on defined anatomical locations, and by connecting them to monitoring platforms. When the ECG monitoring systems are wearable, they are specifically referred as Holter systems. A large variety of alternative devices that monitor averaged HR values (one HR value over a specific time window) or heartbeat intervals based on bio-potential measurements exists. Most of them are employed during sport or daily activities (see [8–10] for examples of commercial devices). Their bio-potential sensors are based on gel, dry, or textile electrode principles [11]. There is also a family of strapless/wireless devices that temporally estimates fingertip bio-potentials using sensors embedded into watches (including Timex's Health Touch Plus [12] and Salutron's SmartHealth [13]).

Phonocardiography (PCG) is an electric-acoustic technology that allows the measurement and analysis of heartbeat sounds. The closing of the atrioventricular valves produces a characteristic sound labeled S1, which is easily identified from PCG signals. These S1 events can be used to estimate heartbeat intervals and consequently to estimate HR. *Echocardiography* is a sonogram (ultrasound measurements) of the heart that monitors the tissue and blood motions related to the pumping action of the heart. It is usually used to visually diagnose specific heart diseases such as heart defect or valve dysfunctions.

Impedance cardiography (ICG) is a bio-electrical technique aimed at monitoring cardio-related displacements of blood within the thorax by injecting electrical currents within the thoracic cavity and measuring resulting changes in the voltages [14]. This technique is usually used to estimate the amount of blood ejection. To our knowledge, no commercialized PCG-, echocardiography-, or ICG-based devices are dedicated to HR monitoring.

Photoplethysmography (PPG) is an optical technology aimed at measuring tissue light propagation changes during cardiac cycle. Its most popular application is monitoring of subject's oxygen saturation (pulse oximetry) [15]. For this purpose, two wavelength lights are used to estimate the arterial blood absorbance, which is linked to the blood oxygenation level. Various commercialized devices are available, including Nonin's [16], Masimo's [17], and Covidien's [18] devices. An extension of this application, namely near-infrared spectroscopy, estimates both oxygenation and deoxygenation of blood on a peripheral scale such as tissue and bones. In the HR monitoring context, the measurement of volumetric changes of a microvascular bed of tissue due to blood flow is the target [19]. This measure brings information on arterial pulsatility content (see Figure 1). Wearable HR monitoring devices using this technology are already available on the market (including Nonin's Onyx 2 [16], MIO's Alpha [20], Basis [21], and Impact Sports Technology's ePulse 2 [22] products).

2. PHOTOPLETHYSMOGRAPHY BASICS

2.1 History

Early in 1936, two independent research groups in New Jersey and Stanford explored the use of a non-invasive optical instrument to assess blood volume changes in rabbit ears [19]. One year later, a first study on the use of PPG to measure blood volume changes in human fingers was published by the team of Alrick Hertzman in St. Louis (US), paving the way toward the introduction of PPG in human monitoring. At that point, the term photoplethysmography was adopted in order to etymologically depict a new technique that could measure changes of volume (*plethysmography*) by optical means (*photo*).

For several decades PPG technology was restrained to physiological studies, kept apart from actual clinical practices. The fact that bulky light sources and sophisticated processing/visualization means were required to acquire and interpret the PPG signals limited wider regular use [19].

It was not until the later appearance of light-emitting diode (LED) technology in 1962 that PPG techniques raised the enthusiasm of a new generation of researchers. The fact that a PPG optical setting could be simplified to a simple LED and a photodetector opened the door to dozens of out-of-lab applications. In particular, the introduction of PPG in clinical routine was undeniably triggered by the development of a so-called pulse oximeter in 1972 by a team of engineers at Nihon Kohden labs [23]. Pulse oximetry was created as a non-invasive spectrometric technique that could provide first-ever real-time estimates on arterial blood gas content by simply placing an optical probe around the fingertip. Since then, pulse oximeters have penetrated many single operating theaters, intensive care units, and practitioners' offices, creating a worldwide market of over a billion dollars. While the original dual-wavelength pulse oximeters of Nihon Kohden provided estimates of arterial oxygen saturation (SpO_2), Masimo Corporation recursively improved the pulse oximetry technology by introducing the concept of perfusion index (a PPG-derived estimate of arterial pulsatility) in 1995, and the concept of pleth variability index (a PPG-derived estimate pulse pressure variation) in 2007. Finally, in 2011, Masimo commercialized the first multi-wavelength pulse oximeter providing simultaneous estimates of arterial saturation on oxygen, carboxyhemoglobin, and methemoglobin.

The introduction of pulse oximetry in clinical routine is undoubtedly associated with the generalized acceptance of PPG as a non-invasive monitoring technique being low cost, unobtrusive, and easy to use. In particular, during the past decades, PPG-derived approaches have been investigated for the assessment of parameters such as HR [24], blood pressure via volume unloading techniques [25], blood pressure via pulse transit time techniques [26], and endothelial dysfunction, among others. However, until very recently, its application to ambulatory HR monitoring was limited due to PPG's sensitivity to movement artifacts.

2.2 Measurement Principles

PPG technique relies on illuminating a living tissue with a light beam, capturing a portion of the light that has propagated through the living tissue, and analyzing said captured

light, depicting functional or structural information on the tissue. The attenuation of light, from the light beam (source) to the photodetector (signal), is typically modeled by the Beer-Lambert law. This law states that in a homogeneous medium, light intensity decays exponentially as a function of path length (l) and light absorption coefficient (α) corresponding to medium properties at a specific wavelength. Accordingly, and assuming the intensity of an injected monochromatic light beam being I_0, one expects the intensity of the transmitted light through the medium to be

$$I = I_0 e^{-\alpha l} \tag{1}$$

The properties of the Beer-Lambert law are valid if more than one substance absorbs light in the medium or if a succession of several media is foreseen. In both cases, each absorber contributes to the total absorbance, as a sum of the individual absorbance. The Beer-Lambert law suggests that the sum of the transmitted and absorbed light is equal to the incident light (Figure 2). Reflection at medium surface as well as other physical processes (e.g., light scattering) are not contemplated by this model. The Beer-Lambert law helps in understanding the absorbance of light traveling through homogeneous layers. However, the blood and other biological tissues are not homogeneous, quite contrary, and therefore absorption of light passing through is not simply proportional to the concentration of hemoglobin and to the optical path length. Blood is an inhomogeneous liquid exhibiting a nonlinear absorbance of light. Absorbance and scattering varies during the cardiac cycle with respect to the orientation of red blood cells during the contraction and relaxation periods of the heart. Absorbance is increased because of light reflection at the skin surface and multiple scattering effects, causing the deviation of the light beam from its initial direction. Furthermore, skin and other tissues are in homogeneous, and the variation of their structures and shapes (mainly due to movement) cause complex changes in the light reflection and absorption.

A living tissue can be modeled as a concatenation of several media, each one being characterized by a different path length and light absorption coefficient. Assuming now

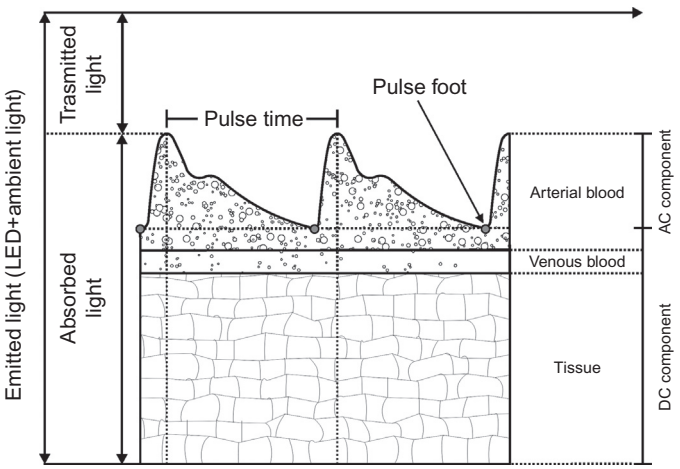

FIGURE 2 Simplified representation of the components of the PPG signal. The AC component due to pulsating arterial blood absorption and the DC component due to a sum of arterial blood, venous blood, and other tissues are displayed *(adapted from [15])*.

that at least one of the illuminated media represents an artery or a vein, each time the heart is beating, a blood pressure pulse is generated and propagates in this blood vessel. When a local increase of the blood pressure occurs, it modifies both the geometry (due to volume change) and the properties (due to changes in blood composition and concentration) of the medium representing the blood vessel. This results in an increase of the light absorption and an attenuation of the transmitted light intensity. The volumetric changes of venous and arterial blood highly contribute to the observed PPG signal variations. These variations are commonly divided into two components universally referred to as AC and DC components (see Figure 2). This nomenclature derives from the electrical engineering domain, where AC indicates a periodically varying level of voltage and DC a static level of voltage. Similarly, in PPG signals, AC refers to the pulsatile arterial blood, while DC refers to the "constant" light absorption due to tissue, venous blood, and diastolic volume of the arterial blood. In reality, the DC component is not constant but varies slowly, typically over several heartbeats. The main factors affecting the DC fluctuations are respiratory and vasomotor activities, and thermoregulation (as described in a later section).

