4

Non Contact Heart Monitoring

Lorenzo Scalise
Dipartimento di Ingegneria Industriale e Scienze Matematiche,
Università Politecnica delle Marche,
Italy

1. Introduction

Heart rate is, among the many vital signs (respiration rate, blood oxygen saturation, arterial blood pressure, etc.), one of the most commonly measured and monitored. Whatever will be the sensing principle or the monitoring method used, data referred to the heart rate can be considered the primary vital sign information which is needed on a patient approach in both emergency and clinical situations. Heart rate data are used to measure anomalous rate or irregular pulse rate (arrhythmias) or heart block. The post-processing of the data can be used to verify trends or single events, providing precious elements to the patient diagnosis. Heart-rate variability (HRV) can be performed on recorded data in order to have an objective measure of eventual cardiac abnormalities (irregular beat-to-beat time is a prognostic factor for atrial fibrillation (Gorgas, 2004). Low HRV is also a known prognostic marker for several cardiovascular diseases. Other possible use of the heart rate data are related to the analysis of the circadian rhythm (sleep), temperature regulation, cardiac sympathetic nervous activity and synchronization with respiration rate.

Since the past centuries, observation of the electrophysiologic effects related to the heart beats are reported. In 1842, Carlo Matteucci, Professor of Physics at the University of Pisa, shows that an electric current accompanies each heart beat. In 1902 Einthoven publishes (Einthoven, 1902) the first electrocardiogram recorded on a string galvanometer, opening the way to the electrocardiography (ECG) era which is still, nowadays, the primary heart rate monitoring procedure.

To date, an enormous series of procedures, methods and devices for ECG monitoring are available on the market (Gorgels PM, 2007; Webster, 1988). The majority of these contributions are based on the need to place some electrodes on standard positions on the body surface (i.e. Einthoven's triangle), as depicted in fig. 1. ECG measurements can be divided into two types according to where electrodes are attached or fixed. The first type involves measurements with conventional "fixed-on-body" electrodes such as Ag – AgCl electrodes, and the other involves measurements using electrodes installed on appliances or furniture.

Even if fixed-on-body electrodes (fig. 2) are reliable and give good signal quality, they are inconvenient and inadequate for long-term, everyday measurements. Moreover the presence of cables (one for each of the electrodes placed) can considerably limit the patient mobility and comfort, forcing him to maintain the initial position (supine) for all the monitoring period of time or limiting his/her movements because of the cables length."

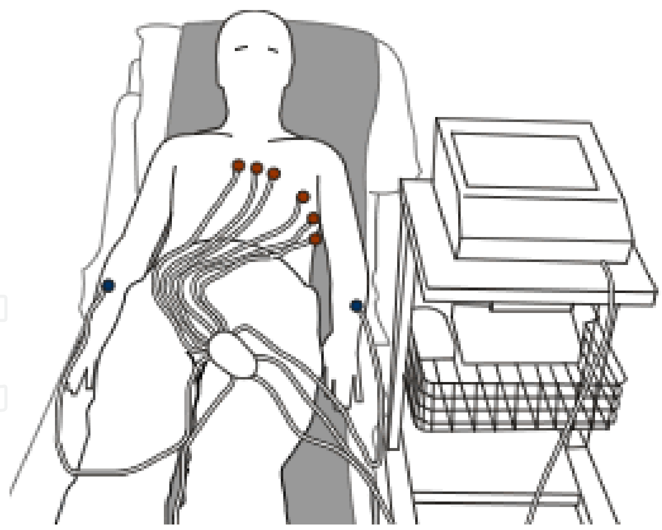


Fig. 1. Example of ECG electrodes placement and patient positioning for standard ECG monitoring.

Fig. 2. Examples of ECG on-body electrodes commonly used for standard ECG: disposable, gel-impregnated, attachable electrodes (left), clamp reusable electrodes (center) and suction, reusable electrodes (right).

"Fixed-in-the-environment" electrodes are nonintrusive and more adequate for long-term monitoring, and even if they also present some shortcomings, the nonintrusive nature of fixed-in-the-environment electrodes makes them an attractive option for daily monitoring. The main limit is the requirement of maintaining direct contact between the bare skin of the subject and the electrodes which is fixed in the environment (chair, seats, etc.), which limits the application of fixed-in-the-environment electrodes to a few cases.
An alternative possibility for heart monitoring is phonocardiography, PCG (Webster, 1988; Dressier, 1970; Fowler, 1962; Durand and Pibarot, 1995) which consists essentially in recording/processing the acoustic waves produced by heartbeats and travelling through the body up to the skin. It was one of the first heart monitoring method developed and it is based on the observation that heart contractions produce acoustic waves travelling through the blood at the corresponding speed of sound. Therefore acoustic pick up of such waves can provide information on the periodic variation of pressure wave due to the heart

pulsation , as well as other information related to the heart mechanics and valves dynamics (fig.3). The typical speed of dilatation pressure waves varies between 4 m/s to 10 m/s. Increasing stiffness of arteries and veins due to aging or cardiac diseases increases the speed of dilatation waves. Consequently, study of propagation of dilatation waves has a clear physiological and clinical interest. PCG can be considered a non-invasive method, but it still requires the contact between the high sensitivity acoustic transducer (typically a piezoelectric microphone) and the patient skin and therefore it can't be considered a solution to the problem of monitoring heart activity (in particular heart rate) without the direct contact between the device and the patient's skin.

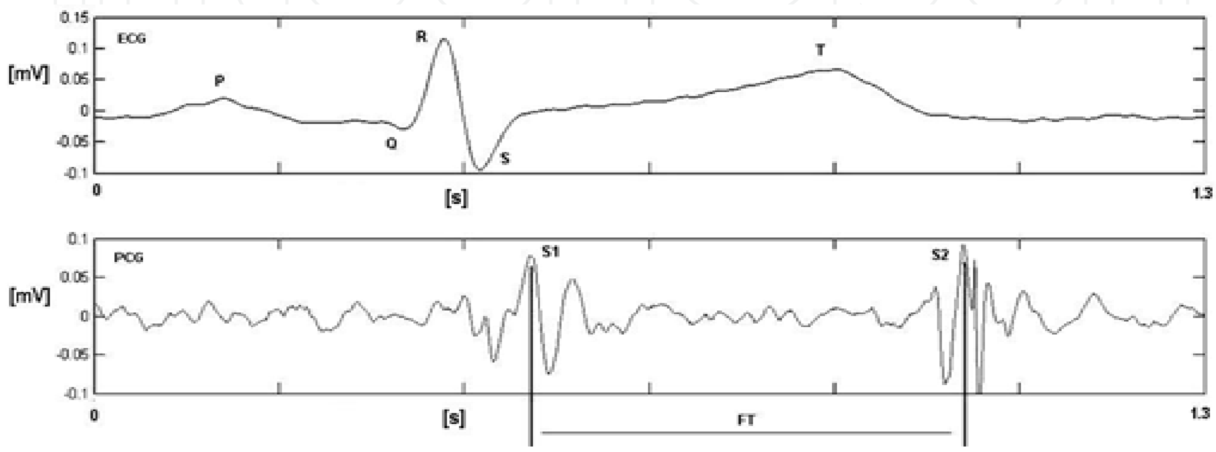


Fig. 3. Example of ECG II-lead trace (top) and simultaneous phonocardiogram (down) traces. S1 and S2 are the main signals measured during an heart cycle; (FT is the filling time).

A medical device frequently used to monitor patients pulse rate, is the pulse oximetry, PO (Webster, 1988; Alexander et al., 1989; Welch et al., 1990; Wahr et al., 1995). The device has been realized for the monitoring of oxygen saturation (percent of hemoglobin in the oxyhemoglobin configuration) in blood and is based on the different spectral absorption of hemoglobin and oxy-hemoglobin. In adults, PO's sensing elements are normally placed on the finger tip (or ear lobe) where the two emitting LEDs and the photodiode are placed (fig. 4). In addition to the digital read-out of O_2 saturation, most pulse oximeters display also a plethysmographic waveform as the blood micro-vessels under the skin expand and contract with every heart beat; from this trace, heart rate is automatically extracted.

As for the PCG microphone, the PO probe requires direct contact with the finger or ear lobe and it can't be considered a non contact method. The main drawback of PO is the sensitivity to motion artifacts which can generate repeated erroneous data and consequently activate false alarms signals.

