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A review of methods for non-invasive heart rate measurement on wrist

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Abstract

When evaluating general health condition on a patient, heart rate is an essential indicator as it is directly representative of the cardiac system state. Continuous measurement methods of heart rate are required for ambulatory monitoring involved in preliminary diagnostic indicators of cardiac diseases or stroke. The growing number of recent developments in wearable devices is reflective of the increasing demand in wrist-worn activity trackers for fitness and health applications. Indeed, the wrist represents a convenient location in terms of form factor and acceptability for patients. While most commercially-available devices are based on optical methods for heart rate measurement, others methods were also developed, based on various physiological phenomena. This review is focused on heart rate measurement methods located on forearm and more specifically on the wrist. For each method, the physiological mechanism involved is described, and the associated transducers for bio-signal acquisition as well as practical developments and prototypes are presented. Methods are discussed on their advantages, limitations and their suitability for an ambulatory use. More specifically, the superposition of motion artefacts over the signal of interest is one of the largest drawbacks for these methods, when used out of laboratory conditions. As such, artefact reduction techniques proposed in the literature are also presented and discussed.

Keywords: heart rate, measurement, methods, wrist, forearm, plethysmography, ballistocardiography, electrocardiography

1. Introduction

With the increasing of life expectancy and development of centenarians lifestyles, cardiovascular chronic diseases are representing a major issue for healthcare systems, both in terms of cost and infrastructures occupancy. It is now widely admitted that personal health (pHealth) and telemedicine devices are efficient solutions to assist medical practitioners with ageing populations. Ideally, these solutions should maintain the autonomy level and independence of elders while monitoring populations at risk. Preceding factors have led to an increasing demand in ambulatory measurement tools to assess patients health condition during daily life activities. Continuous monitoring of heart rate (HR) on a patient enables evaluation of parameters such as heart rate variability (HRV). As these physiological parameters are controlled by autonomous nervous system and influenced by both internal (stress, relaxation, sleep)

and external factors (physical activity) they provide rich information about physical and mental status of an individual [1]. In association with the actimetry assessment, HR and HRV provide general health indicator and have tremendous diagnostic value for cardiovascular diseases.

Having a non-invasive method capable of precisely monitoring heart rate is highly desirable for continuous patient monitoring. Nevertheless, with continuous monitoring comes continuous wearing of a measurement device, and, consequently, the design process should be driven by user acceptance. If such devices cause patient disturbance, they will no longer be worn and accordingly, no measurement at all will be possible. The wrist is a convenient location to wear a monitoring device as it represents an ideal measurement site in terms of patient acceptability [2]. However, physiological manifestations that occur on the wrist have weak amplitudes, requiring transducers capable of working with very low signal-to-noise ratio, thus having high sensitivity. Moreover, during motion episodes, this high sensitivity will also acts as a major drawback by picking-up motion component in addition with physiological information.

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Indeed, in ambulatory conditions, wrist is highly vulnerable to motion artefacts. An ideal measurement method should be a compromise between acceptability, accuracy and motion artefact robustness. Commercial wrist worn devices extensively rely on optical methods for assessing heart rate. However, these wearable devices cannot be used to produce clinical grade physiological indicators in ambulatory conditions. Whereas the literature provides a wide variety of measurement techniques, this paper intends to reference and evaluate them in terms of performance, sensibility and acceptability.

This paper is organised as follow: in section 2, heart beating physiological process is exposed. A classification based on the physical quantities affected by cardiovascular activity is proposed. Then, according to the proposed classification, transduction mechanisms involved in heart rate estimation are discussed in sections 3, 4 and 5. As previously mentioned, wrist is a preferential measurement site, consequently referenced methods and prior art are focused on wrist-worn devices. In section 6 methods are compared on their accuracy and usability in ambulatory conditions, motion artefact reduction techniques and algorithms are classified and discussed on their advantages and limitations. Section 7 offers an overview of reviewed methods.

2. Heart rate measurement methods classification

As heart beating is a complex physiological process which involves the variation of multiple physical quantities within the body, heart rate measurement methods can be classified in function of the phenomenon used for the acquisition.

Electrical-based methods. Electrical-based methods rely on the electrical activity of the cardiac muscle. During each cycle, electrical *action potentials* coming from the transitory depolarisation of the cardiac cells membrane can be acquired from the chest using conventional electrocardiographic (ECG) recorders. These complex electrical waveforms shown on Fig. 1 contain a significant amount of information on the functional state of the cardiovascular system. At the beginning of an auricular systole, a first wave (P) appears on the ECG related to the auricular depolarisation. Then follows the QRS complex corresponding to the depolarisation of ventricles which is responsible for ventricular contraction and for the increase of ventricular and aortic pressure. Finally, the T wave represents the beginning of ventricular re-polarisation when muscular fibers start relaxing.

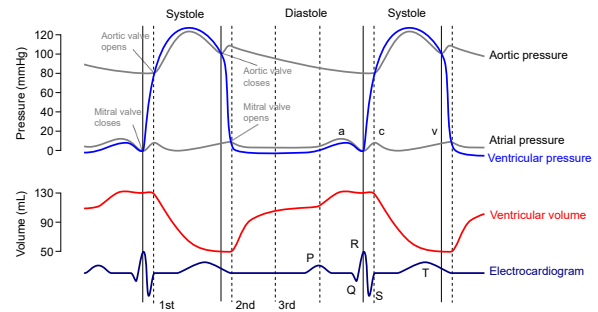


Fig. 1. Wiggers diagram : pressure, volume and ECG records. [3]

Plethysmography. Plethysmographic methods are based on the local modification of blood volume occurring during systole and diastole. On each cardiac cycle, during systolic phase, blood is ejected through the aortic valve in the circulatory system. For an average person, ventricular volume goes from 130 ml down to 50 ml. This causes a global increase of blood pressure and volume in systemic circuit: elastic arteries, muscular arteries and arterioles are mainly affected by this pressure and volume change. On the contrary, pressure in capillaries and venous blood flow remain constant regardless of systolic or diastolic phase, making them unsuitable for plethysmographic-based measurements.

Ballistocardiography. A pulse wave occurs each time blood is ejected in circulatory system, propagating from arterial tree to peripheral blood vessels. Ballistocardiographic methods are mechanical-based methods which take advantage from the deformation caused by this pulse wave. These methods are used in specific locations such as radial artery on wrist or carotid, where the pulsation is the most perceptible.

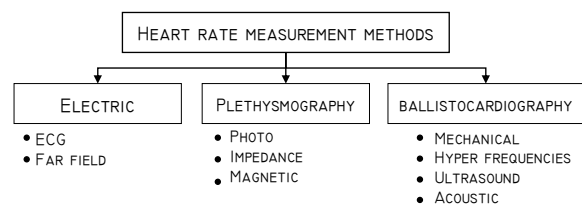


Fig. 2. Proposed classification for heart rate measurement methods

In this paper we consider techniques which are suitable for measuring the heart rate from the wrist using non invasive techniques. A proposed classification is shown on Fig. 2. For each one of the three physiological modalities previously presented, we have identified, in the literature, practical methods used for acquisition

of the corresponding variation of physiological quantities. Prior art using these methods at wrist location is referenced and discussed.

3. Plethysmography

The word "plethysmography" comes from two ancient greek terms: "plethysmos", which is the idea of an increasing quantity, and "graphos" for writing or recording. In modern medicine literature, it commonly designates the recording of volume changes in a live body. Historically, the first referenced attempt for recording these volume changes has been made by Hertzman [4] in 1937 using an optical approach. Since then, others methods have been developed for evaluating volume modification caused by blood flow.