The transmitted light captured by the photodetector might come from two different modes or pathways, as shown in Figure 3. In "transmission" mode, the tissue is illuminated at one side and the light transmitted through it is gathered at the other side. Unfortunately, not all body locations are prone to be monitored via transmission PPG measurements. When aiming at performing PPG analysis at body locations such as the forehead, the sternum, or the ankle, the emitted light is completely absorbed before reaching the opposite side of the body. In these conditions, an alternative operational configuration is available: the "reflectance" mode. In reflectance, the light source is placed next to the detector onto the skin surface, and the predominant light interaction is that of scattering.

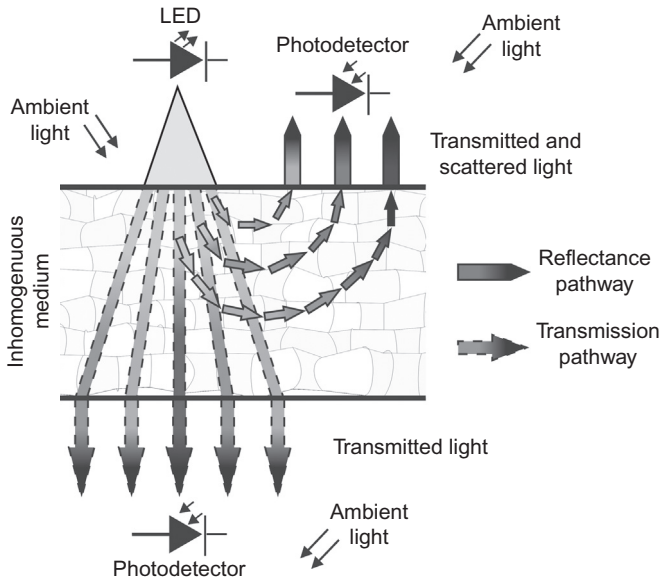

FIGURE 3 Transmission versus reflectance light modes: the roles of absorption and scattering mechanisms *(adapted from [15])*.

The wavelength of the injected light beam is also of paramount importance in the interaction of light and tissues. Each tissue constituent exhibits a specific optical behavior when traversed by a precise wavelength. The absorption spectrum shown in Figure 4 represents the optical behavior in terms of coefficient of absorption/extinction of a particular molecule with respect to light wavelength. The main constituent of tissue, namely water (H_2O), depicts in its absorption spectrum a window that allows wavelengths shorter than 950 nm to be transmitted more efficiently. Melanin is another constituent of tissue that strongly absorbs light wavelengths shorter than 500 nm, and its skin concentration depends on skin pigmentation. Hemoglobin (Hb) is the principal constituent of blood, whose absorbing characteristics change with its chemical binding. Hb molecules that are not able to bind reversibly with molecular oxygen (O_2) are called dysfunctional hemoglobin (e.g., methemoglobin, carboxyhemoglobin, and sulfhemoglobin). Functional hemoglobin is called oxyhemoglobin (HbO_2) if it is fully saturated with oxygen (i.e., carrying four O_2 molecules) and reduced Hb if it is not fully saturated. In healthy persons, most of the Hb molecules are of the functional type.

The choice of the wavelength at which absorbance is monitored is a trade-off and depends on the targeted application, but is usually in the 510 to 920 nm range corresponding to green and infrared lights, respectively. Measurements done on light skins and at normal ambient temperature (around 20°C) have shown that reflected green light has an advantage in terms of AC/DC component ratio over reflected infrared light [27], and might therefore be more suitable for ambulatory monitoring applications. The longer the wavelength is, the deeper the light penetrates, and the scattering effects associated with infrared light in deeper tissues produce a more complex reflected signal. However, in cold ambient conditions, blood microcirculation dramatically decreases and it becomes an

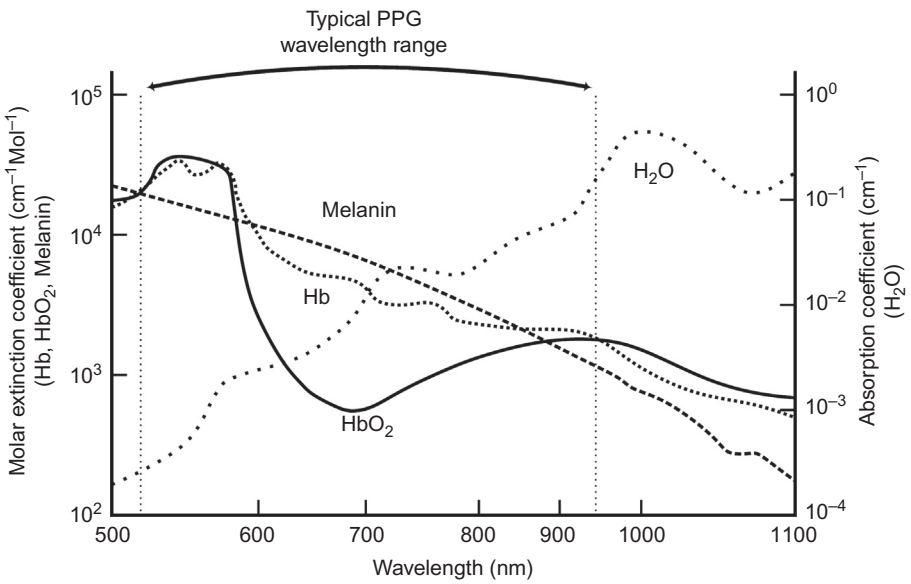

FIGURE 4 Absorption and molar extinction coefficients of main biological tissue constituents (H_2O, Hb, HbO_2, and Melanin) at 500 to 1100 nm window wavelengths.

advantage to reach deeper tissues. The dark skin pigmentation (high melanin concentration) strongly absorbed wavelengths shorter than 650 nm. In these two conditions, infrared light is desired. Therefore, selection of an optimal wavelength for ambulatory HR monitoring depends on targeted applications and usage conditions, and is a compromise between competing factors (vulnerability to artifacts vs. sensitivity during poor skin perfusion).

The measurement of PPG signals requires the use of at least one light-emitting and one light-receiving means. LED is the preferred light source for HR monitoring mainly because of its compactness, low cost, simplicity in use, and limited power consumption within a narrow bandwidth. Common photodetectors (light sensors) used in PPG measurements are photodiodes, but photocells and phototransistors can also be used. The signal coming from the photodetector is pre-amplified, filtered, and digitalized at a fixed sampling frequency, usually of about 25 Hz. For more details regarding the electronic schematic, see [15].

As described earlier, the PPG signal is a combination of DC and AC components as displayed by Figure 1 and Figure 2. The DC component, related to static component, has slow baseline variations and defines the pulse foot. The AC component is characterized by a sequence of PPG waveforms. The shape of the PPG waveform is not unique, but always depicts an inflow phase characterized by a steep rising wave and a runoff phase characterized by a slow decrease of the amplitude. Several parameters can be extracted from the PPG waveform to describe the AC component, and the most commonly used, displayed in Figure 1, are the pulse foot, defined as the lowest amplitude at the beginning of each pulse; the pulse peak, defined as the maximum amplitude between two pulse feet; the pulse height, defined as the amplitude difference between the peak and foot of the pulse; and the pulse time, defined as the time lasted between two following pulse feet (or peak). Other parameters can be found in the literature to evaluate additional physiological parameters, namely pulse area, propagation time, crest time, and inclination time [28].

2.3 Measurement Sites

The most common measurement sites for transmission mode PPG are the fingertip and earlobe, but other measurement sites are also possible, e.g., toes. For the reflectance mode, more variations in the measurement site are possible, including the forearm, the wrist, the ankle, the forehead, and the torso [24,29–31]. One of the practical differences between these modes is that the transmission mode sensors usually use a cuff or a clip to attach the sensor. This causes a probe-dependent increase in transmural pressure, potentially sufficient to collapse the low-pressure venous system, and hence slow changes in the local peripheral blood volume, leading finally to suppression of venous oscillations. The reflectance mode sensors are usually attached to the patient and hence may not apply enough pressure to collapse the veins. Therefore, this mode may be more beneficial if the venous components of PPG variations are of interest [29], and may also be applied to areas that are less affected by vasoconstriction [31]. However, if attachment of the reflectance mode PPG sensor is done by applying pressure to the tissue, it may suffer from the same phenomenon. Optimal measurement site depends, again, on the targeted application. Different anatomical sites differ in terms of density of microvasculature close to skin, skin thickness and structure, tissue characteristics below the skin (e.g., amount and structure of

fat tissue, muscles, large vessels, and tendons), and amount of movements during typical physical activities. Furthermore, usability and user acceptance issues play an important role in this selection. For example, arm and forearm are less prone to movement artifacts and variations in environmental temperature as compared to wrist [32], while wrist may be more accepted as a location by the consumers for long-term use.

2.4 Factors Affecting the Quality of Signal

PPG measurements derive from a complex interaction between light propagation and tissue characteristics. In the following section, an overview model of the PPG phenomena is illustrated. As depicted by Figure 5, the PPG signal derived from the PPG phenomenon is determined by three families of factors: sensing, cardiovascular, and biological factors.

PPG measurements are highly determined by the implemented sensing setup: the amount and nature of emitted light, the coupling between the skin, tissue, and the optical probe, and the response of the photodetector will influence the measured PPG signal.