In this chapter, the possibility to perform heart monitoring without physical contact with the subject is investigated and all available non contact methods are reviewed, trying also to predict the future trends of the technology. Before presenting such methods we will resume the motivations for such exploring the possibility of non contact monitoring of heart rate.

Non contact detection and monitoring of human cardiac activity without contact through bedding and clothing is a valuable tool in intensive care monitoring, long term-monitoring and home health care applications as well as in other non clinical fields such as the case of workers health monitoring (i.e. airplane pilots, firefighters, etc.). Patients with conditions

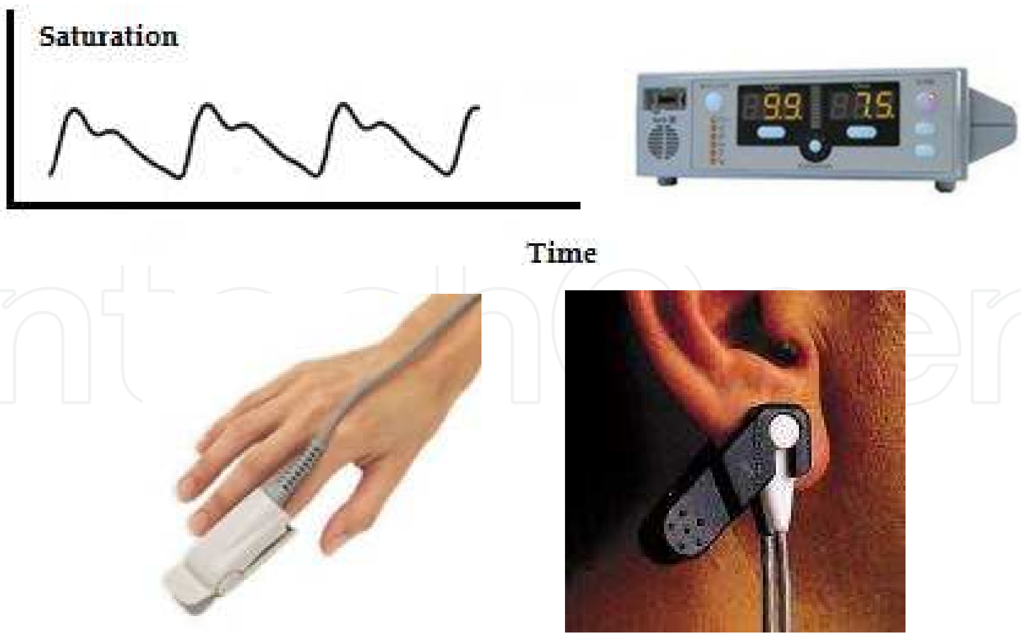


Fig. 4. Common pulsatile signal on a pulse oximeter (top, left); Oxymeter front panel reporting blood saturation and heart rate (top, right). Oxymeter probes: finger clip (down , left) and hear lob clip (down, right).

that can be perturbed or worsened by contact sensors include neonates (fig. 5), infants at risk of sudden infant death syndrome, and burn victims; a noncontact heart rate monitor provides a vital sign without affixed electrodes for these patients. Most alternatives to standard heart monitors need the application of electrodes or transducers (such as the case of thoracic belt) which often require accurate control or re-placement of electrodes/transducers during the monitoring period of time. Such aspect is sometimes critical and can also be impossible or undesirable in many situations.

Fig. 5. Possible skin irritation effect on preterm infant due to prolonged use of skin ECG electrodes.

An important advantage associated to the possibility to have a non contact sensing of the heart rate, it should be recalled, is the intrinsic compliance with the safety recommendations concerning risk of electric hazards which are particularly strong with electromedical apparatus. Moreover the absence of contact is particularly important in specific cases such as security (hidden or suspected subjects identification), monitoring of contaminated patients, presence and conditions of live subjects in biologically and chemically contaminated environments, emergency (i.e. presence and conditions of subjects on a battle field, transport of patients), intensive care (pre-term infants, adults), long monitoring of vital signs as well as home monitoring especially of elderly adults (Abowd, 2002) and automatic activation emergency actions.

A first class of proposals in this field has been based on the usage of non conductivity electrodes for ECG recording for example with electrodes embedded in the bed or the chair (Ishijima, 1993; Matsuda et al., 2008; Lim et al., 2006 and 2007), or have been based on techniques such as the ballistocardiography (Pollock, 1957; Morris, 1954) where the sensing part is installed into the bed, chair or wheelchair, in furniture or is directly based on the vibration monitoring of the skin, such as for seismocardiography (Salerno and Zanetti, 1990; Poliac et al., 1991; Sandham et al., 1998).

More recently a laser-based, single point, non contact measurement method, named vibrocardiography (VCG), for the heart rate (HR) monitoring and the heart rate variability (HRV) assessment has been proposed. First studies using optical vibrometry (laser Doppler vibrometry) has been reported for the identification of the arterial pressure waves (Tomasini et al. in 1998). While a novel method, called vibrocardiography (VCG), has been later proposed to measure the velocity of displacement of the skin in correspondence of the chest wall (Scalise et al., 2005a;2005b;2006; Morbiducci et al., 2006; Scalise and Morbiducci, 2008). VCG has been demonstrated to be valid for the assessment of the cardiac frequency and variability and in Cardiac Resinchronization Theraphy as support in pacemaker programming after installation (Bocconcelli et al., 2006). The same laser-based optical approach has also been explored to have HR data correlated to the emotional state of the subject without the need of a physical contact (Rohrbaugh et al., 2007) as well as for the evaluation of biometric capabilities (Chen et al., 2010).

Other non contact methods, based on electromagnetic (EM) approaches for heart and respiration monitoring, have also been proposed. The first microwave system (a microwave Doppler radar) was firstly proposed in early 1970s (Lin, 1975) for respiration detection and was based on bulky, heavy and expensive components; today many solutions have been proposed demonstrating the possibility of compact, lightweight and inexpensive mass-producible solutions (Staderini, 2002; Matsui et al., 2004a and 2004b; Kim et al., 2007; Droitcour et al., 2001). The interest on the use of EM heart monitoring are based on the possibility to have data from the subject without the necessity of the direct contact with the skin or just the visibility of the skin (electromagnetic waves at certain frequencies can pass through the tissues). A series of different laboratory solutions as well as market available monitoring systems (see web references) have been already proposed. Finally HR monitoring based on image-based methods (Da Costa, 1995; Poh, 2010; Takano and Ohta, 2007; Garbey et al., 2007), electrical impedance (Ischijima, 1993; Harland et al., 2002), acoustic and ultrasound (Tanaka et al., 2002) approaches have also been reported as valid principles for non contact heart rate monitoring.

2. Non contact methods for heart monitoring

Non contact detection and monitoring of human cardiac activity through bedding and clothing would be a valuable tool in intensive care monitoring and home health care applications. Patients with conditions that can be perturbed or worsened by contact sensors include neonates, infants at risk of sudden infant death syndrome, and burn victims; a noncontact heart and respiration rate monitor could provide vital signs monitoring without affixed electrodes for these patients. Most alternatives to standard heart monitors need leads and contacts and often require accurate control or placement; this may be impossible or undesirable in many situations

In this chapter, the non contact methods for heart monitoring proposed in the last years have been divided in four categories, based on the working principle:

- Electromagnetic-based monitoring systems
- Laser-based monitoring systems
- Image-based monitoring systems
- Other methods

The common characteristic for all the methods, presented in the following, is that they aim to measure the surface (skin) displacement taking place because of the heart muscles contractions: change in the volume of the heart and consequently displacement of the external heart walls during systole move the tissues under the ribs and the soft intercostals tissues, causing the well perceived "beat" that everyone can experience simply putting the finger tip on the left-upper part of the thorax. Specific studies – carried out with optical methods - report that the maximum displacement takes place in the correspondence of the heart apex and it is reported to be in the order of about 600 μm (Aubert et al., 1984; Ramachadran and Singh, 1989). It must, anyway, be remarked that the motion of the surface due to the heartbeat, which is indeed detectable also from the right part of the thorax, is largely influenced by the subject health conditions and position (supine, prone or on a side), by the gender, by age and by body shape.

2.1 Electromagnetic-based monitoring methods

The basic principle of radar is to transmit a microwave (radio) signal towards a target. The strength of the backscattered signal is measured. There are two variants of radar sensing used for heart rate monitoring: continuous-wave (CW) and wide band pulsed radar (UWB).