3.1. Photoplethysmography

This technique is widely used in clinical devices, personal activity trackers, fitness heart rate monitors and other commercially available devices for heart rate measurement.

3.1.1. Measurement principle

In its simplest form, a photoplethysmographic device is comprised of a light source and a photodetector. Local volume variations of the radial artery (in case of a wrist-worn device) are used to modulate light intensity due to partial absorption or reflection, resulting in variations of current within the photodetector.

Volume variations and photodetector responses are linked by the Beer-Lambert law (Eq. 1). By illuminating tissues with an absorption coefficient α with a light source of intensity I_0 , and by measuring the fraction of received light I_t , thickness variations d can be evaluated [5].

$$I_t = I_0 \exp(-\alpha d) \quad (1)$$

During systole, blood volume increases, which consequently increases the absorption coefficient, thus decreasing the amount of light received by the photodetector. Conversely, during diastole, blood volume decreases. The absorption coefficient is then minimal and the transmitted light through tissues becomes maximal. According to this phenomenon, photoplethysmographic devices are able to acquire dynamic variations of blood volume in arteries. As systolic and diastolic phases are synchronised on heart rate, it has been established that the signal amplitude obtained from the photosensitive element is directly modulated by heart rate [6].

This variable component of the PPG signal is designated as AC, or Pulsatile Cadiac Component (PCC). A DC, or Low Frequency Component (LFC), is also present and is induced by the static absorption of skin and tissues due to their optical properties. Additionally, a slow baseline drift, synchronised on the respiratory rate, might occur due to the variation in venous blood return during each breathing cycle. Figure 3 presents a graphic representation of the tissues and medium responsible for the PPG signal shape.

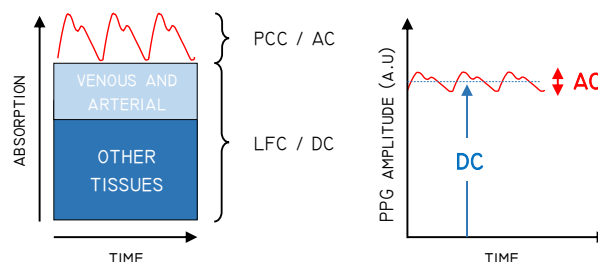


Fig. 3. PPG absorption and signal shape

However, even if this phenomenology is commonly admitted for red and infrared wavelength (historically used in PPG measurements), it does not fully explain the origin of PPG signal when other wavelengths, as for example green light (525 nm), are used. As it will be discussed later, green light does not penetrate tissues down to arteries and consequently should not contain a pulsatile component. Kamshilin and Margaryants [7] suggested that the pulsatile nature of green-recorded PPG signal comes from variations in the density of capillaries. When blood volume variation occurs on the considered measurement site, the diameter of arteries increases, resulting in a compression of the capillary bed located between arteries and skin. Consequently, absorption and scattering coefficients at such a wavelength would also reflect heart rate.

In addition to heart rate information, PPG signals also contain information on the hemodynamic functional state and can be used for diagnosing malfunctions in cardiac function. For example, waveform steepness during systolic phases is an indicator of the force of ventricular contraction. Secondly, the position of dirotic notch (secondary peak observed in the decreasing part of the pulse wave, see Fig. 3) offers an information on vasomotor tone [8]. Additionally, respiratory rate can be acquired using PPG : slow amplitude variations synchronised on respiratory rate are superimposed onto cardiac component [9]. Pressure modifications occurring in chest cavity on each breath cycle

causes fluctuation in peripheral venous blood circulation [10]. Evaluation of this respiratory component can be performed using Fast Fourier Transform (FFT) as proposed by Wang et al. [11], this method shown high degree of correlation (>0.97) with medical monitoring device. Finally, photoplethysmography is also used for the evaluation of pulsed oxygen saturation (SpO_2). For this purpose, a second light source is required; the first one should have a preferential emission domain in red (660 nm), while the other one within the infrared domain (940 nm). These wavelengths correspond to the maximal absorption of oxyhemoglobin (HbO₂) and deoxyhemoglobin (HHb). By assessing absorption coefficients corresponding to each wavelength, using a wide-band photodetector, the pulse oxygen saturation (SpO_2) can therefore be evaluated, using an empirical model [12].

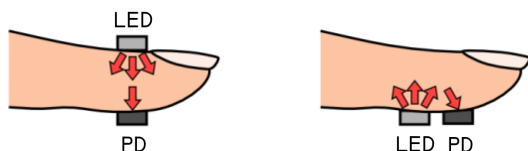


Fig. 4. Transmission and reflection PPG working principle. [13]

As a function of the relative position between the photoemitter and receptor, photoplethysmographic (PPG) acquisition systems can be qualified either of reflection PPG or transmission PPG (see Fig. 4). When using reflection photoplethysmography, the light source and the sensitive element are placed on the same side (for example above the radial artery). This configuration is used when tissues introduce a significant absorption coefficient in the light path, typically in case of wrist-worn devices. In transmission PPG, the light source and the detector are placed across the measurement side, on the same axis. An important part of light is then scattered and absorbed by tissues, depending on the thickness of the tissues crossed. However, a fraction of blood-modulated light is still transmitted to the photodetector.

3.1.2. Parameters

Anatomical site. Photoplethysmographic signal quality mainly relies on the measurement location. By using a multiple-wavelength PPG device, signal quality has been evaluated on different locations along the arm and the hand by Maeda et al. [14]. Measurements were conducted on 11 subjects, and upon 7 different locations on the left arm simultaneously using green and infrared wavelengths. When using green light (525 nm), lateral

upper arm has shown the highest SNR; others sites such as wrist and forearm are also suitable for these measurements but it should be kept in mind that measurements located on peripherals such as finger or wrist are more exposed to motion artefacts due to the variable optical contact quality. In addition with optical path modifications, the acceleration during motions causes a change of blood flow in the vessels, as peripheral sites such as fingers or hands are more vascularized.

Wavelength. The wavelength used for PPG measurements is also a predominant parameter regarding the signal quality as well as the device's sensitivity to motion artefacts. In particular, green wavelength has shown significant advantages in comparison to conventional red/infrared based PPG devices [15]. As the penetration depth of light in human tissues is wavelength dependant (the shorter the wavelength is, the thinner the penetration is [16]), the use of different wavelengths enables to target variations of light scattering and absorption in the different biological layers at the measurement site. More specifically, blood volume variations occur at the depth where arteries are located, while motion artefacts are produced on superficial skin layers, as they are mainly due to disturbances of the optical contact. Thus, multiple wavelength measurements can be used for motion artefact detection and reduction [17]. As wavelength can influence Signal-to-Noise-Ratio (SNR) and sensitivity to motion artefacts, it will also cause inaccuracies on the estimated heart rate. Measurements performed with green light (525 nm) and blue light (470 nm) shown the best agreement in terms of earth rate with synchronised ECG ambulatory measurements [18]. In a similar approach, by using a multi-spectral PPG device during physical activities, Blanos et al. [19] demonstrated that green light features highest agreement with simultaneous ECG recording (with a bias of 0.1261 BPM and standard deviation of 1.244 BPM). Using multiple wavelength laser sources and associated wide-band spectrometer, Spigulis et al. [16] evaluated differences in PPG waveform shape. It has been found that as wavelength increases, components such as dirotic notch become less pronounced or even invisible in the infrared domain (1064 nm). Similar experiments conducted by Asare et al. [20] brought out temporal differences in systole's maximum peak and rising time, in addition with differences in shape and amplitude, as function of the wavelength. This time-shift occurring between PPG waveforms acquired at different wavelengths has been used by Liu et al. [21] for continuously monitoring blood pressure using Pulse Transit Time (PTT) method. As penetration depth is wave-

length dependant, the contribution from capillary and arterial blood can be differentiated. Consequently, arteriolar pulse transit time is then used for blood pressure estimation [17]. Finally, wavelength must be chosen carefully as it influences sensor response as function of skin pigmentation or dyes used in tattoos [22]. Performance of yellow-orange PPG system was evaluated over different skins tone by [23], in terms of Perfusion Index (PI) which represents the ratio of AC to DC components. This PI indicator was evaluated on a population whose skin pigmentation is representative of Fitzpatrick scale. It has been demonstrated that yellow-orange wavelength has better immunity in regard to skin tone variation, higher PI was measured over 95 % of individuals.