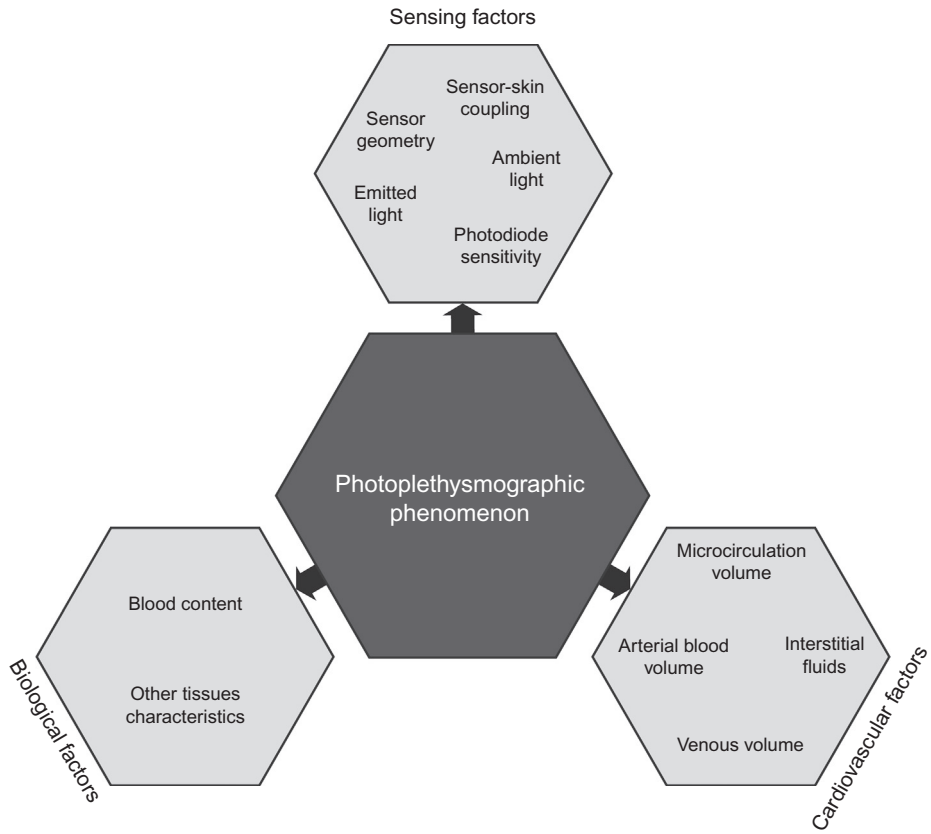

FIGURE 5 An overview model of the PPG phenomena and its three families of factors that influence PPG signal.

Sensor geometry and ergonomics constitute an important factor that will also highly determine the so-called optical shunting effect, which is the amount of direct light traveling from the light emitter toward the photodetector without penetrating the biological tissue. Perturbations due to ambient light on PPG measurements are to be minimized as well by an adequate sensor design. Optimal distance between light emitters and receivers is to be chosen as a trade-off between the desired depths of tissue penetration and the achievable light intensity to be injected into the skin. Larger distances will depict deeper penetration of photons through the scattering process while smaller distances will require smaller light intensities to achieve a reasonable amount of photons reaching the photodiode. Empirical studies have shown that optimal separation distances are in the range of 6 to 10 mm for infrared light [33] and $\sim$2 mm for green light [34]. Reflectance PPG measurements should be avoided in regions where large arteries induce pulsatility in all surrounding tissues [35]. In some body locations, reflectance PPG relying on light absorption (instead of scattering) can be achieved as well. In such operational configurations a bone surface is used to reflect part of the injected photons back.

Concerning the biological factors, the inherent tissue characteristics such as blood content and skin pigmentation have important impacts on the amplitude of the measured PPG signal by modifying the absorption and scattering properties of the tissue. Time-variant cardiovascular factors influenced by body position, age, and cardiovascular stresses also influence the nature and morphology of the measured PPG waveforms, and more precisely, the AC component behavior.

The next section addresses the problem related to PPG motion artifacts in the context of HR monitoring. The different sources of motion artifacts related or not to the sensing factors are detailed and appropriate optomechanical and signal-processing designs are proposed.

2.5 Motion Artifact Minimization and Removal

Due to its measurement principle, the PPG signals are quite sensitive to motion artifacts. The origins of these artifacts are multiples and often occur concomitantly. Three different causes can generate the motion artifacts.

2.5.1 Tissue Modifications Due to Movements

Voluntary or involuntary movements produce modifications of the inner tissues (e.g., motion of the muscles and tendons, and compression or dilatation of the tissues) that change the content of the tissues spanned by the light and thus modify the received signals. The motion-related acceleration or the gravity also affects especially the shape of the soft tissues (e.g., fat) and changes the distribution of the fluids in the tissue due to inertial forces. These factors result in changes along the optical path and modify the received optical signals. These modifications depend on the location of the optical probes on the body (e.g., an ear-located sensor is less prone to be affected by such artifacts than a wrist-located sensor) as well as the mechanism for how the sensor is attached to the skin (sensor pressure, mechanical support provided by the sensor housing, and possible strap, etc.).

2.5.2 *Relative Motion of the Sensor-Skin Interface*

The optical probe is attached to the skin via some binding (e.g., clutch, strap, adhesive media, and clothes). This link is not perfectly rigid and operates therefore as a mass-spring system. A local or global movement of the body generates accelerations that may produce a displacement of the sensor relative to the skin surface. This displacement changes the optical path of light and, as the tissues are generally not homogeneous, modifies the optical signals.

2.5.3 *Changes in the Pressure between the Optical Probe and the Skin*

The pressure applied by the probe on the skin surface modulates the amplitude of the received signals. As the skin surface is not perfectly flat, an initial increase of the pressure results in an augmentation of the pulsating component of the PPG due to an improvement of the optical interface between the probe and the skin. When the applied pressure exceeds some threshold value the amplitude of the pulsating component decreases due to squashing of blood vessels. The accelerations produced by motion induce variations in the skin-probe pressure, and may cause re-distribution of the fluids within the tissue and result in modulation of the optical-signal amplitude and waveform. The importance of this type of perturbation is directly related to the mass of the probe and to the mechanical properties of the body-attachment solution.

In order to cope with the motion-related artifacts, it is mandatory to model the effects of motion on the optical signals. Most of the actual methods postulate that motion artifacts are additives. This assumption is seductive because it allows formulating simple signal-processing techniques to reduce or to remove the artifact components from PPG signals. However, its validity is limited. The combination of multiplicative and additive models, possibly with nonlinear relations, is certainly more representative of the real relations that exist between the motion and the motion artifacts observed in the PPG signals (see Figure 6). In this context, the observed signal $y(t)$ is represented by the sum of the pulse component $s(t)$ and a weighted multiplication of both pulse and motion components $s(t)$ and $m(t)$ as described by the following equation: $y(t) = (1 + \alpha \cdots m(t)) \cdots s(t)$, and the observed frequency components Y are described by the sum of pulse components S and the weighted convolution of pulse and motion components S and M as described by the following equation: $Y = S + \alpha \cdots S * M$.

Typical use cases of an HR monitoring system allow classifying the motion artifacts into three different categories. The first category is related to rhythmical motions generally produced during endurance activities such as walking, running, and biking. Under such circumstances, motion artifacts behave as a stationary process and the PPG signals can be enhanced by signal-processing techniques. The two other categories are related to non-rhythmical motions and are categorized as intermittent or continuous. These two categories of motion artifacts are generally more difficult to cope with, especially the continuous one. The successful estimation of the HR with optical systems during activity is dependent on two main factors: the optomechanical design of the probe and the signal-processing algorithms.

2.6 Optomechanical Design

The design of the optical probe has to take into account different constraints in order to minimize the effects of the motion on the PPG signals. To reduce motion artifacts