Constant wave (CW) radar emits a continuous stream of electro-magnetic radiation. An antenna is used as transmitter and it radiates a signal to a target, the energy reflected from the target is detected by an antenna (it can be the same antenna used for transmission) and a mixer diode provides a tension proportional to the phase between the transmitted and received signal (which is related to the target movement). A filter section is needed to separate heartbeat from the respiration; valid measurements could be taken at a range exceeding 10 m. Microwave apexcardiography was demonstrated firstly with a continuous-wave 2 GHz antenna placed in correspondence to the apex and precordial motions were detected (Lin et al., 1979). In general, CW radar methods, reported in literature, appear simpler respect to UWB radar, but it presents problems when multiple reflections, due to scattering characteristics of the surrounding environment, are present. With CW radar, the phase of the received signal is containing the information on the displacement of the target *x(t)* and, if we report the transmitted signal *T(t)* as:

$$T(t) = A\cos\left(2\pi ft + \phi(t)\right) \quad (1)$$

Where, *f* is the frequency of the transmitted signal and *Φ(t)* is its phase; then, the received signal *R(t)* can be approximated, as:

$$R(t) = KA\cos\left(2\pi ft - \frac{4\pi d_0}{\lambda} - \frac{4\pi x(t)}{\lambda} + \phi\left(t - \frac{2d_0}{c}\right) + \theta_0\right) \quad (2)$$

where, K is the reduction of the amplitude A of the originally transmitted signal, θ_0 is the phase shift due to the reflection at the surface, d_0 is the distance between system and the skin surface. When the phase of $R(t)$ is demodulated a signal proportional to the chest displacement (about ±1cm max, caused by the respiration activity and ± 0.6 mm, max caused by heart activity) can be inferred. The CW radar monitoring principle is reported in figure 6.

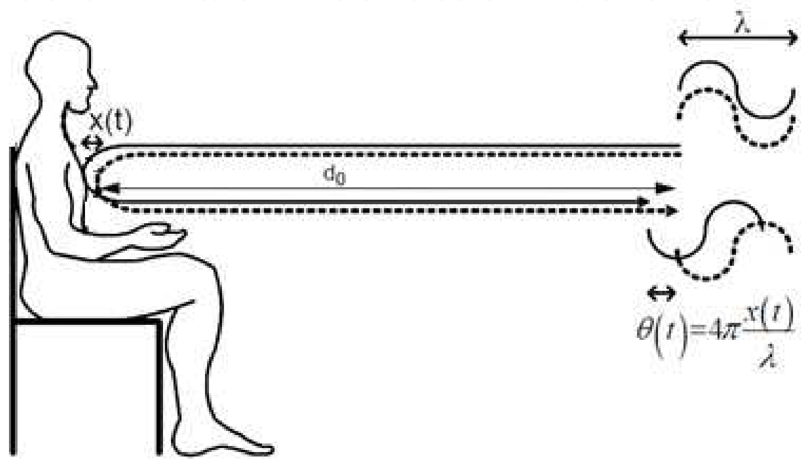


Fig. 6. Principle of CW radar monitoring of the chest movement: phase shift $\theta(t)$ caused on the reflected wave by the chest displacement x(t).

CW microwave reflectometers typically have a phase resolution of 1/200 fringe which corresponds to a spatial resolution of λ/ 400, where λ is the incident wavelength (Tateischi et al., 2007). For example, if a CW system working at a carrier frequency of 10 GHz is used (λ=3.0 cm) is used, the spatial resolution becomes 75 μm, allowing a full detection of the heartbeats. A typical technique used to extract from the detected signal the part due to the heart rate, is by use of an high-pass filter with a cut-off frequency of 0.7 Hz. Nevertheless, most of the authors propose specific signal processing in order to enhance the SNR of the heart beat signal and for the removal of residual motion artifacts (Nagae & Mase, 2010). With opportune filtering and signal processing, it was demonstrated that heart rate and respiration rate can simultaneously be detected by the same microwave apparatus (Chan et al., 1987). Particular measuring conditions taking into consideration the presence of persons in rubble (Chuang et al., 1990), athletes (Greneker et al., 1997) and persons behind walls (Chen et al., 2000) were explored, demonstrating that the approach is robust and mature for a wide employment on different scenarios. An example of a possible laboratory set-up using CW microwave reflectometry to detect respiratory and heart rate activities is reported in figure 7 (Scalise et al., 2011).
Experimental unfiltered phase signals acquired with the measurement system reported in fig. 7 (emitted signal at 6 GHz, 1 mW) are reported in fig. 8, for the case of a volunteer sitting on a chair and normally breathing (left) and holding the breath (right).
For HR monitoring scope, the direct comparison between phase variation, as detected using the CW set-up reported in figure 7 and the simultaneously acquired ECG trace of the subject are reported (fig.9), putting in evidence the high time correlation between the two signals.

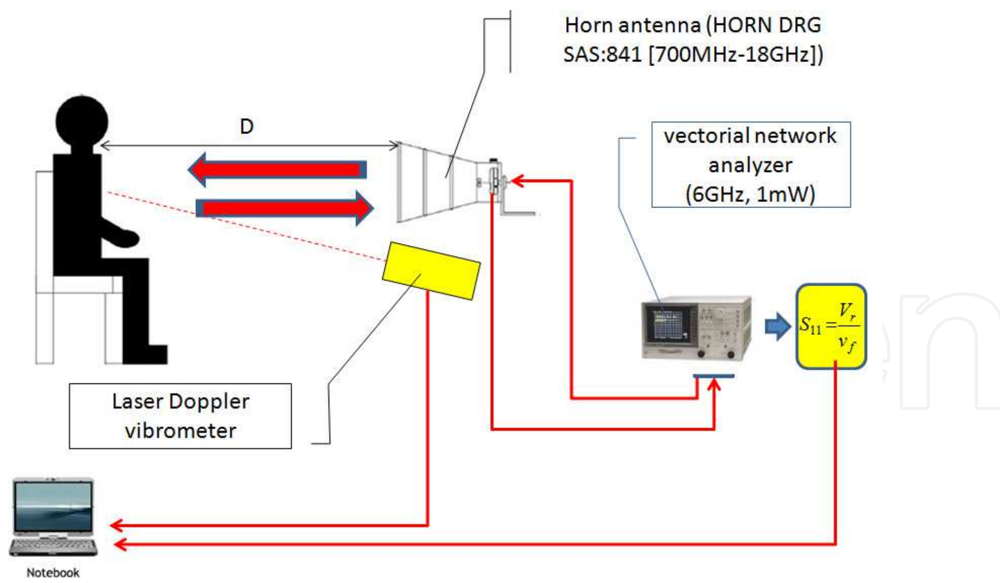


Fig. 7. Laboratory set-up for CW microwave reflectometry for respiratory and heart rate activities monitoring.