Temperature. Optical properties of tissues also depend on body temperature. When temperature decreases, circulation in capillaries is reduced. This vasoconstriction phenomenon reduces AC component amplitude in PPG, especially when short wavelengths are used as their penetration depth is lower [15].

Acquisition. Methods used for signal conditioning is an other parameter for signal quality improvement. Ambient light acts as the main interference source along with motion artefacts. As ambient light is an external noise source, a synchronous detection is beneficial for removing day light and fluorescent bulb lights from desired PPG recording [24]. In this configuration, illuminating LEDs are driven by a reference signal which is later used for demodulation. An other approach for external noise reduction is to measure ambient light contribution while LEDs are not driven. This measured contribution is later subtracted from PPG samples acquired while LEDs are powered on [25]. This technique is commonly used in commercial PPG Analog Front Ends (AFE). In case of integration in a wearable device, PPG could represent a high power consumption source as it requires illumination of LEDs for measurement. A strategy to limit this consumption is the adaptive intensity control of LEDs: depending on perfusion quality and ambient conditions, LEDs power can be adjusted [26].

Contact force. Attachment of optical elements onto the skin influences AC component of PPG signals. Experiments conducted by Teng and Zhang [27] suggested that adding a small amount of pressure in the range of 0.2 to 0.4 N upon the measurement site increases AC cardiac amplitude, thus enhancing SNR. In a walking situation, when motion artefacts are more likely to appear, a constrained PPG device shown better agreement

with simultaneous ECG recording than a device without any pressure applied [28]. Similar results were found by Dresler [29], where an optimal contact force will increase PPG signal amplitude. However, beyond this optimal point, vessel constricting occurs, resulting in a reduced signal amplitude. Sensor housing and attachment will also affect signal amplitude and immunity to motion artefacts. For example, external light barriers required for minimising cross-talk between photoemitters and receivers can also act as a mechanical damper. Optical elements manufacturers, such as OSRAM usually provides technical guidelines for integration and sensor response optimisation [30].

Sensor layout. Arrangements between photoemitter and detector also influence PPG signal amplitude and thus SNR. In order to minimise effects due to poor sensor placement or tilt, redundancy in light sources is required to maintain optimal illumination onto the photodetector. For example, placing multiple LEDs around the photodiode in a "T" configuration as suggested by Mohapatra et al. [31] will ensure optimal illumination. This symmetrical configuration around the photodiode is widely used in commercially available PPG devices. However, an optical isolation has to be used between elements as light radiated by LEDs should not illuminate the photodiode directly. Ideally, only the contribution of reflected light from skin comprising the pulsatile component should reach the photodiode. Consequently, the distance between emitter and receiver is also critical [32]. The distance between optical elements will influence the shape of PPG signal in terms of Perfusion Index (PI). In addition with separation distance, the PI depends on skin coloration, ambient exposure and various physiological parameters such as temperature. Typically, PI varies from 0.02% for very small pulsatile waveform and up to 2% in ideal conditions. Using a dedicated optical simulation software, Kao et al. [33] evaluated the optimal distance between optical elements. From these simulations, it has been determined that for infrared light (850 nm - 950 nm), the optimal PI is reached for a distance of 2 mm between optical elements. However, another parameter has to be considered: if the elements are placed too close from each others, the detector response will be saturated by direct illumination and penetration in tissues will be insufficient to obtain a significant AC component [34]. On the other hand, while increasing the distance produces better perfusion index, higher current injection level in emitters will be required to maintain the amount of light received by detector [35]. Using a micro-translation stage to adjust distance from the source to the detec-

tor, Huang et al. [36] evaluated the ideal distance for multiple wavelength. In this study, perfusion index and signal-to-noise ratio were considered as relevant indicators for determining the optimal distance. For green wavelength (543 nm - 571 nm), the ideal separation is located between 2 and 3.2 mm. Consequently, the ideal distance is a compromise between the PI (thus, penetration depth and path length in tissues) and the amount of power required to maintain this perfusion index (while minimising optical cross-talk that comes with higher injection levels).

3.2. Impedance-plethysmography

In addition with PPG, this acquisition method is also used for heart rate measurements on commercial devices such as the UP3 (Jawbone, San Francisco, California).

3.2.1. Measurement principle

Volume variations can also be acquired using impedance plethysmography (IPG). The measurement principle is illustrated on Fig. 5: an alternative current source is used to inject current in measurement site through the outer electrodes. Inner electrodes are then used to measure a resulting voltage signal proportional to the instantaneous impedance of tissues. As previously described, the systolic phase causes a volume modification in artery, resulting in a modification of the average impedance under the skin. During systolic phase, blood flow increases, resulting in a lower impedance. Conversely, during diastole, a higher impedance is measured [37].

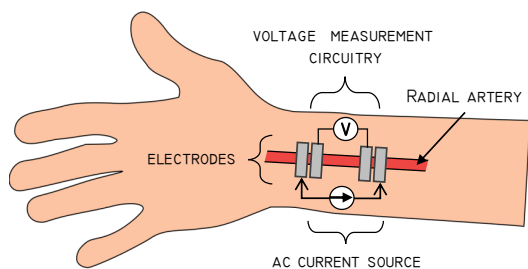


Fig. 5. Impedance plethysmography measurement principle

As long as this impedance measurement is performed at a sufficient sampling rate, the heart rate component will be visible on signals [38]. In this configuration, envelope of the acquired signal is comprised of a pulsatile component $Z(t)$ representative of cardiac activity, and a static component Z_0 introduced by the average impedance of tissues (see Fig. 6). However, the acquired waveform has to be demodulated to retrieve the

impedance variations. One of the main challenge for IPG is thus to assess the very low impedance changes of few milliohms within a signal amplitude in the order of several decades of ohms [39].

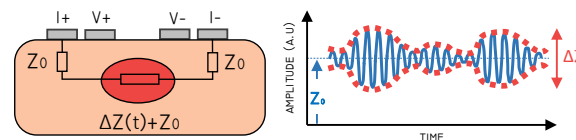


Fig. 6. IPG waveform and simplified model

A commonly used measurement setup is composed of four dry electrodes, as tetrapolar (four-wire) method is used to reduce the contribution of skin-electrode interface. The contact impedance introduced by electrodes represents an unknown parameter as it will change depending on individuals [40], as well as over time. Indeed, sweating is also acting as a natural electrolyte and will cause a long-time drift in case of prolonged measurement [41].