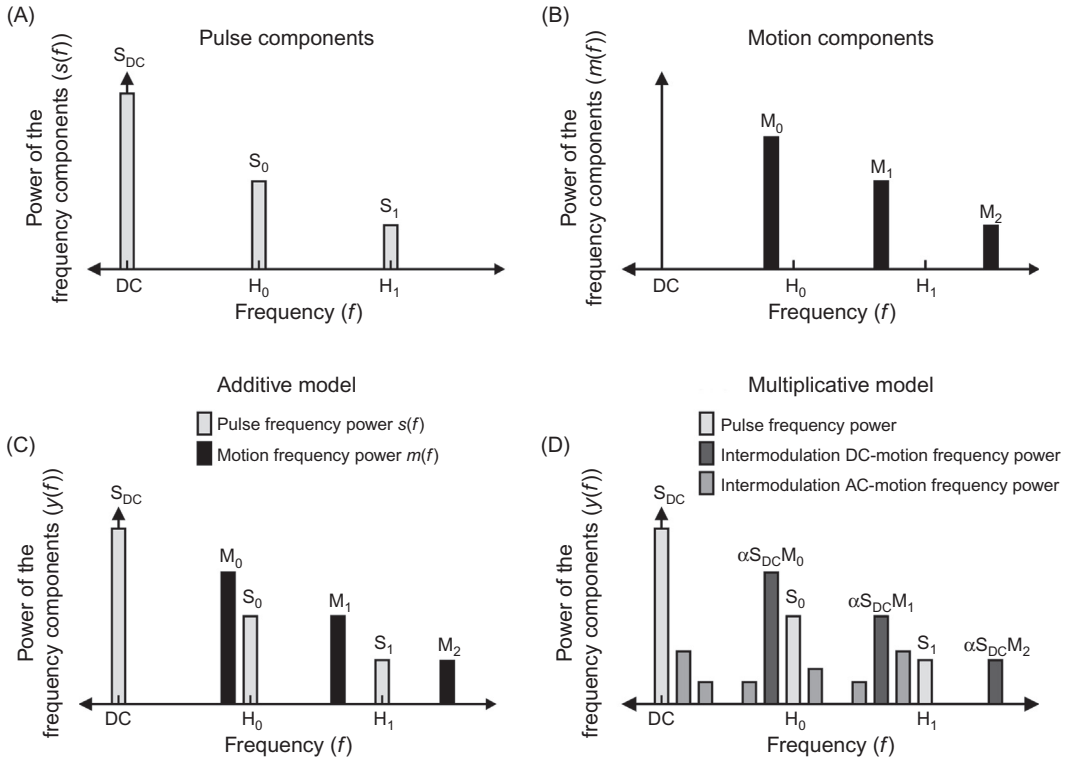

FIGURE 6 (a) Frequency components S of the optical pulsating signal $s(t)$. (b) Frequency components M of the optical motion-artifact signal $m(t)$. (c) Frequency components Y of the observed signal $y(t)$ following an additive model. (d) Frequency components Y of the observed signal $y(t)$ following a multiplicative model.

produced by the inertial forces induced by the motion (relative skin-sensor displacements and pressure variations), the mass of the measurement system has to be minimized. The probe can also be designed such that friction force takes place between the sensor and the skin surface to reduce the relative displacement. The attachment of the sensor to the body has to avoid insufficient stiffness that increases its sensitivity to motion artifacts. The attachment also has to provide an adequate pressure of the sensor on the skin surface resulting in an optimal optical interface. The probe fixation should also ensure that the applied pressure is not excessive, which could possibly result in blood-vessel clutching and discomfort. Finally, the distribution of the blood vessels in the tissues is generally non-uniform, and small displacements of the optical emitter-receiver pair can result in drastic changes in the amplitude of the pulsating component (AC component) and therefore increase sensitivity of the optical signal to motion-artifact corruption. The optomechanical design of the probe is therefore of tremendous importance concerning the sensitivity to motion artifacts.

2.7 Dedicated Signal Processing

No optomechanical design alone is able to reduce the sensitivity to motion artifacts to a suitable level: a robust estimation of the HR requires as well the implementation of a signal processing algorithm. Typically, the processing scheme of the optical signals involves three main steps, namely the enhancement, the spectral estimation, and the robust estimation of HR.

The PPG signal enhancement consists of suppressing, or at least reducing, the motion artifacts in the observed signals while preserving the pulsating component. When only one PPG signal is available without other signals of other sources, the possibility of enhancement is limited. Under such circumstances, the enhancement is restricted to the acceptation of uncorrupted segments and to the removal of motion-corrupted segments. This selection is generally obtained by the analysis of the morphological properties of the signal such as the amplitude and its stability.

In order to facilitate the removal of the motion artifacts in the PPG signal, it is possible to use extra signals that contain information on the skin and/or sensor motions. A possible solution is to obtain a motion reference signal by using an extra light emitter at a different wavelength. The wavelength has to be selected such that its sensibility to the optical attenuation of the blood is minimal (outside of the typical PPG wavelength range). This will ensure that the extra signal mostly contains motion artifacts. For optimal efficiency, the probe has to be designed such that the pulse-measurement and the motion-measurement wavelengths share the same optical path. The integration of pressure sensors in the probe is also possible. Another possibility consists of the addition of 3D accelerometers to combine optical and direct sensor-motion signals.

Different algorithmic approaches combine the optical and the motion signals to reduce the artifacts. The simplest approach consists of using the motion reference signal to discard motion-corrupted segments. More sophisticated approaches are based on the assumption that motion artifacts follow an additive model, as described before. Different approaches have been developed to identify the relation existing between the motion reference signal and the motion component present in the pulse signal. The first approach consists of estimating the spectrum of both optical and motion signals, then identifying the spectral peaks that are related to the motion present in both signals, and keeping the non-motion peaks present only in the optical signal by filtering processes. The main limitation of these approaches is that it can only be used when the motion artifacts are rhythmical.

Another approach uses an adaptive filter [36] to find the model parameters that map the motion signal on the motion components present in the optical signal. At the end, these motion components are subtracted from the optical signals [37]. This approach is more robust because it does not require rhythmical motion. Practically, it works on rhythmical and also on limited non-rhythmical motion conditions.

Finally, an efficient method that is suitable for rhythmical motion consists of the estimation of the fundamental frequency of the motion from a reference signal and the use of notch filters centered on the harmonics of this frequency to remove the artifacts in the pulse signals [38].

Different HR estimation approaches might be applied to PPG signals once the optical signals have been enhanced. These methods are divided into two categories: the ones

operating in the frequency or the ones in the time domains. Frequency domain approaches estimate the spectral density of the signal using either a non-parametric (fast Fourier, discrete cosine, or wavelet transforms) or parametric methods (autoregressive model). In order to ensure a rejection of the erroneous estimations, every measurement is associated with an index of reliability. For the frequency domain approaches, the reliability index is generally the entropy value of the spectrum (high reliability when only one dominant peak is present in the spectrum and low reliability when the spectrum contains several possible dominant frequencies).

The approaches operating in the time domain are divided into two categories that are based on the detection of events related to the heartbeat or on the tracking of the instantaneous dominant frequency. Event detection-based approaches consist of the detection of characteristic events related to heartbeats in the PPG signals, typically maxima, minima, or zero-crossings. The temporal intervals between these events expressing heartbeat intervals are used to estimate the HR. The reliability of the measurement is estimated from the dispersion of the values of these intervals. Finally, adaptive frequency tracking approaches are based on a model whose parameters are adapted to track the dominant frequency observed in the PPG signals. With the appropriate filter setting, the tracked frequency expressed averaged HR values. The reliability is estimated from the ratio of the energy present at the dominant frequency over the energy of the whole signal.

Finally, the cardiac frequency values resulting from the spectral analysis or the event temporal intervals are processed to obtain the current HR values usually expressed in bpm. Different formulations such as Bayesian estimation, reliability-dependent autoregressive estimation, outlier rejection, and a model of the dynamics of the cardiovascular system can be used to obtain the final HR estimate.

3. APPLICATIONS

3.1 Sport and Fitness

Monitoring HR during exercise is especially useful in endurance training, professional training planning, or fitness workout. Maximal oxygen uptake and energy expenditure can be accurately estimated from HR measurements [39]. HR monitoring in real time during training allows a user to control his training intensity accurately to optimize training and to avoid too low or too high training loads. Also, training effect, i.e., excess post-exercise oxygen consumption, can be accurately estimated from HR recorded during the training session. Furthermore, use of HR monitoring devices adds motivation for users to exercise [40].

The original idea for wearable HR monitoring dedicated to sports came from cross-country skiing training in the late 1970s. Professor Seppo Säynäjäkangas from the University of Oulu made a prototype of a wearable HR monitor that used a wired connection. In 1983 the first chest-strap wearable HR monitor was produced by Polar Electro. This device consisted of two parts: a watch receiver and a transmitter on an ECG-based chest strap [41]. Today, ECG-based chest straps are widely used for monitoring HR during sports, and their annual sales exceed 10 million pieces worldwide. Chest-strap-based HR monitors provide relatively accurate monitoring of HR, but they suffer from reduced

comfort from the chest strap, especially for female users. In addition, their reliability may be compromised with dry skin, dirty electrodes, or poor strap placement.

Optical monitoring of HR has been recently introduced as an alternative to overcome especially the usability and user acceptance-related challenges in chest-strap HR monitors. Optical monitoring of HR during sports may be done from different body positions, including ear [42,43], forearm [44], and wrist [20]. Most of the commercialized devices use a green light source (one or several LEDs) combined with a single photodetector. It has been shown that green light has a better signal-to-noise ratio for AC components during movement than longer wavelengths [17]. Table 1 gives the characteristics of two typical devices designed for sports based on optical HR monitoring, namely the Scosche's RHYTHM and the Mio's Alpha (Figure 7).

The reliability of the currently available devices has not been studied widely. Figure 8 shows an example of two optical HR monitors (forearm-located Scosche's RHYTHM and wrist-located Mio Alpha) as compared with reference HR (ECG-based chest strap) during walking and running on a treadmill with increasing load. The examples show that these devices may provide high-quality HR monitoring during exercise, but significant errors are also possible. Possible reasons for poor performance in some cases may include poor device attachment (sensor-skin contact), poor skin perfusion, or algorithm failures.

Figure 9 shows an example of optical HR monitoring during cross-country skiing with the same ECG-based chest strap reference and PPG-based devices (forearm-located Scosche's RHYTHM and wrist-located Mio Alpha). The device performance during this 48 minutes of cross-country skiing results in RMSE (normalized correlation) values of 13.12 bpm (0.76) for the forearm-located device and 18.07 bpm (0.47) for the wrist-located device. Both optical HR monitors fail at the beginning of the exercise due to low skin temperature, but work reliably once skin temperature and thereafter blood perfusion close to skin increased.