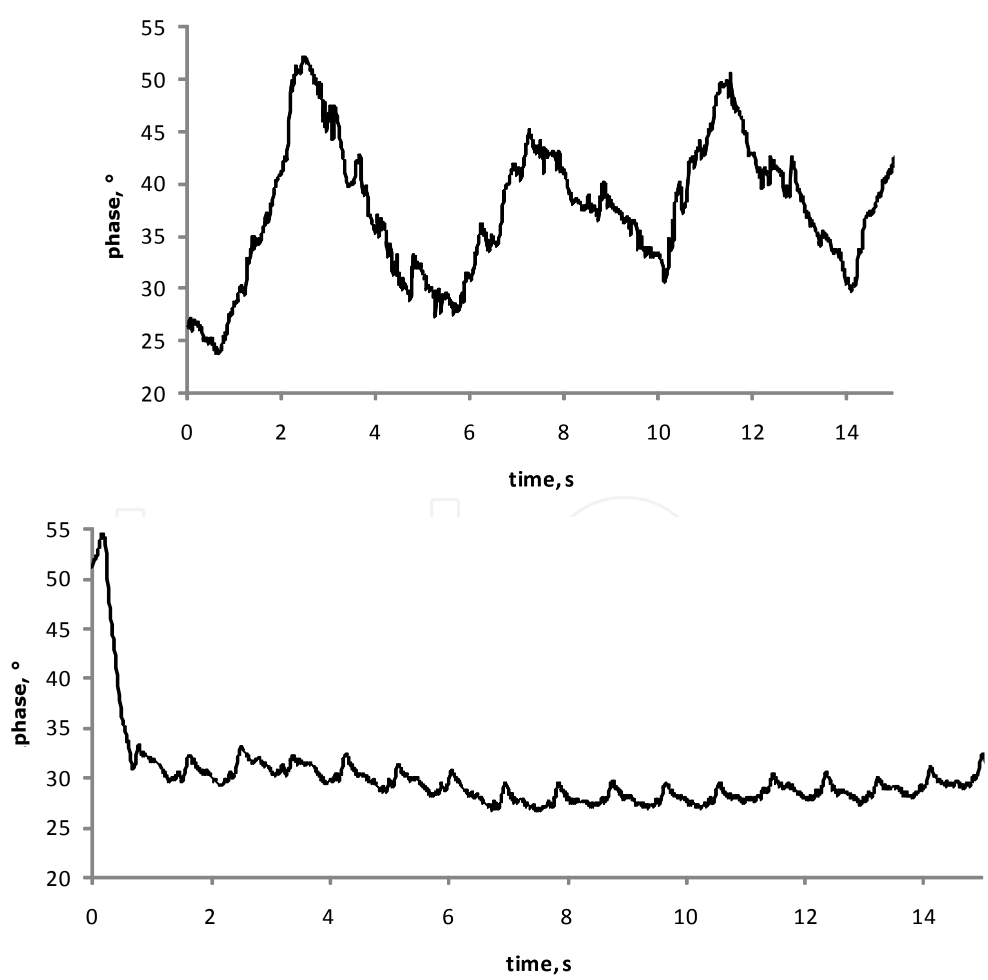


Fig. 8. CW microwave reflectometry: Respiratory and heart beat activities (top) and only cardiac activity (bottom).

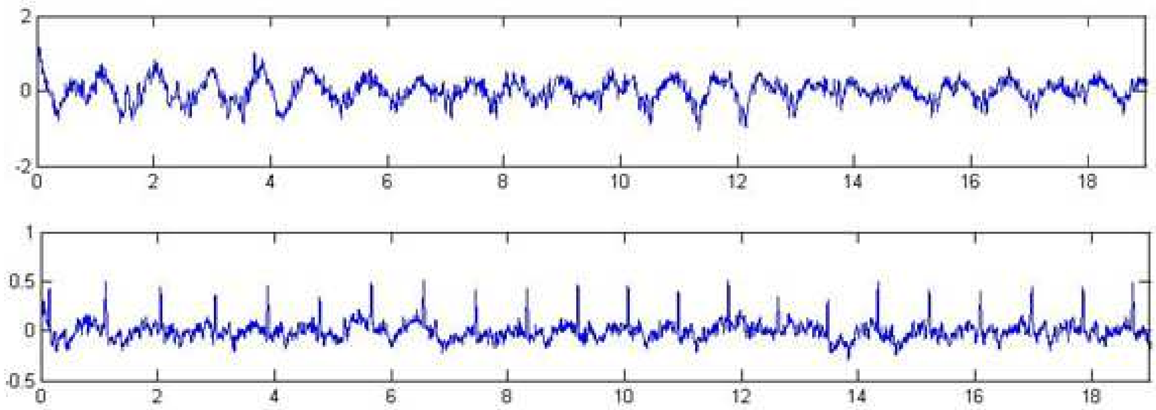


Fig. 9. Phase variation (normalized data respect to the max phase value, top) and ECG trace (down).

HRV analysis with a 24 GHz CW microwave system monitoring a volunteer from the back of a sitting subject at a distance of 34 mm, showed a strong correlation of the HRV measures from the microwave system with that from the ECG system in 2-minute recordings in rest and during simple arithmetic task. Comparison of 5 min recordings demonstrated also that there were no significant differences in the temporal, frequency domains and in non-linear dynamic analysis of HRV measures derived from heartbeat and ECG, which suggested this technique may used as practical alternative to ECG for HRV analysis. (Lu et al., 2009).

The second possible approach to the radar monitoring of the heart activity is based on the use of pulse radar (Staderini, 2002; Chia et al., 2005; Immorev & Tao, 2008) which are in the majority of the cases ultra wide band pulsed (UWB) radar. Unlike narrowband systems, which transmit continuous waveforms at a specific frequency, ultra-wideband (UWB) systems transmit narrow impulse-like signals that span a broad frequency range. The pulse width of such system is typically within a range of 100's of ps to several ns, with rise times as fast as 50 ps, corresponding to a frequency range that can span several GHz. Since the energy of the pulse is distributed across a frequency band, the power spectral density is much lower in magnitude than a narrowband system reducing also the eventual interferences with other RF or MW apparatus.

UWB application used to be limited mainly in military areas, however, since 2002, FCC has gradually allowed the commercial usage of these bandwidths (frequency for the UWB technique is 3.1-0.6 GHz in USA and 3.4-4.8 GHz and 6-8.5 GHz, in Europe) The power radiation requirement of UWB is strict and it usually it does not disturb the other equipments because UWB's spectrum is normally very low. Emitted pulses are spread over a wide frequency spectrum having a very short time duration (in the order of ns or sub-nano seconds of duration); the main advantage of such type of radar appears to be the low energy consumption due to the short pulses. Pulse radars make use of a pulse generator to allow the e.m. pulse transmission from the antenna and simultaneously the activate a so-called delay line used for controlling the sampling of the received echoes. Time duration between emitted and received echo is proportional to the target distance; the receiver can be activated at very short time intervals triggered by the delay-line (range gating). Thus, the length of the delay-line ensures that only pulses back-scattered from a certain distance are received. UWB is ideal in range measurement but can present some ambiguity in both range and velocity measurements. A typical set-up using a UWB-radar system is reported in figure 10.

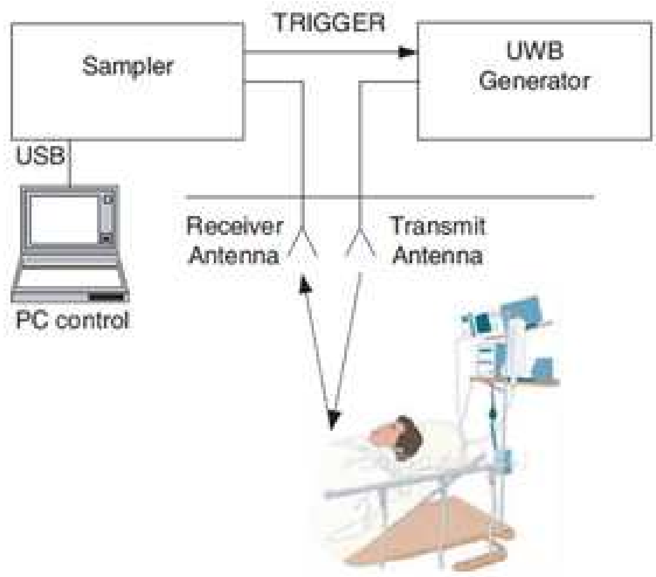


Fig. 10. Laboratory set-up for UWB-radar sensing of heart rate.

As for the case of CW systems, also UWB have been used to prove their capability in vital signs assessment. For example, data from 40 human volunteers at Walter Reed Army Institute of Research were collected using a micro-power impulse radar range finding prototype (Paulson et al., 2005). Readings from an ECG and pulse oximeter were captured simultaneously with MIR readings (Azevedo, 1996). MIR sensor readings were collected from each volunteer in four different body positions: standing upright, lying face up, lying on the right side, and lying face down. Since the readings of the MIR range finder prototype correspond to reflections off of tissue interfaces, rather than electrical impulses as in the ECG, body position was expected to be an important factor. Fixed range radar signals were compared to ECG, cardiac impedance, and acoustic heart signals showing well correlated relation among standard contact techniques (ECG) and the UWB-pulsed radar.
UWB's venture into the commercial domain of medical industry is now becoming a reality. Several vendors have been working on this technology to develop high value products aimed at the low-end and mid-range market segments in the patient monitoring and medical imaging markets (i.e. VSM, LifeWave and PAM™3000, see references). Some UWB systems are now available on the market allowing the check of occupancy of the bed and the vital signs monitoring of patients (primary elderly). In particular, if the at-risk patient should leave the bed without control or needed assistance, the UWB system, placed underneath the patient mattress, will detect it and send the alarm. Moreover the system is designed to wirelessly transfer heart rate and respiration rate data to the patient monitor or the nursering station. In figure 11, some example of market-available sytems are reported.
As against several imaging and patient monitoring modalities, UWB based medical technologies are expected to be impacted by the positive regulatory policies by European governments. In the United States, Federal Communication Commission (FCC) has mandated the use of UWB technology within the vast bandwidth of 3.1 to 10.6 GHz at power levels of -41 dB/MHz. On the other hand, there are no strict regulations mandating the use of certain frequency bandwidths for UWB devices for medical applications. In February 2007, the European Commission (EC) suggested that the use of UWB technology without mitigation is within the bandwidth of 4.2 to 4.8 GHz. It also accepted that in terms