3.2.2. Parameters

Acquisition technique. Standard IPG measurements require a minimal electronic circuitry. Jonghwa Lee and SeongHwan Cho [42] used a simplistic architecture comprised of an instrumentation amplifier followed by a passive diode envelope detector for amplitude demodulation. This demodulated voltage is directly representative of an impedance variation and is digitized for further signal processing, including heart rate estimation, using a custom algorithm. To overcome variability issues inherent to this principle of measurement (such as contact quality drifting over time), an enhanced circuitry build around the impedance measurement front-end AD5933 (Analog Devices, Norwood, Massachusetts) has been developed by Schneider et al. [43]. In this approach, the injected current level and the input voltage gain are adjusted to maintain an optimal SNR. For enhanced noise reduction, Xu et al. [44] designed a wrist-worn device with a demodulation circuit made using synchronous detection. A reference voltage derived from the injected current is used in a mixer for down-converting the resulting voltage across the measurement electrodes. This architecture features better accuracy and noise immunity (such as power line interferences) than direct demodulation. However, motion artefacts can not be addressed using this technique as they are directly causing impedance variation. An other architecture developed by Rossi et al. [45] used a digital version of synchronous detection which exploits the

time-frequency domain duality. Instead of using continuous time and analog approach, it uses a sample and hold circuit driven by a phase shifted version of the excitation signal injected in the electrodes. This analog front end design features low consumption level (down to 750 μA), in comparison with typical 10 mA for the AD5933 [46]. Programmable System-On-Chips (PSoCs) have also been used by He et al. [37] in a wrist-worn device. Both analog and digital functions required for heart rate measurement using IPG were implemented in a single chip solution. In this case, waveform demodulation is made using a digital lock-in demodulation algorithm so it does not require any external hardware.

Electrodes. As they represent the interface between user and measurement device, electrodes are a fundamental parameter. For a continuous use and integration in a wrist-worn device, wet electrodes such as disposable silver - silver chloride electrode are not suitable, as the hydrogel will degrade over time [47]. Dry electrodes are mainly used for long term monitoring in IPG, however, this type of electrode is more sensitive to motion artefacts : contrarily to wet electrodes, they do not use hydrogel which normally acts as a mechanical damper [40]. The influence of the measurement site was evaluated by Cho et al. [48]. They developed a bio-impedance measurement system for pulse wave velocity (PWV) and heart rate monitoring on the wrist. Twelve positions were evaluated along ulnar and radial artery using 0.75 cm² dry aluminium electrodes with 100 kHz excitation current. In this configuration, the highest variations of impedance were found to be along ulnar artery, approximately 2 cm away from the wrist. In addition with sensitivity, electrodes placement also influences contribution of motion artefacts. Using finite elements simulation on wrist section, Jivet [49] suggested that transversal electrodes placement could reduce their influence. Electrode size and their relative position between current injection and voltage measurement influence current distribution [50] and thus, sensitivity. Indeed, the distance between electrode pairs affects the targeted tissues within and under the skin: the closer the two pairs are, the higher the sensitivity to vessels near surface is [51].

Excitation frequency. Commonly used frequencies in impedance plethysmography are in the range of 10 kHz to 100 kHz associated to a current injection level lower than a milliamperere. According to Anderson [51], in this frequency range, blood has a higher conductivity than other tissues such as muscle and fat. However, blood's conductivity itself varies with frequency [52, 53] but it

does not significantly affect acquired waveform appearance or amplitude [54].

3.3. Magneto-plethysmography

Several approaches were explored to acquire the cardiac activity based on variations of a magnetic field on the wrist. Magneto-cardiography (MCG), as first proposed by Baule and McFee [55] in 1963, uses the magnetic field produced by cardiac activity for vital signal recording and pathology diagnosis. However this requires complex and high-end equipment that need to be run at liquid helium temperature. Another approach used in medical imaging for haemodynamic response evaluation is based on the magnetic properties of haemoglobin [56]. As it will be discussed in the following section, new methods suggest the use of off-the-shelve magnetic sensors for pulse wave acquisition and cardiac frequency measurement.

3.3.1. Measurement principle

In medical imaging, by using functional Magnetic Resonance Imaging (fMRI), brain activation areas can be identified using haemoglobin magnetic properties. When a specific brain area is stimulated, the corresponding neuronal activity causes an increasing demand in oxygen [57]. The oxygen requirement highers the local blood flow in the concerned area, but the supply of oxygenated haemoglobin (HbO_2) exceeds the oxygen required by neuronal activation. This local and instantaneous oxygenated haemoglobin excess can then be detected by a fMRI device, as HbO_2 is diamagnetic and de-oxygenated haemoglobin (Hb) is paramagnetic, so that HbO_2 excess results in a modification of the magnetic susceptibility in tissues. This oxygen level modification in haemoglobin resulting from neuronal activity is known as Blood Oxygen Level Dependant (BOLD) signal. Measuring BOLD with fMRI, therefore, requires specialised imaging devices with magnetic field in the order of 3 Tesla.

In a wearable-compatible version, Phua et al. [58] suggests that haemoglobin magnetic properties can be used to evaluate cardiac frequency from wrist, even without the requirement of specialised imaging device, which are usually reserved for research or clinical purpose, such as Superconducting QUantum Interference Device (SQUID) used in magnetocardiography. As shown on Fig. 7, this approach uses a permanent magnet to create a magnetic field in the range of 0.1 to 0.2 Tesla. Close to an artery, a magnetic sensitive sensor then records perturbations caused by the circulation of blood [59]. This magnetic field modulation effect

is qualified of *Modulated Magnetic Signature of Blood* (MMSB) by Phua et al. [60].

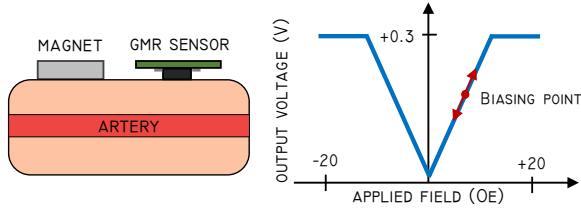


Fig. 7. MMSB measurement setup and sensor typical response

Transducers typically used for MMSB acquisition are Giant Magneto-Resistance (GMR). These sensors are based on a magnetoresistive effect discovered by Baibich et al. [61] in 1988, which occurs in thin metallic films consisting of magnetic layers with a thickness of few nanometers, separated by thin non-magnetic layers. Applying a magnetic field on GMR results in a large resistance decrease [62]. In commercially available sensors from NVE Corporation (Eden Prairie, Minnesota, USA), sensors are placed in a bridge configuration, as shown on Fig. 8. Once biased by a voltage source (or current source for higher temperature stability), the sensor produces a differential output voltage proportional to magnetic field applied according to theoretical characteristics shown on Fig. 7. By adjusting the relative distance D_0 between magnet and GMR sensor, the biasing point can be set in a linear region to obtain a maximal dynamic excursion and sensitivity. An instrumentation amplifier is then used to amplify the differential voltage and a low pass filter, acting both as anti-aliasing and noise filter, is applied prior to digitizer.