Current solutions are mostly based on green light with the advantage of providing robustness against motion artifacts and the disadvantage of being sensitive to poor skin perfusion. As displayed by Figure 8 and Figure 9, these solutions do not yet reach the reliability of chest-strap HR monitoring in a wide range of conditions. Cold conditions especially appear challenging for the PPG HR technique (see Figure 9). Furthermore, it is likely

TABLE 1 Technological Description and Features of Representative Devices for *ad hoc* Performance Tests

Features	Scosche's RHYTHM	Mio's Alpha
Source light color and #LEDs	2 infrared LEDs	2 green LEDs
Number of photo detectors	1	1
Location	Forearm	Wrist
Band type	Textile	Plastic rubber
Wireless connectivity	Bluetooth	Bluetooth/ANT +
Data storing	Without memory	Without memory
Display	No display	Dot − Matrix LCD

Available devices differ in terms of sensor location, exact design of the sensor element (wavelength(s) used, number of LEDs and photodetectors, LED − PD spacing), and algorithms to extract HR from motion-disturbed signal.

FIGURE 7 (a) Scosche's RHYTHM device located at the forearm with a view of the light sources (extremities) and photodetector (middle) design (dashed line). (b) Mio's Alpha device located at the wrist with a view of the light sources (extremities) and photodetector (middle) design (dashed line). Scosche uses two infrared LEDs and Mio uses two green LEDs.

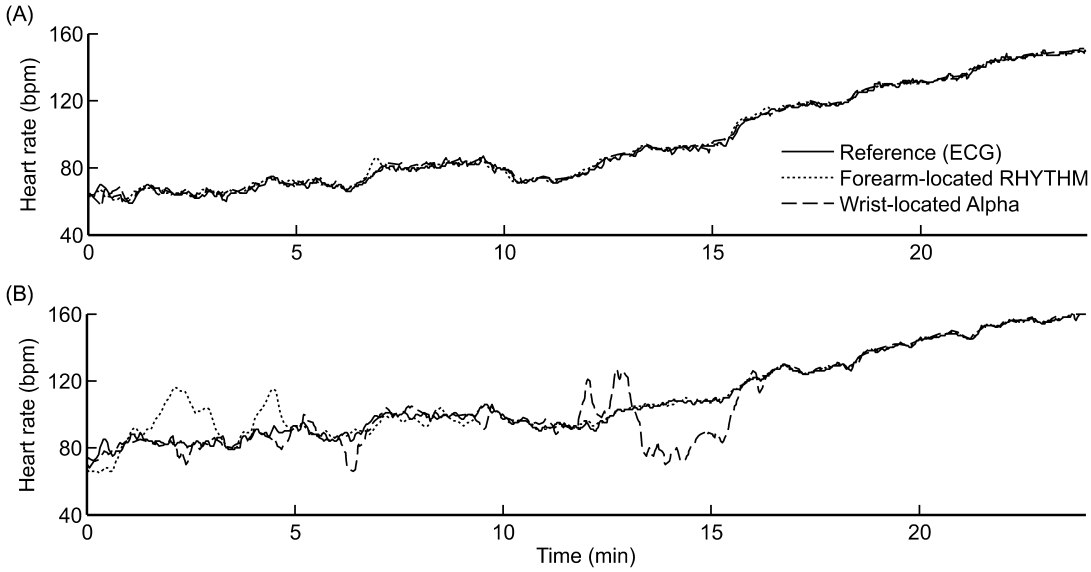

FIGURE 8 Optical HR monitoring from forearm and wrist locations during treadmill walking and running with increasing load, as compared to chest-strap ECG system on healthy volunteers. (a) Example of good quality experiment (RMSE values of 1.79 and 1.73; correlation values of 0.99 and 0.99, respectively). (b) Example of poor quality experiment (RMSE values of 7.27 and 9.60 and correlation values of 0.96 and 0.94, respectively).

that any sport that includes vigorous movements of the body part where the sensor is attached will be challenging for optical HR monitoring (e.g., racket games for wrist- or forearm-located sensors). However, if these challenges could be overcome, optical HR monitoring during sports could become an attractive alternative to a broad range of fitness consumers.

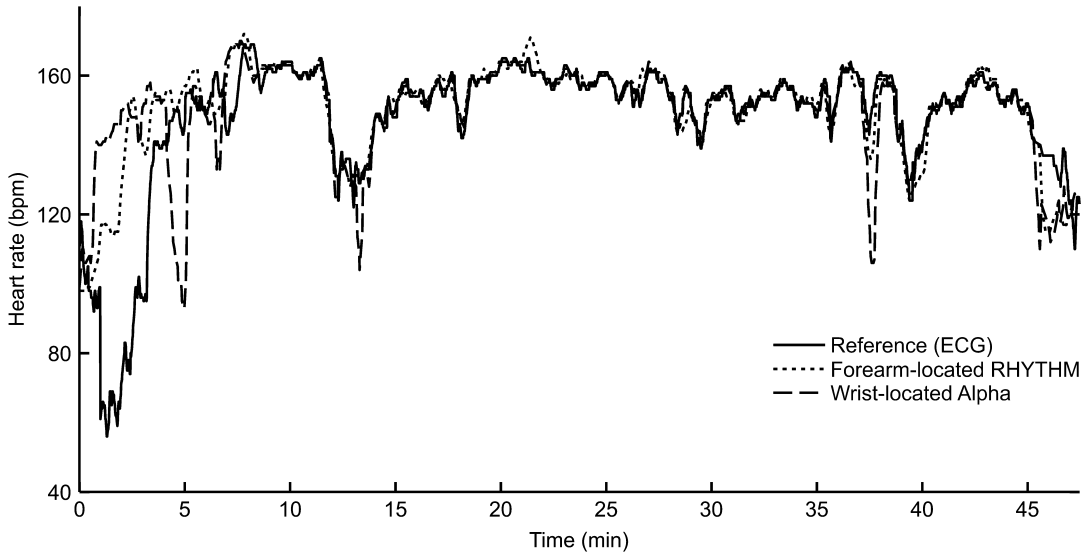

FIGURE 9 Optical HR monitoring from forearm and wrist locations during cross-country skiing as compared to chest-strap ECG system (RMSE values of 13.12 and 18.08 and correlation values of 0.76 and 0.47, respectively).

3.2 Daily Life

Consumer interest in wearable sensors beyond sports is increasing rapidly. Monitoring of movement (acceleration) during daily life allows quantifying patterns and amount of physical activity, step count, and rough estimates of energy expenditure, while monitoring during sleep allows estimation of sleep duration and to some extent sleep quality. Typical sensor solutions are accelerometry-based devices that are worn on the wrist or trunk. HR monitoring during daily life would allow more accurate estimation of physical activity and energy expenditure [45], but also physiological stress and recovery [46]. However, chest-strap or electrode-based solutions are not widely acceptable for long-term use.

Optical HR monitoring is potentially more acceptable to users due to its potentially better wearability and unobtrusiveness. However, reaching continuous reliable optical HR monitoring during daily life is challenging. While relatively reliable HR monitoring during sports may be achieved with current solutions (see previous section) these solutions may not be directly applicable to daily life. Comfortable, snug, and fit-to-the-skin sensors are essential for reliable monitoring as poor sensor contact drastically increases motion artifacts. However, long-term continuous sensor and strap contact require a solution that allows skin ventilation and does not compress the vascular bed. For example, optical HR monitoring designs targeting sport applications may not be acceptable for daily-life users. Finally, optical sensing requires significant power due to inherent power consumption of LEDs and required circuitry, and significant attention needs to be paid to solutions to extend battery life beyond current solutions to reach full 24/7 monitoring.

Today, some solutions for extending optical heart rate monitoring to daily life exist [21]. However, continuous reliable monitoring of heart rate during daily life has not yet been reached.

3.3 HRV Applications

The analysis of HR and its variability (HRV) has been the subject of numerous clinical studies concerning cardiological diseases, sleep analysis and apnea, physiologic phenomena, pharmacological responses, and risk stratification. However, a clinical consensus has been reached only in two scenarios: (1) the prediction of risk after acute myocardial infraction and (2) early detection of diabetic neuropathy. In patients following acute myocardial infraction, depressed HRV has proven to be a good predictor of mortality and arrhythmic complications, and independent of other established factors [47]. As for the assessment of diabetic autonomic neuropathy, the short- and long-term HRV analyses have proven to be accurate in its early detection [48]. Other promising studies have also investigated the potential of HRV in other cardiological diseases such as hypertension [49], congestive heart failure (insufficient pump action) [50–52], arrhythmias [53–55], and sudden death or cardiac arrest [56,57]. All of these studies are based on gold-standard HRV features. In order to extract these features, the techniques used to monitor HR have to be able to detect and record accurate heartbeat time locations (e.g., the timing of R wave peaks from ECG signals).

Although PPG monitoring devices are medically accepted as a means to assess average heart rate values, little is known about the reliability of PPG signals to extract these relevant HRV features [58,59]. However, one thing is certain, the methodology associated with the PPG post-processing is important. Successive heartbeat detection, motion artifact correction, normal beat detection, and uniformly distributed heartbeat interval processes have to be applied before extracting the desired HRV features.