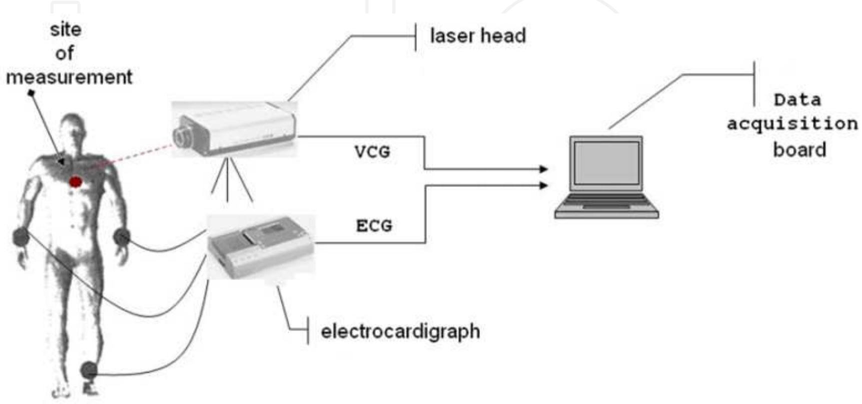
Fig. 11. Kai Sensors’ technology platform enables wireless monitoring of vital signs (left); Wireless2000 (center), Microimpulse’s VSM (right).

of the global use of this technology for medical as well as communication purposes, all UWB based devices shall operate above 6 GHz. Europe is likely to conform to this regulation by 2011. Furthermore, since most hospital devices operate at frequencies much lower than UWB, the possibility of electromagnetic interference caused by such devices is also minimal. This is likely to allow vendors convince the end users about the safety and utility of such devices, thereby easing the route to market entry.

2.2 Laser-based monitoring methods

In the 90’s the first studies of skin deflections due to arterial blood pulses appeared; these were conducted with optical measurement systems (mostly laser interferometers), which, in those years, started to be available on the market. By means of a bulky, fiber-optic, Michelson interferometer, pulsatile movement of the artery (arterial wall) were firstly measured near a stenotic lesion in an embedded vessel, the maximum gradient of skin displacement has been shown to be proportional to the time derivative of the blood pressure profile within the underlying vasculature (Hong & Fox, 1997). Data were taken firstly on a phantom, then at eight different sites for manual palpation of rhythm and strength of arterial pulse, radial, brachial, carotid, temporal, popliteal, femoral, posterior tibial, and dorsalis pedis. Three cardiac auscultation points were also assessed by stethoscope: pulmonary (second left intercostal), tricuspid (lower left sternal border), and mitral valves (fifth intercostal space at the midclavicular line). This attempt was consider extremely promising, but still possible only in a laboratory environment were the optical interferometer can operate correctly. Nevertheless the fiber-optical probe used in this approach was placed over the skin by means of a skin adapter using a stethoscope bell placed on the distal side of the fiber (measurement side) need in order to minimize unwanted signal artifacts due to body movement. The authors observe that further research may be necessary to see whether this stretches or alters the skin at the sensing site, thus perturbing the natural skin surface vibration effect. This approach must therefore be considered (minimally) invasive and with contact.

The possibility to detect the skin deflection due to the pressure pulse and correlate with the carotid pulse was firstly demonstrated (Tomasini et al., 1998) using a laser Doppler vibrometer (Castellini et al., 2006); in this case, the observation was carried out without any physical contact with a distance between the sensing head and the volunteers of about 1.5 m. The set-up proposed was based on the use of a single point Laser Doppler Vibrometer (LDVi) aiming to the volunteer neck in correspondence to the skin overlaying the carotid. Some years after, researchers from the same group started an extensive and systematic analysis of the opportunities offered by the use of the LDVi as a non contact method (named vibrocardiography, VCG) to measure heart beat (Scalise et al., 2005, 2006 and 2008; Morbiducci et al., 2006), as well as the respiration activity (Scalise et al., 2011), the heart mechanics (Bocconcelli et al., 2006) and the artery stiffness (De Melis et al., 2008). Typically (Scalise et al., 2005), VCG, is used to monitor superficial chest displacement due to the heart activity (fig. 12).



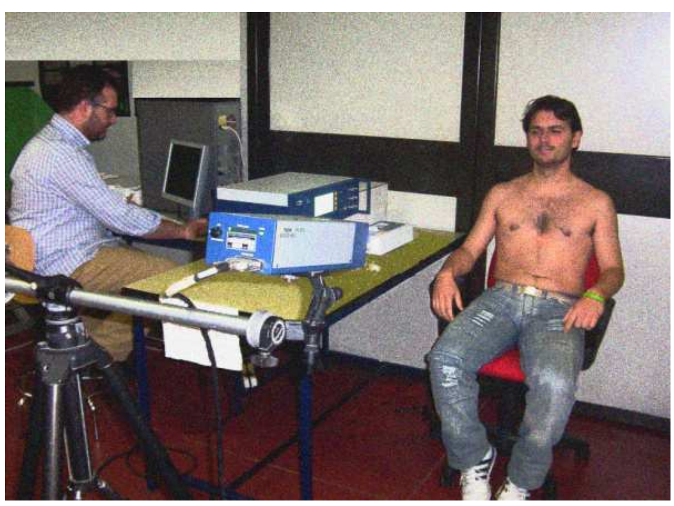


Fig. 12. Vibrocardiography set-up; Top: block diagram of the experimental set-up with the reference instrument (ECG); Bottom: Image of a test session.

Examples of typical signals, measured on two subjects (one male and one female), are reported in figure 13 with synchronous data acquisition from ECG reported for direct comparison.
Very high shore have been obtained with heart rate variability (HRV) carried out using the VCG (Morbiducci et al., 2006; Scalise and Morbiducci, 2008). Scatter plot of the HR measured by VCG and standard instrument (ECG) on the previously examined subjects, are reported in figure 14; the time series (fig. 15) built up from the time intervals between consecutive R peaks (RR), and VCG time intervals from the thorax (VV). Right side depicts a detail (50 beats) to show the close relationship of the two time series.
From results reported (Morbiducci et al., 2006; Scalise and Morbiducci, 2008), the results of the test of Bland–Altman put in evidence that significant differences are not present, from a clinical viewpoint. In fact, results showed mean percent differences of VCG derived descriptors, with respect to ECG ones, that do not threshold the 4.80% (3.03% mean value) for the LF/HF index and even lower for the other standard HRV parameters .