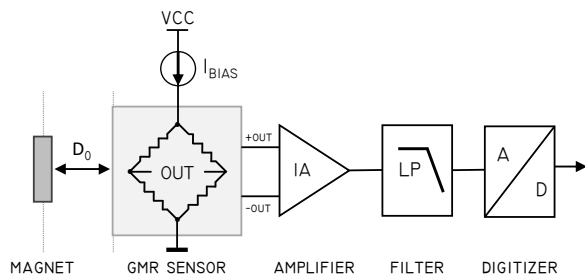


Fig. 8. GMR sensor for MMSB acquisition typical circuit

However, the origin of the previously described MMSB signal is not fully understood and remains a controversial topic discussed in literature. Simulations and experimentations conducted by Sinatra [63] suggested that blood magnetic contribution is too weak for being acquired using GMR in a non-shielded environ-

ment. They also stand that disturbances caused by magnetic properties of blood are in the order of 10^{-5} G which is much less than earth's magnetic field (≈ 0.3 G). Increasing magnetic field contribution of blood is possible by using larger magnet, however it requires the sensor to maintain a high sensitivity at high field levels. In another approach, Zhang et al. [64] and Li et al. [65] consider that mechanical effects caused by pulse wave should also be responsible for the signal acquired by GMR sensor. With this hypothesis, MMSB acquired as described by Phua et al. [58] could not be longer classified as a volume-change based method (plethysmographic) as it relies on the mechanical deformation caused by pulse wave, resulting in distance modification between magnet and GMR sensor.

3.3.2. Parameters

Transducer used. Most MMSB devices developed for heart rate measurement from wrist uses single giant magnetoresistance as transducer [58, 66, 67, 68]. However, when used in a dual configuration it can be used for Pulse Wave Velocity (PWV) measurement and blood pressure evaluation using PTT method [69, 70]. Hall sensors in a dual configuration have also been used by P M et al. [71].

Magnetic field source. Instead of using permanent magnet as a biasing source for GMR, Chandrasekhar et al. [69] used an electromagnet made from a U-shaped magnetic core and an associated excitation coil. Although this technique has a higher power consumption, it enables dynamic adjustments of the magnetic field level for an optimal biasing of GMR sensor [72], and can also be used for synchronous detection by feeding an AC signal into the coil.

3.4. Temperature

The ability of recording respiratory rate using high resolution temperature measurement has already been demonstrated [73]. The hypothesis formulated in this work is that the fresh air breathed in during inhalation is causing slight temperature reduction onto arterial blood.

As proposed by Cuadras and Casas [74], heart rate can also be acquired from the wrist using negative temperature coefficient thermistors (NTC). The principle of measurement is shown on Fig. 9: sensors are placed in a differential configuration, TH1 being placed over the artery while an auxiliary sensor TH2 is being used as a reference for skin temperature compensation, thus offering a higher dynamic range. With the associated electronic circuit, temperature variations of 0.5 m°C synchronised with heart rate were recorded. Fast Fourier

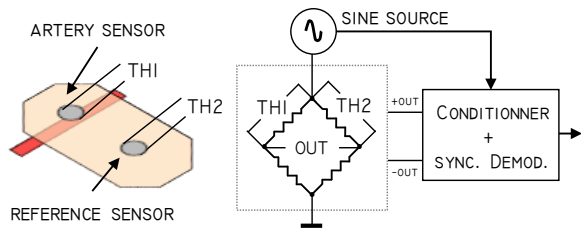


Fig. 9. Temperature measurement for heart rate evaluation

Transform (FFT) analysis on temperature records and simultaneous PPG have shown identical fundamental frequencies corresponding to the heart rate.

4. Ballistocardiography

Ballistometric methods are based on the acquisition of perceptible mechanical pulse waves from the skin's surface. During each diastolic phase, blood is ejected from the aorta and is sent through the circulatory system. This ejection causes a local increase of pressure and therefore gives rise to a pulse wave, travelling through arteries and causing perceptible mechanical deformations. Using mechanical transducers, this pulse wave can be acquired, and therefore, the heart rate can be estimated. The first recording of arterial pulse wave was done in 1854 by Karl von Vierordt using a mechanical device called sphygmograph, however this device was only able to record crude sine-shaped waveform from the wrist. Few years later in 1858, a french physiologist Étienne-Jules Marey developed an enhanced version of the sphygmograph, enabling the acquisition of a detailed pulse wave [75]. These techniques are especially suitable for wrist-based measurements as the heart rate is usually evaluated by palpating artery at *guan* position.

4.1. Mechanical

Methods presented in this section are based on mechanical transducers that require a direct skin contact for operation. The generic working principle is described on Fig. 10. Whatever the transducer, the main challenge is to develop a sensor which features a high sensitivity with a form factor that does not reduce acceptability nor causes disagreement in case of long-term monitoring. In addition, a strong motion artefacts immunity is required for an ambulatory use as mechanical-based methods usually suffer from high artefacts sensitivity. As mechanical manifestation of physiological phenomenon is only perceptible at precise locations, SNR quickly

decays if transducer is not precisely positioned. When motion occurs, transducer can slightly shift away from pulsatile location, resulting in poor physiological parameter estimation. Previously exposed methods such as PPG or IPG are less concerned by this phenomenon as physiological manifestation, volume variations, occurs in much more large areas. As it will be discussed later, multiplexing transducers to automatically select most pulsatile location is a commonly used method to overcome these limitations.

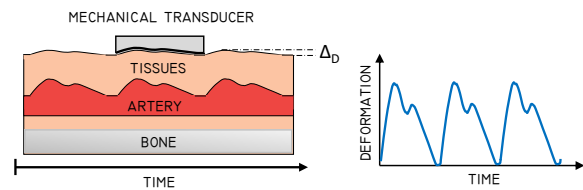


Fig. 10. Pulse wave mechanical acquisition principle

4.1.1. Transducers

In the following section, types of sensors used for mechanical deformation acquisition and their measurement principle are exposed.

Piezoelectric. By using off-the-shelf PVDF piezoelectric films and associated conditioning electronics, Almeida et al. [76] have demonstrated that it was possible to acquire the pulse wave from sites such as radial artery. Measurements conducted by Foo and Lim [77], using the same type of transducer and with a simultaneous heart rate recording using an ECG holter, have shown a good agreement between both measurements (standard deviation less than +/- 1 bpm in resting conditions). However, in ambulatory situations, vital sign sensors will experience large disturbances caused by motion artefacts. To minimise SNR degradation, Buxi et al. [78] proposed a new approach by stacking two piezoelectric sensors. The first one, close to skin, is used to derive heart rate, while the second acts as a reference sensor for the evaluation of the contribution of motion artefacts, so that signal processing methods such as adaptive filtering can be applied to reduce the contribution of motion artefacts from the first sensor.

Piezoresistive. Shu et al. [79] developed a flexible high sensitivity piezoresistive sensor using a silicon rubber nanocomposite film as functional material. The sensing element is then mounted on an elastic belt for pre-pressure bias. With associated conditioning and processing electronics, a wrist-worn prototype directly displays heart rate.

Capacitive. Capacitive sensors can also be used for pulse wave acquisition, as shown by Jovanov [80]. The capacitive TSI (*Touch Sense Input*) (NXP, Eindhoven, Netherlands) of a microcontroller is used in order to capture capacitive variations with an order of magnitude between 0.1 pF to 0.3 pF. Tissues capacitance variations are picked up by co-planar electrodes directly built onto PCB. However, this technique is highly sensitive to motion artefacts, and for instance is not suitable for ambulatory use yet. Commercially-available capacitive MEMS have also been used as a mechanical transducer by Kaisti et al. [81] in a wrist-wearable sensor array. The system benefits from the use of an array of sensors by selecting the sensor featuring the highest SNR for heart rate evaluation.