Figure 10 shows an example of the correlation between heartbeats estimated from ECG and PPG signals. In this example, the heartbeat detection algorithm was obtained from a multi-channel first derivative signal in which the superior envelope was estimated. An adaptive threshold approach was applied to this superior envelope to detect heartbeat locations. In order to analyze the behavior of the autonomous nervous system, any ectopic, premature beats, or outliers were also rejected from the heartbeat time series. The entire post-processing approach is described in detail by Arberet et al. [60]. This example shows that with a high-quality PPG signal it is possible to extract HRV time series that closely resemble that extracted from ECG.

An interesting study [24] evaluated the heartbeat detection performance of a reflectance PPG sensor integrated into a wrist device in the context of sleep monitoring of subjects affected by chronic mountain sickness and sleep-disordered breathing. Figure 11 provides a Bland-Altman plot summary of the comparison between ECG- and PPG-estimated heartbeat time series from the analyzed dataset ($N = 26$ subject, $\approx 930'000$ heartbeats). The overall mean absolute error and its standard deviation ($\mu \pm \sigma$) when estimating RR intervals are 0.05 ± 17.96 ms.

Once the proper algorithm is applied to detect the normal heartbeat locations, a uniformly resampled process must be applied to the heartbeat time series before any

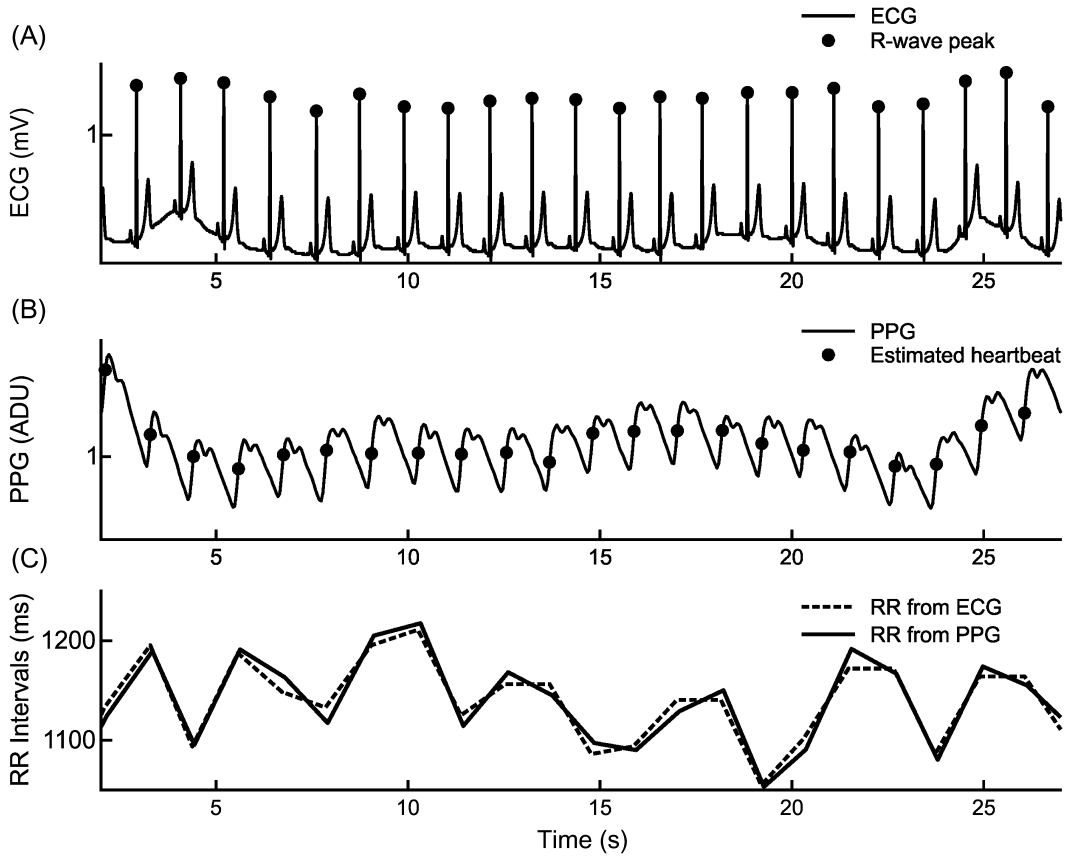

FIGURE 10 Illustration of typical R-wave peak detection (cardiac muscle contraction) observed from ECG signals (a), the corresponding heartbeats detected on PPG signals (b) and the resulting heartbeat intervals from both origins (b) [60].

time- and frequency-domain HRV feature extraction; state-of-the-art guidelines suggest resampling at 4 Hz [4]. Then, the gold-standard HRV features can be computed from the resulting uniformly resampled RR signals. These HRV features might be estimated over 5-minute or 24-hour segments, depending on the physiological behavior in observation. In the time domain, the most common variable to calculate is the standard deviation of the heartbeat intervals, labeled SDNN, which stands for standard deviation (SD) of the normal beat (NN) intervals. The standard deviation of differences between adjacent heartbeat intervals, labeled SDSD, which stands for standard deviation (SD) of the heartbeat standard deviation (SD) values, is another gold-standard feature generally used in HRV analysis. Figure 12 displays an example of the evolution of these two features estimated from ECG and PPG signals.

In the frequency domain, various non-parametric and parametric methods exist to estimate the frequency components [4], the most popular one being the fitting of an

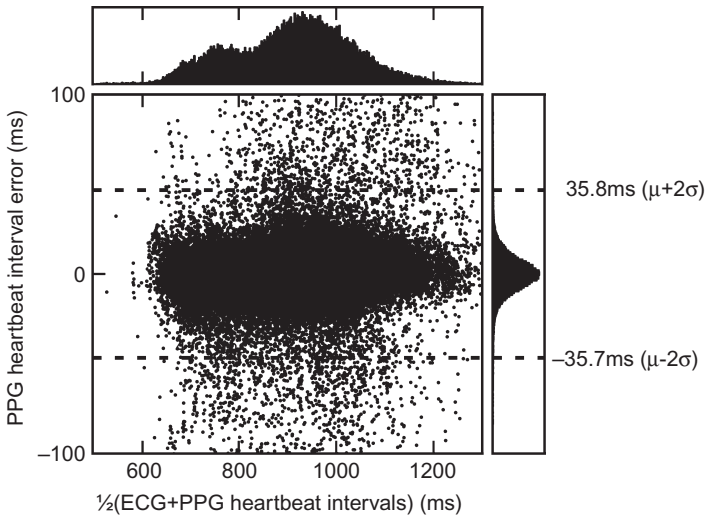

FIGURE 11 Bland-Altman plot comparing reference ECG-derived heartbeat intervals (RR intervals as measured by ECG) to associated PPG-derived heartbeat intervals (as estimated by the PPG-wrist device). The entire dataset contains a total of $N \approx 933k$ heartbeats from 26 subjects affected by chronic mountain sickness and sleep-disorder breathing. The overall error μ is 0.05 ± 17.96 ms and its standard deviation σ is 18 ± 2 ms [24].

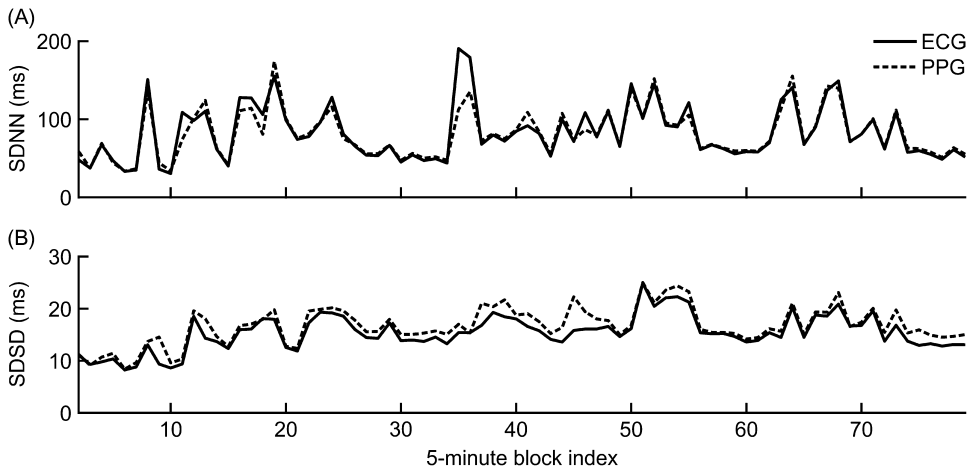

FIGURE 12 Time-domain HRV features (SDNN and SDSD) estimated from ECG and PPG signals. The correlation (normalized absolute error values) are 0.93 (0.15) and 0.95 (0.11), respectively [60].

auto-regressive model of a given order (usually 12) to the heartbeat interval time series. A spectral analysis is then applied to estimate the powers located in the very low, low and high frequencies (0.003-0.04, 0.04−0.15, 0.15−0.4 Hz, respectively). The ratio between low- and high-frequency powers is usually added to provide information on the relationship between sympathetic (low) and parasympathetic (high) nerve activities. Figure 13 displays an example of the evolution of the described four frequency-domain HRV features estimated from ECG and PPG signals. One can observe that the general behavior of the HRV features estimated from PPG signals is well correlated to the ones estimated from ECG signals.