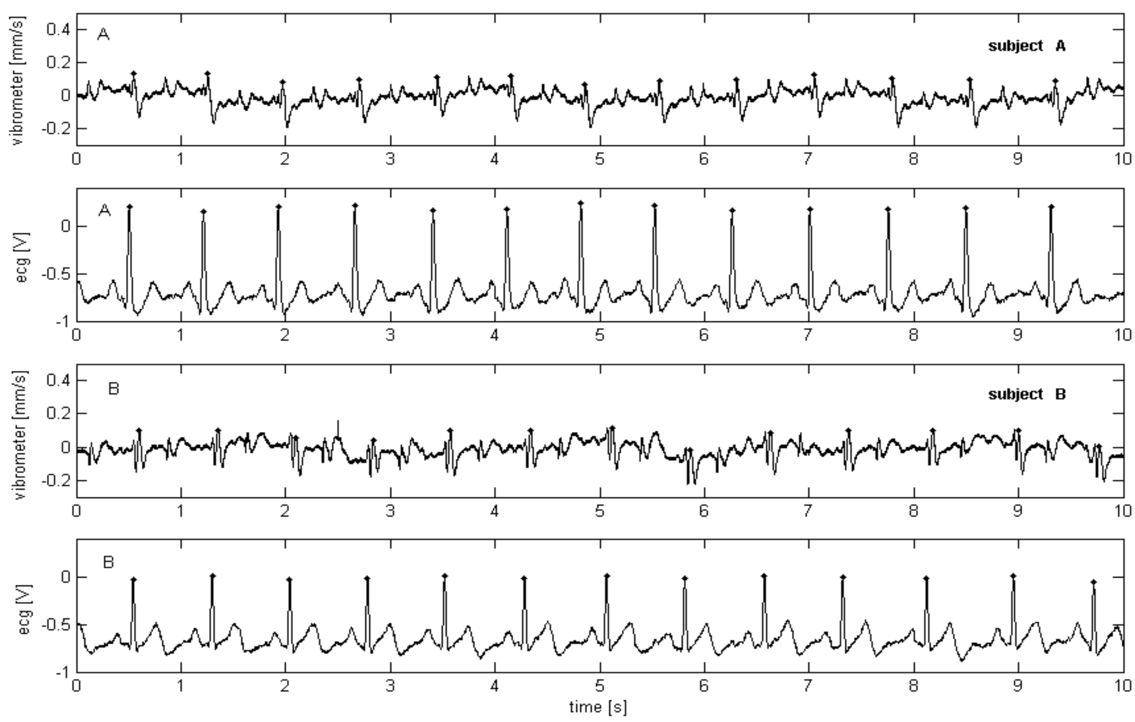


Fig. 13. Chest wall vibration (velocity) and ECG traces synchronously acquired, measured on two subjects (subject A: male, subject B: female).

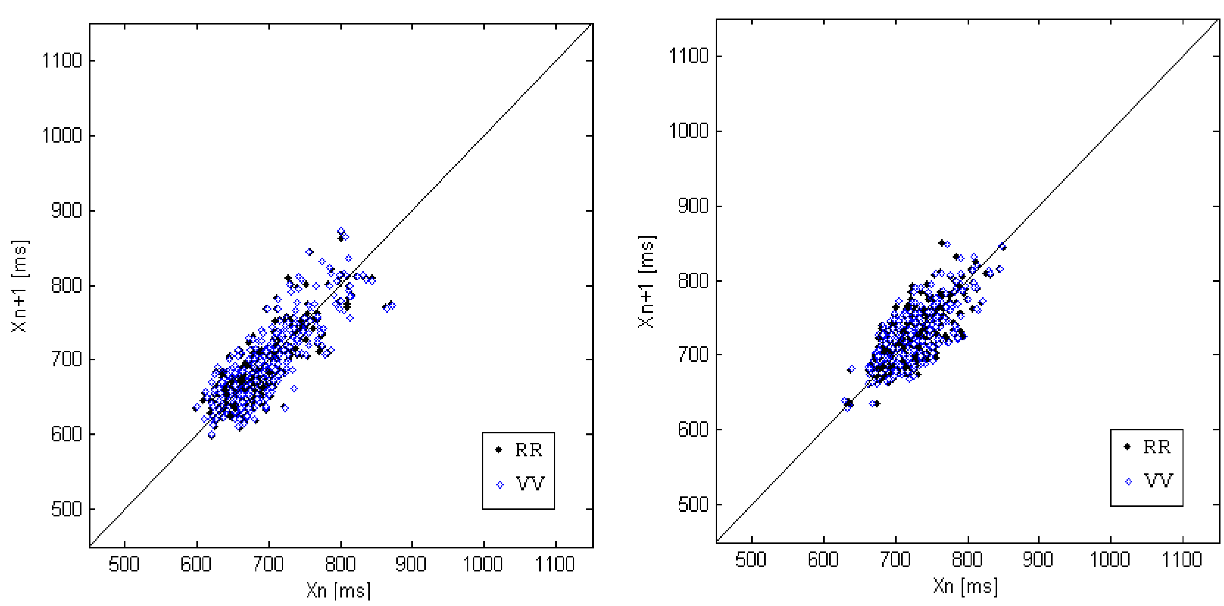


Fig. 14. Scatter plot of the heart rate measured by VCG (VV data) vs HR measured by ECG (RR).

The possibility to access the heart rate data from a different site respect to the chest wall was demonstrated (Scalise and Morbiducci, 2008) using the neck region (carotid site) as measurement site. High quality results are obtainable (fig. 16) and like for optical VCG signals measured in different sites of the chest wall (Morbiducci et al., 2006; Scalise et al., 2006), common morphologic features in the traces measured on the neck were observed,

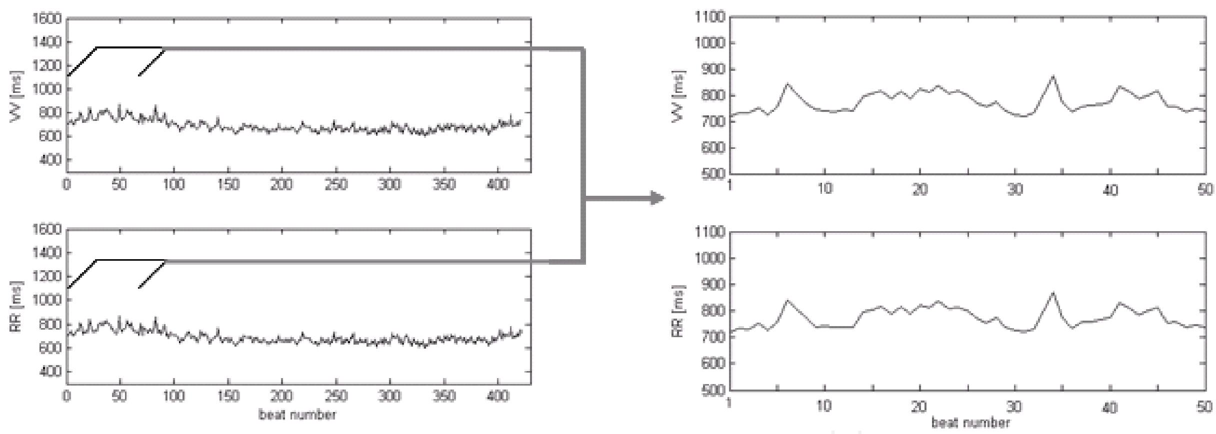


Fig. 15. Time series (relative to subject A and B of fig. 13) built up from the time intervals between consecutive R peaks (RR), and VCG time intervals from the thorax (VV). Right side depicts a detail (50 beats) to show the close relationship of the two time series.

with only inter-subject minor differences. HRV analysis, performed using VCG on the neck, agrees with the one derived from ECG with mean percent differences of VCG derived descriptors, with respect to ECG ones, < 2.94%. (relative to RMSSD) for the time domain and the 3.12% (relative to the sympatho-vagal balance LF/HF) for the frequency domain.

A very interesting comparison among ECG, phonocardiography (PCG) and VCG traces have been carried out in order to demonstrate the possibility to identify events of the cardiac mechanics, correlating the heart sounds relative to the closure of the mitral valve, and the following closure of the aortic and pulmonary valve with characteristic deflections identifiable on VCG traces (De Melis et al., 2007). Simultaneous acquisition of cardiac acoustic sounds, II-lead ECG trace and VCG are reported in figure 17, with indicated the S_1-S_2 and W_1-W_2 intervals (differences < 1.4 ms).

VCG was also explored (De Melis et al., 2008) as an alternative fully non contact method for the assessment of carotid-femoral pulse wave velocity (PWV), which is considered a gold standard method for assessing the stiffness of large arteries (high PWV values have been demonstrated to be associated with increased cardiovascular morbidity and mortality), and pulse transit time (PTT). With a simple approach (fig. 18), PWV and PTT were accurately measured with a mean PTTs of 75.85± 8.61 ms and 74.86± 8.63 ms for applanation tonometry (reference method) and VCG, respectively. A non contact method like optical VCG for the evaluation of large artery stiffness is considered of great interest because it could overcome limitations inherent to a contact method like arterial tonometry (based on contact piezo transducers manually kept in place), for which the debate on the influence of the contact of the probe over the skin on the measurements is still open in the scientific community.

A recent application of VCG has been reported as preliminary study for the cardiac resinchronization of patient using pacing devices (Bocconcelli et al., 2006). In figure 19, an example of simultaneous VCG and ECG beat monitoring is reported and compared with the same data acquired on a patient with weak ventricular depolarisation. Authors demonstrated the feasibility of VCG to provide data needed for a correct setting of the cardiac stimulator in order to provide optimal cardiac re-synchronization.