Pressure-sensitive transistor. Associated with a MOSFET for charge amplification, a PZT film can be used for higher sensitivity than piezoelectric transducers alone. Dagdeviren et al. [82] developed a wrist-worn biocompatible wearable film. Each pulse produces a mechanical constraint on the PZT array, and the charges generated are transferred to the Gate of a MOSFET transistor. When V_{GS} and V_{DS} are properly biased using voltage sources, current on the drain-source channel is representative of the mechanical strain applied onto the PZT array and thus, of the pulse wave. In a similar approach, Schwartz et al. [83] used microstructured PDMS as gate dielectric in a field effect transistor. The capacitance changes resulting from applied pressure causes a drain-source current variation representative of pulse wave.

Magnetic. Using custom designed magnetoelastic sensor comprised of two coils made of magnetic amorphous wire, Hlenschi et al. [84] were able to record deformation at wrist skin surface. Each time blood pulse causes a deformation of the flexible sensor, this results into a modification of the permeability and thus of the coupling between the two coils. The excitation coil is fed with a sine wave, and the variations of waveform amplitude picked up by secondary coil are representative of coupling modifications and is then demodulated using a simple envelope detector. An other approach proposed by Lee et al. [85] uses an array of 4×8 Hall sensors in association with a matrix of small permanent magnets for measuring deformations caused by the pulse wave through the radial artery and surrounding tissues. Sensors and magnets are separated by a conformable air chamber which transmits the deformation from the skin to an array of permanent magnets. Hall sensors are then used to create a three dimensional

surface mapping resulting from the deformation. This method reduces the need for precise placement of sensors as the most pulsatile location is directly identified.

4.2. Microwaves

Unlike mechanical transducers, hyperfrequencies-based methods can be used for detecting pulse wave without direct skin contact. This principle has been already considered in non-contact physiological parameters evaluation, by measuring motions of the chest [86, 87]. The working principle, applied to the wrist, is illustrated on Fig. 11. One or more resonator elements are placed at a fixed distance from the skin. Pulse wave, when travelling through arteries, causes near field modifications and thus the reflected signal measured is modulated by artery motion [88].

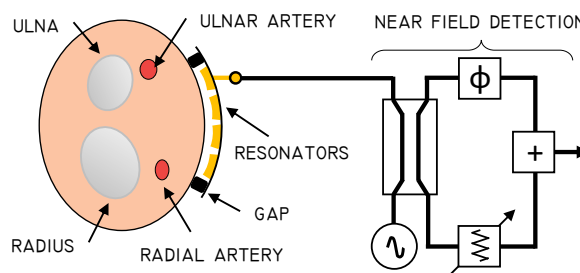


Fig. 11. Microwave acquisition form factor and principle

Using a directional coupler, the transmitted signal is fed to the variable attenuator and the reflected wave to the phase shifter. The transmitted signal is attenuated in order to have the same magnitude as the reflected signal collected from the resonators. A variable phase shifter is used to create a version of this reflected signal in phase opposition with the transmitted one, and both signals are then fed to a power combiner. It results in a signal which is only representative of pulsatile cardiac component detected from wrist [88].

However, this approach requires numerous external components which are not always easily integrable in a wearable device. The architecture proposed by Kim et al. [89] relies on an injection-locked, Phase-Locked Loop (PLL) for near field detection. As PLLs are commonly available in RF front-ends (as for example for Bluetooth or WiFi), it does not require specific external components, and features a higher degree of interoperability with existing communication systems. PLL is then directly used for measuring frequency variations resulting from cardiac activity. In a similar approach, Wang et al. [90] proposed the use of Self Injection Locked (SIL) radar technology for measuring heart rate

on wrist. This method relies on the Doppler effect created by pulse wave, resulting frequency shift is directly measured using a frequency discriminator [91].

The device developed by An et al. [92] uses four resonators and associated detection electronic on a small form factor of 30×30 mm. The only additional instruments used are a 2.4 GHz signal generator and a DAQ for digitising the output of the power combiner. Switching between resonators enables the selection of the most pulsatile wrist location and can therefore be used for better evaluation of heart rate.

4.3. Echocardiography

Echocardiographic methods are based on the use of ultrasounds to assess arteries motions caused by cardiac activity. To the best of our knowledge, there is no academic literature except from a single patent [93] using this method on a wrist worn device for assessing heart rate.

4.4. Acoustic

Acoustic methods are also suitable to assess the mechanical pulses from the radial artery. The recording of heart sounds from the chest cavity is a common technique to evaluate the correct sequencing of cardiac cycle. Typical phonocardiograms records are comprised of two main sounds qualified of S1 and S2. S1 is representative of the activity from mitral and tricuspid valves while S2 represents the activity from aortic and pulmonary valves. Consequently, the heart rate can be derived from the phonocardiogram by measuring the successive time intervals.

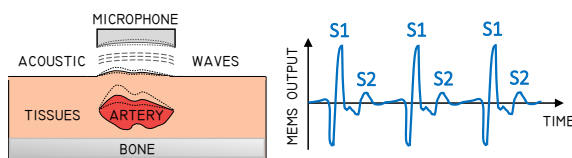


Fig. 12. Acoustic acquisition working principle

Recently, Sharma et al. [94] developed a wrist-worn device capable to sense the acoustic waves caused by the dilatation and constriction that occurs in the radial artery. The developed device embeds an omnidirectional MEMS microphone whose diaphragm is affected by the periodic waves correlated with the cardiac activity. Working principle is represented on Fig. 12. In the proposed work, the heart rate is estimated using a peak-finding method along with an artefact identification and reduction strategy.

5. Electric

These methods are based upon perceptible electrical activity of heart, commonly recorded from chest using clinical grade 12-lead ECG recorder or wearable-compatible version such as cardiac holter or patch [95]. However, this approach might not be suitable for all applications, especially for patients with reduced mobility or if periodic replacement of electrodes is required. As the measurement site considered for this review is the wrist, we will focus on methods enabling long-term ECG recording from this location.

5.1. Dual wrist ECG

Unipolar measurement of the ECG from a single location would not be possible because of its differential nature, as well as the common mode of this signal which is floating and can vary largely. To obtain the highest SNR, two electrodes are usually placed from each side of the heart where the cardiac action potentials take place. Unfortunately the signal is rapidly attenuated when electrodes are moved away from the heart, and especially if both electrodes are placed on the same side of the heart, like for example along a single arm. However, if measured from both hands, the amplitude remains high enough to extract heart rate conveniently. As such, a first approach consists in placing a device on the wrist which has to be touched by the other hand, thus ensuring the electrical contact with the skin of both arms (Fig. 13). Two electrodes (ECG_N and VCM) are placed underneath the device, and a top-sided electrode (ECG_P) allows placement of a finger from the opposite hand. As potentials measured using this approach can reach amplitudes in the range of 1 mVpp [96], it can be acquired using an off-the-shelf ECG Analog Front End (AFE) as used in the device developed by Thomas et al. [97]. This method is widely used in wrist worn heart rate monitoring devices [98, 99].

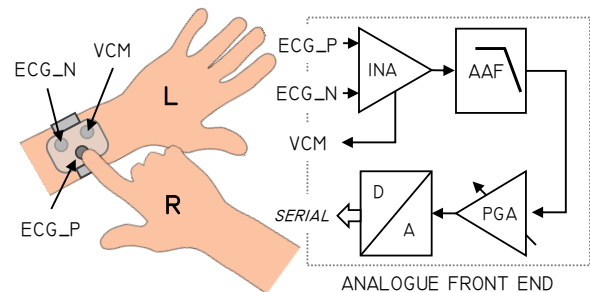


Fig. 13. Wrist-based ECG measurements and simplified AFE

As this technique requires the user to place his finger onto the monitoring device for measurement, it is not

really suitable for a continuous ECG recording. It can however be used for punctual monitoring in case of a cardiac event. To overcome this limitation, electrodes can be continuously worn on both wrist and connected by wire to a single wrist-worn monitoring device [100, 101, 96] this additional constraint degrades drastically the acceptability of such device by patients.