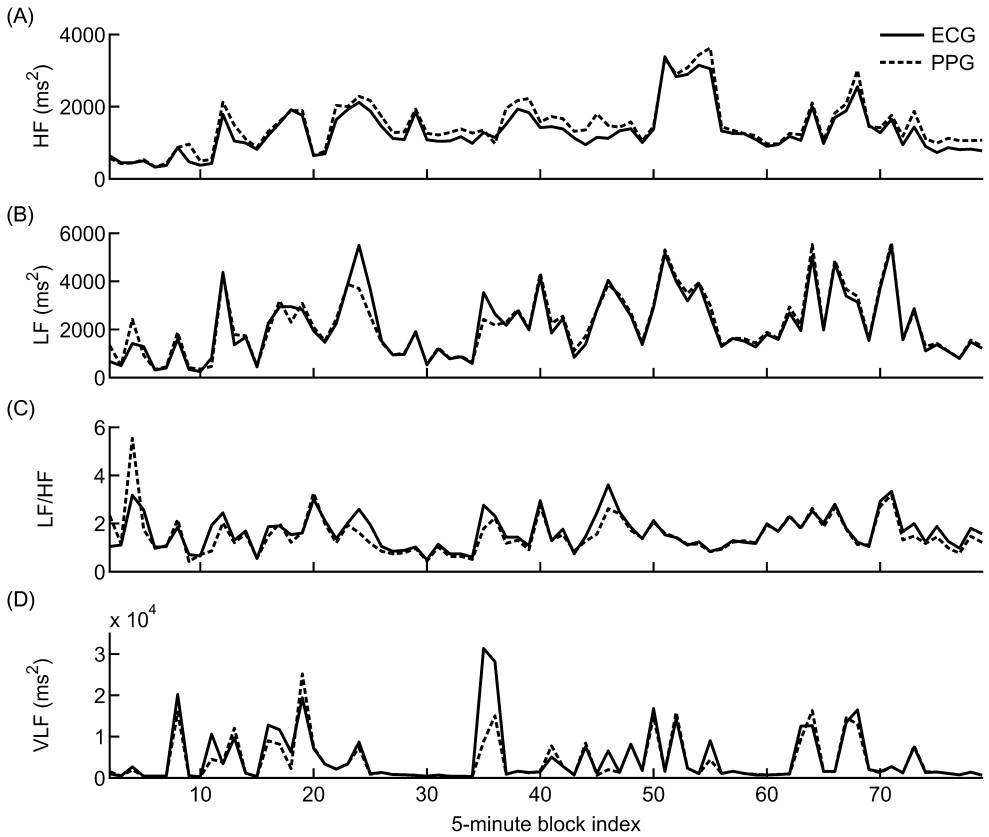

FIGURE 13 Frequency-domain HRV features (HF, LF, LF/HF, and VLF) estimated from ECG and PPG signals. The correlation (normalized absolute error) values are 0.98 (0.16), 0.96 (0.14), 0.84 (0.25), and 0.86 (0.43), respectively [60].

Clinical applications of PPG-based HR monitoring devices are at the embryonic stage. Most of the HR-based clinical studies based on Holter and other wearable HR monitoring devices could be reproduced using a less cumbersome, inconspicuous, and more daily-life long-term appropriate PPG-wrist device. The advantage of such devices is also related to the accuracy of the diagnostic. Cumbersome systems affect the behavior of the subjects, especially during night monitoring. Therefore, a light and well-integrated PPG device might in the future provide novel insights into long-term cardiovascular regulation mechanisms.

4. CONCLUSION AND OUTLOOK

This chapter aimed at providing an overview of the current developments in the field of optical HR monitoring. Starting from the basics of the PPG technology, this chapter

highlighted the importance of fully understanding the PPG opto-electrical phenomenon. Because of the complex interactions between mechanical sensor parts, optical properties of the sensors, and particularities of living tissues, the design of optical HR monitors should be driven by a thorough analysis of the specific PPG configuration to be implemented. Different trade-offs are to be expected, for instance, when targeting a long-term wellness monitor to be used during sleep, than when targeting an HR monitor for short-term use during a marathon or sport. The optical-electrical-mechanical design of any PPG-based sensor is thus a critical issue that might compromise the usability of the HR monitor.

A second important aspect covered by this chapter is the design of a signal-processing strategy that allows deriving the desired health indexes from the raw PPG signals. The take-home-message of this chapter is that there is no universal algorithm to be implemented within an optical HR monitor, and deep understanding of the underlying physiological problem will guide the developer toward the optimal algorithmic configuration.

Finally, this chapter covered the most recent advances of optical HR monitoring in two different fields: from the monitoring of athletes toward hospitalized patients. The goal of the provided material was to demonstrate that, supported by an optimized design, the PPG technology can cover a very large spectrum of applications going from very precise HRV analysis of sleep to HR monitoring during cross-country skiing.

What is preparing the future? Imagination. The technological floor is now set in the optical heart rate monitoring domain: the principal elements of the PPG puzzle are gradually becoming reality and how to combine them in order to meet particular demands is currently the main challenge. The explosion of smartphones and smartwatches offers an excellent perspective to these techniques: optical monitoring is non-obtrusive and comfortable, and can be easily integrated in such devices. Smart textiles, integrating functional garments and patches, are other promising technologies where optical HR could play an important role. And last but not least, contactless sensors, using a modified PPG-technology, could monitor computer users via webcams.

In conclusion, optical HR monitors have the potential to become one of the central technologies to support the development of the twenty-first century's health and well-being assessment revolution. The technology is there; the implementation is in your field.

NOMENCLATURE

AC Alternating current
BPM Beats per minute
DC Direct current
ECG Electrocardiogram
H_2O Water
Hb Hemoglobin
HbO_2 Oxyhemoglobin
HR Heart rate
HRV Heart rate variability
ICG Impedance cardiography
LED light-emitting diode
ms Millisecond
O_2 Oxygen

PCG Phonocardiography
PPG Photoplethysmography
RMSE Root-mean-square error

Acknowledgments

This chapter was made possible by grants from the Wilsdorf Foundation and the Swiss National Science Foundation. The authors would like to thanks the collaborators of CHUV (Claudio Sartori *et al.*) responsible for data acquisition. The authors would also like to thank all CSEM's collaborators who were involved in the design and development of the wrist-located PPG device.

References

[1] P. Brodal, The Central Nervous System: Structure and Function, third ed, Oxford University Press, Inc, New York, 2004.
[2] A.E. Aubert, B. Seps, F. Beckers, Heart rate variability in athletes, Sports Med. 33 (2003) 889–919.
[3] M. Malik, J. Bigger, A. Camm, R. Kleiger, A. Malliani, A. Moss, et al., Heart rate variability: Standards of measurement, physiological interpretation, and clinical use, Eur. Heart J. 17 (1996) 354–381.
[4] M.H. Crawford, S.J. Bernstein, P.C. Deedwania, J.P. DiMarco, K.J. Ferrick, A. Garson, et al., ACC/AHA guidelines for ambulatory electrocardiography: executive summary and recommendations: a report of the American College of Cardiology/American Heart Association task force on practice guidelines, Circulation 100 (1999) 886–893.
[5] G.-X. Yan, R.S. Lankipalli, J.F. Burke, S. Musco, P.R. Kowey, Ventricular repolarization components on the electrocardiogram, J. Am. Coll. Cardiol. 42 (2003) 401–409.
[6] J. Sztajzel, Heart rate variability: a noninvasive electrocardiographic method to measure the autonomic nervous system, Swiss. Med. wkly. 134 (2004) 514–522.
[7] M.H. Bonnet, D.L. Arand, Heart rate variability: sleep stage, time of night, and arousal influences, Electroencephalogr. Clin. Neurophysiol. 102 (1997) 390–396.
[8] <http://www.polar.com>, (Accessed Last: 23.06.14).
[9] <http://www.suunto.com>, (Accessed Last: 23.06.14).
[10] <http://www.garmin.com>, (Accessed Last: 23.06.14).
[11] J.G. Webster, The measurement, instrumentation and sensors6 handbook, first ed., CRC Press, Inc., Boca Raton, 1999.
[12] Timex Group USA Inc. (2013). Heart Rate Monitor | Global Timex. [Online]. Available: <http://global.timex.com/collections/heart-rate-monitor>. (Last Accessed: 07.07.14).
[13] Salutron Inc. (2013). Smart Health-Salutron. [Online]. Available: <http://salutron.com/smart-health>. (Last Accessed: 07.07.14).
[14] M.J.E. Parry, J. McFetridge-Durdle, Ambulatory Impedance Cardiography, Nurs. Res. 55 (2006) 283–291.
[15] J.G. Webster, Design of Pulse Oximeters, first ed., IOP Publishing Ltd, London, 1997.
[16] Nonin Medical Inc. (2013). Nonin - Pulse Oximeter (Sp02) Monitoring Solutions. [Online]. Available: <http://www.nonin.com/pulseoximetry>. (Last Accessed: 07.07.14).
[17] Masimo Corporation. (2013). Masimo - close to the heart. [Online]. Available: <http://www.masimo.com>. (Last Accessed: 07.07.14).
[18] Covidien. (2013). Nellcor Pulse Oximetry Monitoring from Covidien. [Online]. Available: <http://www.covidien.com>. (Last Accessed: 07.07.14).
[19] J. Allen, Photoplethysmography and its application in clinical physiological measurement, Physiol. Meas. 28 (2007) R1.
[20] Physical Enterprises Inc. (2013). MIO Global. [Online]. Available: <http://www.mioglobal.com>. (Last Accessed: 07.07.14).
[21] BASIS Science Inc. (2013). Basis — health and heart rate monitor for wellness and fitness. [Online]. Available: <http://www.mybasis.com>. (Last Accessed: 07.07.14).