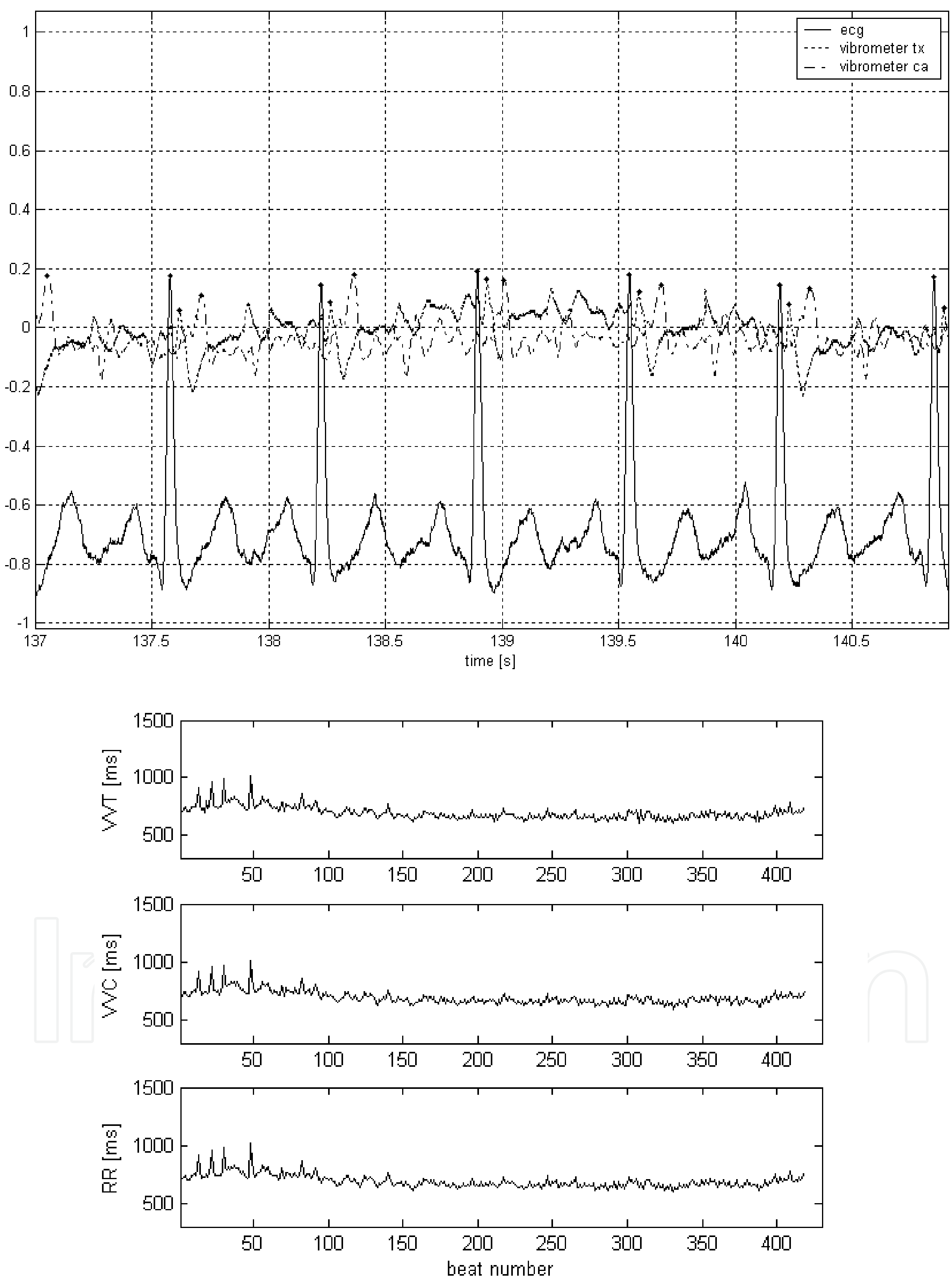


Fig. 16. ECG monitoring (in V) on a volunteer and simultaneous VCG trace (in mm/s) from chest wall and from carotid site (top);

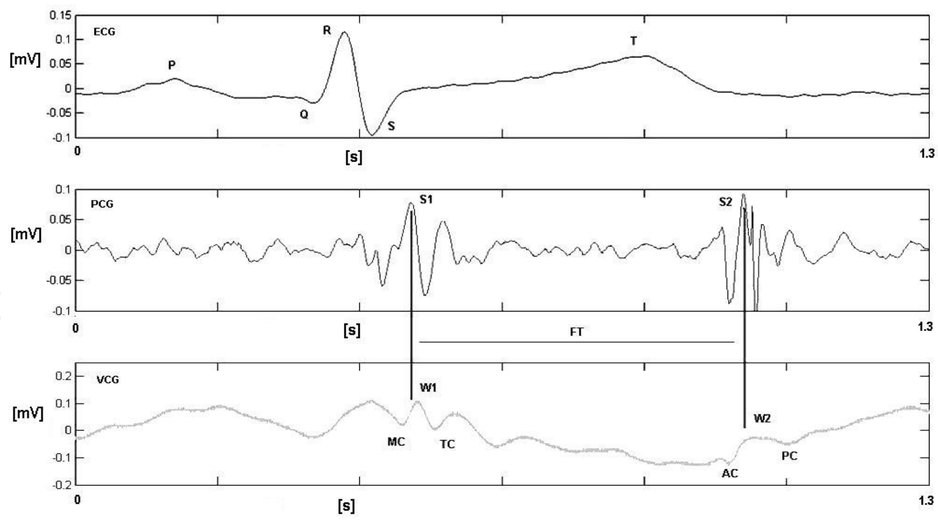


Fig. 17. Example of one beat recordings from VCG and PCG with II lead ECG compared with a gold standard vital signs representation.

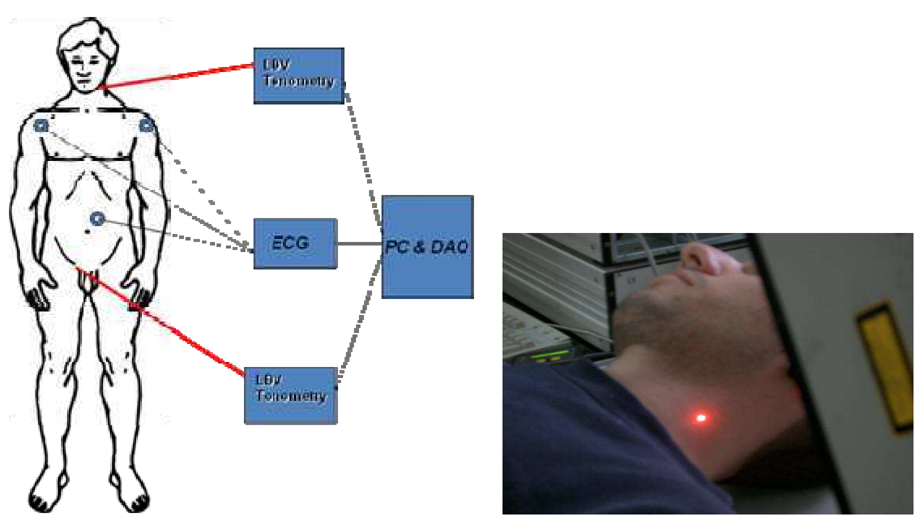


Fig. 18. Sketch of the measurement setup (left) and picture of measurement site (carotid artery).