5.2. Far-field ECG

To overcome the limitations described previously, an approach called *far-field ECG* enables recording of ECG from a single arm, without the need for an electrode onto the opposed side. Even if this represents a seductive alternative to dual wrist ECG, measuring from distant sites makes signal detection really difficult as electrical activity of heart is highly attenuated and others bioelectrical interferences such as electromyographic noise will be superimposed [102]. Conventional measurement setup used for far-field ECG is represented on Fig. 14 with electrodes on various measurement sites.

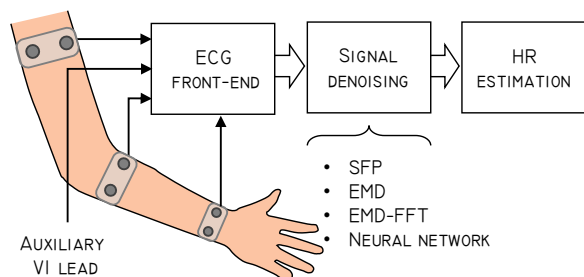


Fig. 14. Wrist-based ECG measurements and simplified AFE

In order to measure ECG signals from a single arm, Lynn et al. [103] investigated electrical potential levels from different locations, from wrist to upper arm using wet electrodes. It has been found that even if the signal is strongly attenuated, an averaging process enables to keep the QRS inflexion point visible. Signal Averaging (SA) and Single Fiducial Point (SFP) offers a basic signal extraction mode, and the denoising process is driven by simultaneously recording a reference ECG (such as precordial lead V_1). This additional source gives a time-domain information on zero-crossing point after R-wave detection, and is then used afterwards to align signals recorded from arm leads [104]. Others methods such as Empirical Mode Decomposition (EMD) do not require this additional information source and have been used by Escalona et al. [105] for denoising signals recorded from the upper arm. More recently, Vizcaya

et al. [106] proposed an approach using an artificial neural network for estimating conventional chest lead ECG waveforms from those recorded on forearm. To the best of our knowledge, there is no literature reporting a device capable of recording ECG from a single wrist directly, without using electrodes on the upper arm to obtain a minimal signal amplitude for the ECG signal.

6. Discussion

6.1. Accuracy and integration

Direct comparison between methods on their ability to accurately evaluate heart rate is hardly relevant as it strongly depends on the experimental setup, like for example the algorithms used for heart rate evaluation, or the resting state of user, or the method used for motion artefact reduction. In a steady situation with laboratory conditions, all the proposed methods in the literature have shown excellent heart rate estimations. However, as soon as the user starts a physical activity, the MA superimposed lead to erroneous estimations. Instead of discussing methods on their intrinsic precision, they will be discussed on their suitability for ambulatory use and their integration in a long term monitoring device (Table 1).

6.2. Limitations and motion artefacts

Regardless of the physiological phenomenon measured, and independently from the acquisition method, a common limitation of the presented methods is their ability of being used in an ambulatory situation were motion artefacts (MA) are more likely to appear. External noise sources such as the ambient light (in PPG), common-mode interferences (in ECG) or other external noise sources can be addressed using methods such as synchronous detection or linear filtering. However, motion artefacts are directly affecting the physical quantity measured. Motion artefacts are caused by the physical activity of the users. Various activities encountered in ambulatory conditions may affect the response from wrist-worn physiological sensors. The causes of motion artefacts can therefore be classified by their physical origin as following [35]:

- **Relative motion between sensor and skin:** In case of intense physical activity and wide amplitude forearm motions, the measurement device will experience acceleration levels that will result in device displacement. Attachment and fixture of the measurement device via strap, wristband or repositionable adhesive is an important part of the

Table 1: Comparison of heart rate measurement methods

	Method	Advantages	Drawbacks & limitations
Plethysmographic	PPG	Most used method on wearable devices, consequently, variety of analog front-ends available. Wide range of motion artefact reduction techniques in literature. Suitable for continuous monitoring. Easy to integrate in wearable devices.	Relative high power consumption. Motion artefacts and ambient exposure sensitive. Variability on signal quality across populations (skin tone or obesity and diabetes also reducing perfusion).
	IPG	Also used on wearable devices for HR assessment. Off-the-shelf front ends available that can be used to access additional physiological parameters such as bio-impedance and electrodermal activity.	Electrodes require extensive development. Might not be suitable for long term monitoring. Also sensitive to motion artefacts.
	MPG	Low power method, sensor biased by permanent magnet. Low complexity electronic required.	Physiological process involved not fully understood. Motion artefact sensitive.
Ballistic	Mechanical	Very low power method and simplistic conditioning electronic in case of piezo transducers.	If required, pre-strain system can cause disagreement in case of long term monitoring. Specific pulsatile measurement location required, high sensitivity to motions.
	RF	Contact-less method.	Custom electronic required for conditioning. High sensitivity to motion artefacts.
Electrical	ECG	Access to gold standard measurement from wrist, in addition with PPG, can be used to evaluate blood pressure using PTT method. Low power required for operation.	The user needs to place his finger on electrode. Not suitable for continuous monitoring.
	Far-field	Enhanced user acceptability compared to ECG, no action is required from user for measurement.	Requires extensive signal processing that might not be suitable for an embedded device. Muscular electric activity (EMG) superimposed. Still in development.

design that will determine the robustness of the measurement device for this type of motion artefacts.

- **Motion of tissues underneath the skin:** This particular subgroup of motion artefacts, also known as micro-motions, are caused by the contraction of muscles and modification in soft tissues during hand, fingers and wrist motions. In case of hand flexion and extension, muscles such as the *flexor digitorum superficialis* or the *extensor carpi ulnaris* located in the forearm will cause deformations in tissues. For photoplethysmographic methods, it results in a modification of the scattering and reflection coefficients of the light. For ballistic methods, it results in deformations at skin surface that will be picked up by the transducers.
- **Contact pressure between the sensor and skin:** Finally, the contact pressure between the sensor and skin can lead to the superimposition of a motion component over the cardiac component. In PPG measurements, this phenomenon causes variations in optical contact with the skin and affects blood circulation, resulting in modifications of scattering and diffusion coefficients, and thus in the light detected by the receiving element. Similar consequences of motion artefacts can be found despite the acquisition method used: mechanical, IPG or MPG.

These three main causes of artefacts will give rise to superimposition of artefacts components in addition with the cardiac component of interest. Additive or multiplicative models were proposed in the literature to describe the interactions between motion artefacts and cardiac components [35]. In addition to their physical origin, motion artefacts can be classified by their harmonic content:

- **Periodic motions**, usually caused by intense physical activity such as running where arm swigging occurs. In these conditions, the wrist is affected by a periodic acceleration resulting in a periodic pattern for motion artefacts.
- **Non-periodic/random motions**, naturally encountered in human gestures during everyday life activities. Because of their transient and non-stationary content, this type of MA are more difficult to reject (when compared with periodic and stationary artefacts).

6.3. Motion artefacts reduction techniques

The challenge in addressing these artefacts remains in differentiating physical quantities variations caused by cardiovascular activity, from those caused by user motions. For this purpose, additional information sources can be used, such as a physical reference correlated with motion, or a synthetic reference composed of previously recorded motion artefacts. However, an additional reference could be avoided, as multiple techniques for MA reduction were shown not to require such a signal. On Fig. 15, we propose a classification of MA reduction techniques independently of the modality of measurement. Generally, most of the methods found in the literature apply to PPG signals which is the most common way to assess heart rate from the wrist.