[22] Impact Sports Technologies. (2013). ePulse2. [Online]. Available:<http://www.impactsports.com/epulseii.html>. (Last Accessed: 07.07.14).
[23] J.W. Severinghaus, Takuo Aoyagi: discovery of pulse oximetry, Anesth. Analg. 105 (2007) S1–S4.
[24] P. Renevey, J. Sola, P. Theurillat, M. Bertschi, J. Krauss, A. Daniela, et al., Validation of a wrist monitor for accurate estimation of RR intervals during sleep, Conf. Proc. IEEE Eng. Med. Biolol. Soc. (2013) 5493–5496.
[25] Penaz, J. (1973). Photoelectric measurement of blood pressure, volume and flow in the finger. In proceedings Digest of the 10th international conference on medical and biological engineering, 104.
[26] J. Sola, M. Proenca, D. Ferrario, J.-A. Porchet, A. Falhi, O. Grossenbacher, et al., Non-invasive and non-occlusive blood pressure estimation via a chest sensor, IEEE Trans. Biomed. Eng. (2013).
[27] Y. Maeda, M. Sekine, T. Tamura, A. Moriya, T. Suzuki, K. Kameyama, Comparison of reflected green light and infrared photoplethysmography, Conf. Proc. IEEE Eng. Med. Biol. Soc. (2008) 2270–2272.
[28] G. Natalini, A. Rosano, M.E. Franceschetti, P. Facchetti, A. Bernardini, Variations in arterial blood pressure and photoplethysmography during mechanical ventilation, Anesth. Analg. 103 (2006) 1182–1188.
[29] L. Nilsson, A. Johansson, S. Kalman, Respiration can be monitored by photoplethysmography with high sensitivity and specificity regardless of anaesthesia and ventilatory mode, Acta. Anaesthesiol. Scand. 49 (2005) 1157–1162.
[30] B. Jönsson, C. Laurent, T. Skau, L.-G. Lindberg, A new probe for ankle systolic pressure measurement using photoplethysmography (PPG), Ann. Biomed. Eng. 33 (2005) 232–239.
[31] K.H. Shelley, D. Tamai, D. Jablonka, M. Gesquiere, R.G. Stout, D.G. Silverman, The effect of venous pulsation on the forehead pulse oximeter wave form as a possible source of error in Spo2 calculation, Anesth. Analg. 100 (2005) 743–747.
[32] Y. Maeda, M. Sekine, T. Tamura, Relationship between measurement site and motion artifacts in wearable reflected photoplethysmography, J. Med. Syst. 35 (2011) 969–976.
[33] Y. Mendelson, B.D. Ochs, Noninvasive pulse oximetry utilizing skin reflectance: noninvasive pulse oximetry utilizing skin reflectance, IEEE Trans. Biomed. Eng. 35 (1988) 798–805.
[34] Huang, F., Yuan, P., Lin, K., Chang, H. and Tsai, C. (2011). Analysis of reflectance photoplethysmograph sensors. in proceedings World Academy of Science, Engineering and Technology (59) 1266–1269.
[35] J.L. Reuss, Arterial pulsatility and the modeling of reflectance pulse oximetry, Conf. Proc. IEEE Eng. Med. Biol. Soc. 3 (2003) 1901–1904.
[36] S.S. Haykin, Adaptive Filter Theory, forth ed., Prentice Hall, New Jersey, 2001.
[37] Renevey, P., Vetter, R., Krauss, J., Celka, P., and Depeursinge, Y. (2001). Wrist-located pulse detection using IR signals, activity and nonlinear artifact cancellation. In proceedings 23rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society (3), 3030–3033.
[38] B. Lee, Y. Kee, J. Han, W.J. Yi, Adaptive comb filtering for motion artifact reduction from PPG with a structure of adaptive lattice IIR notch filter, Conf. Proc. IEEE Eng. Med. Biol. Soc. (2011) 7937–7940.
[39] J. Achten, A.E. Jeukendrup, Heart rate monitoring: applications and limitations, Sports Med. 33 (2003) 517–538.
[40] A. Ahtinen, J. Mantyjarvi, J. Hakkila, Using heart rate monitors for personal wellness--the user experience perspective, Conf. Proc. IEEE Eng. Med. Biol. Soc. (2008) 1591–1597.
[41] R.M.T. Laukkanen, P.K. Virtanen, Heart rate monitors: state of the art, J. Sports Sci. 16 (1998) 3–7.
[42] El-Khoury, M., Sola, J., Neuman, V. and Krauss, J. (2007). Portable SpO2 monitor: a fast response approach. in proceedings IEEE International Conference on Portable Information Devices, 1–5.
[43] Valencell Inc. (2013). PerformTek®. [Online]. Available: <http://www.valencell.com>. (Last Accessed: 07.07.14).
[44] Scosche Industries. (2013). RHYTHM - The Best Heart Rate Monitor for iPhone. [Online]. Available: <http://www.scosche.com/rhythm>. (Last Accessed: 07.07.14).
[45] P.G. Montgomery, D.J. Green, N. Etxebarria, D.B. Pyne, P.U. Saunders, C.L. Minahan, Validation of heart rate monitor-based predictions of oxygen uptake and energy expenditure, J. Strength Cond. Res. 23 (2009) 1489–1495.
[46] A. Uusitalo, T. Mets, K. Martinmäki, S. Mauno, U. Kinnunen, H. Rusko, Heart rate variability related to effort at work, Appl. Ergon. 42 (2011) 830–838.
[47] R.E. Kleiger, J.P. Miller, J.T. Bigger, A.J. Moss, Decreased heart rate variability and its association with increased mortality after acute myocardial infarction, Am. J. Cardiol. 59 (1987) 256–262.

[48] F. Bellavere, I. Balzani, G. De Masi, M. Carraro, P. Carenza, C. Cobelli, et al., Power spectral analysis of heart-rate variations improves assessment of diabetic cardiac autonomic neuropathy, Diabetes 41 (1992) 633–640.
[49] S. Guzzetti, S. Dassi, M. Pecis, R. Casati, A.M. Masu, P. Longoni, et al., Altered pattern of circadian neural control of heart period in mild hypertension, J. Hypertens. 9 (1991) 831–838.
[50] G. Casolo, E. Balli, T. Taddei, J. Amuhasi, C. Gori, Decreased spontaneous heart rate variability in congestive heart failure, Am. J. Cardiol. 64 (1989) 1162–1167.
[51] P.F. Binkley, G.J. Haas, R.C. Starling, E. Nunziata, P.A. Hatton, C.V. Leier, et al., Sustained augmentation of parasympathetic tone with angiotensin-converting enzyme inhibition in patients with congestive heart failure, Am. J. Cardiol. 21 (1993) 655–661.
[52] J.N. Townend, J.N. West, M.K. Davies, W.A. Littler, Effect of quinapril on blood pressure and heart rate in congestive heart failure, Am. J. Cardiol. 69 (1992) 1587–1590.
[53] H.V. Huikuri, J.O. Valkama, K.E. Airaksinen, T. Seppänen, K.M. Kessler, J.T. Takkunen, et al., Frequency domain measures of heart rate variability before the onset of nonsustained and sustained ventricular tachycardia in patients with coronary artery disease, Circulation 87 (1993) 1220–1228.
[54] M.G. Tsipouras, D.I. Fotiadis, D. Sideris, An arrhythmia classification system based on the RR-interval signal, Artif. Intell. Med. 33 (2005) 237–250.
[55] N. Al-Rawahi, M.S. Green, Diagnosis of supraventricular tachycardia, J. Assoc. Physicians India 55 (2007) 21–24.
[56] C.M. Dougherty, R.L. Burr, Comparison of heart rate variability in survivors and nonsurvivors of sudden cardiac arrest, Am. J. Cardiol. 70 (1992) 441–448.
[57] A. Algra, J.G. Tijssen, J.R. Roelandt, J. Pool, J. Lubsen, "Heart rate variability from 24-hour electrocardiography and the 2-year risk for sudden death,", Circulation vol. 88 (1993) 180–185.
[58] K.H. Shelley, Photoplethysmography: beyond the calculation of arterial oxygen saturation and heart rate, Anesth. Analg. 105 (2007) S31–S36.
[59] P. Dehkordi, A. Garde, W. Karlen, D. Wensley, J.M. Ansermino, G.A. Dumont, Pulse rate variability compared with heart rate variability in children with and without sleep disordered breathing, Conf. Proc. IEEE Eng. Med. Biol. Soc. (2013) 6563–6566.
[60] Arberet, S., Lemay, M., Renevey, P., Sola, J., Grossenbacher, O., Andries, D., Sartori, C., Bertschi M, (2013). Photoplethysmography-based ambulatory heartbeat monitoring embedded into a dedicated bracelet. In proceedings 40th Conference on Computers in Cardiology, (in press).