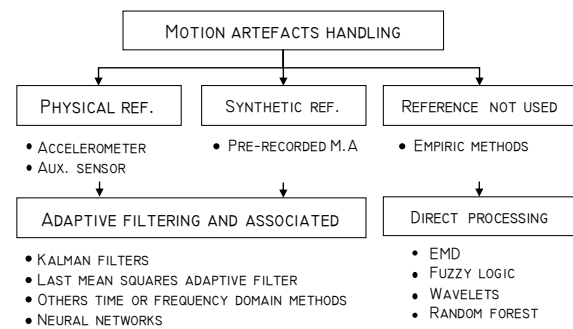


Fig. 15. Proposed classification for motion artefacts handling

Physical reference. Methods using an auxiliary physical signal are common in PPG. The most used reference in literature is accelerometry as motion data is highly correlated with MA during physical activity [107]. In order to exploit mutual information contained in acceleration signals, a wide range of time-domain or frequency-domain algorithm have been developed for addressing both periodic and random motions. In particular, time-domain adaptive filtering has been widely used with accelerometer signals [108, 109, 110, 111]. Denoising can also be performed using frequency-domain methods [112, 113], for example sparse signal reconstruction [114], joint sparse spectrum reconstruction [115] or empirical mode decomposition with spectrum subtraction [116]. However, as reported in [117], both adaptive filtering and wavelet-based methods introduce nonlinearities and variability in phase component of PPG data and could potentially lead to erroneous levels of physiological parameters evaluated using time-domain signals and processing. In addition, these methods may be strongly limited when an accelerometer is not able to detect small amplitude motions close to the

measurement site, or when detected motion is not representative of what is really happening on the PPG device. For this purpose, Zhang et al. [118] used the infrared channel of a PPG device as a reference in continuous wavelet transform (CWT) subtraction. The acquired signal in IR domain is directly representative of the phenomenon happening during physical activity, including micro-motions. This multi-spectral approach can also use lower computing-cost algorithms such as adaptive filtering as proposed by Yousefi et al. [119]. Similar techniques using the same transduction principle for both cardiac and motion artefact evaluation can also be found in mechanical measurement methods [78]. Lastly, the use of interferometric references was proposed by [120] for measuring one dimensional vertical motion. This reference is acquired using self mixing interferometry in association with Fizeau-Doppler demodulation. However, in this preliminary study, experiments were only conducted in laboratory conditions using skin phantom and computer-controlled motion platform. Moreover, form factor of interferometric measurement device used is not suitable for integration in a wrist-worn device. In further work, the interest of this method to reduce periodic motion artefacts was demonstrated in ambulatory conditions [121].

Synthetic reference. The signal used as a reference for MA reduction can also be estimated online directly from the physiological signal. A method proposed by Ram et al. [122] implements a synthetic reference generator that produces an estimation of the noise corrupting PPG signals, by using different signal processing approaches (FFT, SVD, ICA). A randomness test is then applied to select the most relevant reference for adaptive filtering. A similar approach is also implemented in a proprietary algorithm used in commercial instrumentations for SpO₂ estimation [123]. The use of artificial neural networks for MA reduction was also considered as a synthetic reference. In [124] a multi-layer perceptron neural network has been trained with both clean and noisy PPG records. It is then able to reduce motion artefacts from same subject on whose training dataset has been recorded. In an other approach, Lee et al. [125] used previously recorded clean PPG signals as a reference for selecting a filter within a bank. Correlation is used as the indicator to choose the filter that will produce the optimal PPG signal. The main advantage of these methods remains in the fact that no supplementary hardware is required.

Reference not used. Finally, methods based on direct signal processing without using any external reference

are extensively represented in the literature. Independent component analysis (ICA) has been used by Kim and Yoo [126] for MA reduction; it is based on the assumption that heart rate component and motion artefacts are independent of each others. However, ICA-based methods requires a knowledge model to be adjusted in function of sources which are supposed to be independent. Moreover, in case of physical activity synchronised with heart rate, the independence between MA and cardiac component is no longer verified, resulting in poor denoising performance. Regarding SpO₂ measurements, which are carried out using PPG amplitude, the use of adaptive filtering in addition with ICA can be used to preserve amplitude during MA events [127]. Time-frequency domains methods and statistical analysis were also used for MA reduction [128, 129, 130]. Finally, based on the analysis of normal PPG waveform characteristics such as slope, peaks and valleys, the use of fuzzy logic has been proposed by Liu et al. [131] for estimating heart rate during MA events.

The previously discussed methods, regardless of whether a reference signal is required or not, were evaluated by Naraharisetti and Bawa [132] in terms of accuracy for the estimation of heart rate. In this study, cycle Fourier series analysis (CFSA) and single value decomposition (SVD) shown best results in comparison with adaptive filtering and wavelet transform. However, the conclusion of a better accuracy using non-referenced methods was obtained against referenced methods using a synthetic signal in adaptive filter instead of a real accelerometer.

7. Conclusion

Direct comparison between methods presented in this paper is made difficult mainly due to the experimental conditions. Firstly most of these conditions are not precisely described, and secondly most of them are resting conditions. Performing an evaluation in ambulatory conditions requires portability of the measurement device, which is not systematic, especially with early prototypes. Particularly, motion artefacts inherent to ambulatory conditions will inevitably deteriorate physiological signals, thus, reducing precision of the overall system. In addition, there is no universal, reproducible testing protocol independent from the measurement method. In most studies, validation against ECG gold standard is not systematic, and in worst cases, PPG has been used as the gold standard. Relying on indirect

signals such as PPG is a questionable choice. As candidates to surrogate ECG for the assessment of HR, ECG should be used as a gold standard when demonstrating the performances of new methods. Due to the lack of agreement measurements with gold standard, precision of exploratory methods such as microwaves or mechanical based methods cannot be evaluated.

Based on prior art referenced in this study, two measurement methods can be highlighted : PPG and IPG. Statistically, most wrist-based measurement devices described in literature rely on these methods. This statement is also valid for commercial devices where PPG is extensively used in personal fitness trackers, smartphones or smart watches. Impedance plethysmography is also used in the Jawbone UP2 (San Francisco, California, US) connected wristband. Main motivations to use these methods are their ability to be integrated in small form factor devices, their relative low power, their mechanical robustness and high acceptability score for users. Because these methods are ideal candidates for consumer electronic market, integrated circuits (ICs) manufacturers provides a variety of analog front-ends dedicated to these applications. These methods can today be considered as industry standards. It is nowadays common for major actors of electronic industry to offer a wide range of evaluation kits and reference designs for bio-sensing using PPG or IPG. Consequently, time-to-market and hardware development costs are reduced for both industry and academics.

As the associated acquisition and measurement equipment become more accessible, physiological measurements based on PPG are performed by a larger academic audience. Thus, motion artefact reduction techniques previously discussed are essentially designed and validated with PPG signals. Regarding artefact reduction techniques and as previously discussed, there is no ideal technique with optimal performance along all types of motion artefacts. Additionally, a trade-off must be found to benefit from methods relevant for wide amplitude motions (for example, based on accelerometry) and those based on auxiliary wavelength especially suited for micro-motions. Finally, as the use of PPG for assessing cardiovascular system parameters in ambulatory conditions tends to generalise in both academic and industrial sectors, it is very likely that further work will be focused on artefact reduction dedicated to this measurement technique.

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