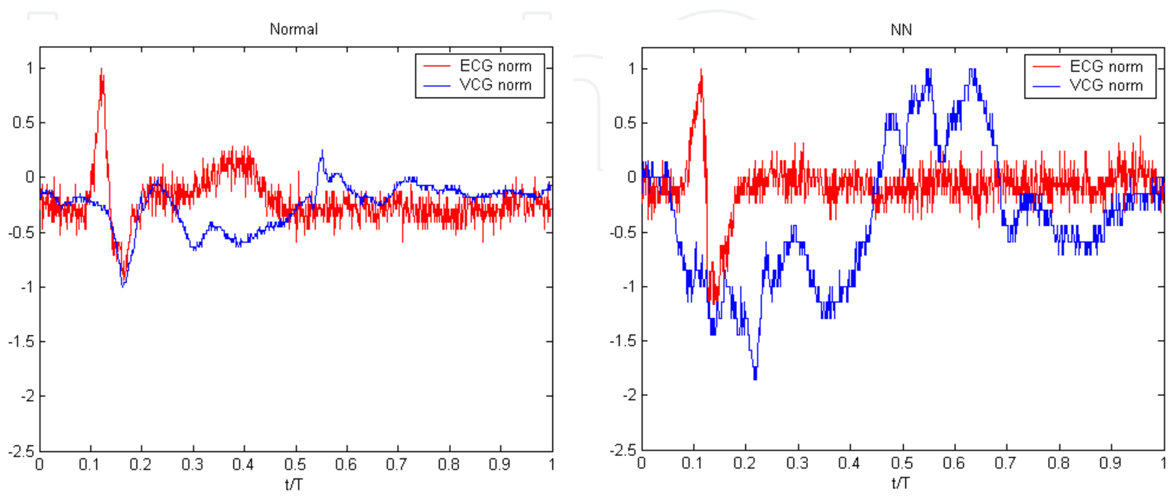


Fig. 19. ECG (in V) and VCG traces (in mm/s) for an healthy subject (left) and for a patient (right).

Very promising use of laser Doppler vibrometry have been proposed in the field of biometrics where the measure of cardiovascular activity as biometric marker appear to be promising (Chen et al., 2010; Rohrbaugh et al., 2007). In contrast to the traditional, contact ECG approach for cardiac activity monitoring, the signal examined is derived from movements of the skin associated with the blood pressure pulse in the underlying carotid artery using LDVi. The varied structure of the skin surface displacements (including some in the submicrometer range) comprising the LDVi signal would seem nearly impossible to mimic. Our studies confirm that the LDVi pulse signal is sufficiently textured to support a high level of discriminability among individuals. The signal changes in appearance somewhat from one occasion to the next, and is affected by factors such as physical exercise and mental stress. Data collected from 285 individuals showed good performance of LDVi and of the data processing method described especially on data collected within the same session; the discriminability characteristic of the LDV pulse is well maintained across an individual 5-min session. The increasing match error of the intersession tests may be due to time-varying factors.

Another field of application for laser-based heart rate monitoring methods has been proposed for distance monitoring of vital signs in risk environments. Proper functioning of the heart ensures the performance of the human body as well as sustenance of life, it can be used as a forward-looking indicator for assessing the performance limits of warfighters and athletes, prognosis of various health conditions, as well as for the determination of life signs in certain situations like triaging and battlefield management. Thus, in principle, detecting surface vibrations from the human subject allows information retrieval regarding the functionality and performance of the human heart. A novel laser-based vibrometer whose optical speckle-tolerant nature makes it ideal for assessing the surface vibration characteristics of, for example, human subjects whose surface quality is far from optically flat, is reported (Wang et al., 2010). The combination of excellent surface vibration detection sensitivity and optical speckle tolerance makes the novel pulsed laser vibrometer an ideal candidate for standoff monitoring and assessment of a human subject's cardiac activity and functionality. At the moment only preliminary, qualitative results are reported for laboratory tests with this special experimental set-up which seems to be extremely powerful allowing life signs detection in non-ideal measurement conditions (such as the case of directly on skin or directly on specific part of the body).

2.3 Image-based monitoring methods

The application of image-based methods for heart rate monitoring is based on the observation that the deflection of a human vessels due to heartbeats can be visually observed from the consequent deflection that the skin surrounding an important vase. One of the first use of image-based methods for HR monitoring (Da Costa, 1995; Parra & Da Costa, 2001) proposes to video acquire the deflections proposing two methods (fig. 20). In the first the skin is illuminated by a 2 mW HeNe laser beam in the neighborhood of the vein. The reflected speckle pattern is acquired, digitized and a specifically developed code, based on speckle image processing, is used to evaluate the skin displacement caused by the mechanical deflection produced by the arterial wall deflections due to the systolic pressure wave travelling along the vascular tree. In the second method a small mirror glued to the skin is illuminated by the laser beam. The position of the light spot resulting from intersection of the reflected beam with an opaque observation plane is recorded and plotted

as a function of time. Both the methods proposed allows the remote sensing of the heartbeats the later being (minimally) invasive requiring the need to glue a mirror on the subject skin.
Both the approach proposed don't report quantitative results; only a graph is reported and non correlations with reference signals (i.e. simultaneous ECG) are reported. Therefore, even if the solutions proposed were promising at the time when were presented, they can't be transferred to a real clinical environment or anyway can't be operated out of controlled conditions usual in laboratories of optics in the reported form. In fact, full filed digital speckle image experimental set-up, as well as laser reflection systems, need to have precise and accurate control of the subject to the set-up relative positions in order to isolate any artifact/vibration disturbance to the measurement data.

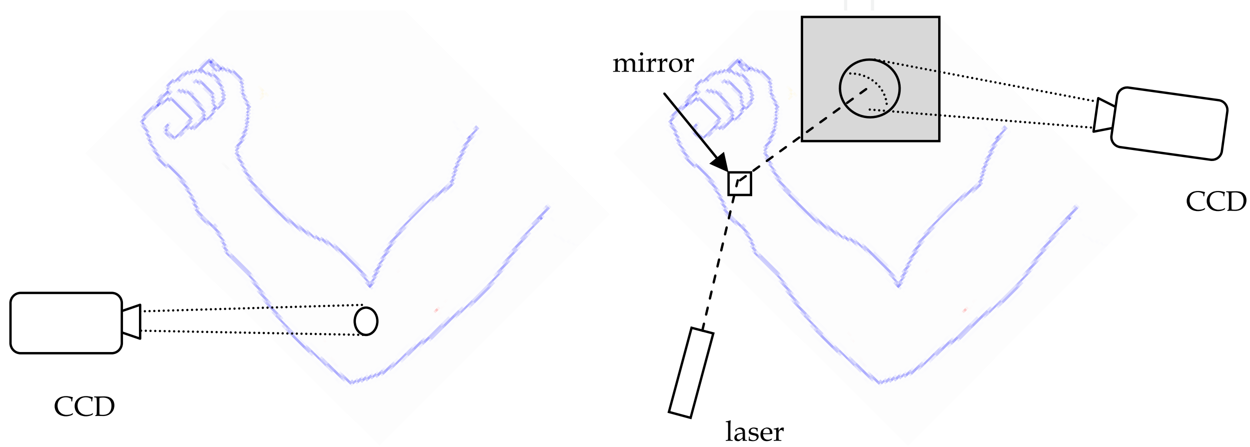


Fig. 20. Speckle image acquisition of skin surface deflections due to the systolic pressure wave (left) and laser reflectormetry of the mirror glued in correspondence of a large vase (right).

More recently (Takano & Ohta, 2007), it has been proposed a new device combining a time-lapse image from a handy video camera and image processing on a PC, and found that it could measure the 30s average heart and respiratory rates based on the changes in the brightness of the ROI set around the cheek of the unrestricted subject. Measurements were successfully conducted for subjects with or without facial cosmetics and that the system tended to detect the pulse rate more clearly around typical palpation points such as the common carotid artery and ulnar artery. Correlation coefficients of 0.90 has been obtained with respect heart rate measured with a pulse oxymeter.
One of the most recent and promising application of image-based techniques is the "cardiocam", as it has been named by its authors, which is a low-cost, non contact technology for measurement of physiological signals such as heart rate and breathing rate using a basic digital imaging device such as a webcam (Poh et al., 2010; Poh et al, 2011). The ability to perform remote measurements of vital signs is promising for enhancing the delivery of primary health care. It has been reported (Poh et al., 2011) Pearson's correlation coefficients among reference sensor and the web-camera-based system of r=1 for heart rate detection and of r=0.92 for HF and LF (common heart rate variability analysis parameters); the root-mean-squared error of the HR was 1.24 beats/min. In particular the system, developed at the Affective Computing research group at MIT laboratory, is suited for home applications and particularly for telemedicine, in fact the only sensing element utilised for heart rate monitoring is a standard web-camera which could be easily integrated on already

existing home furniture or home components such as mirrors or also utilising already existing camera such as the one integrated in most of the modern notebook or mobile phones. Despite these definitely promising advantage, still the robustness of the system and the efficiency of the procedure on large scale need to be demonstrated and improvement on the frame rate is required for adequate heart rate variability's analysis.
Finally another image-based system has been developed in collaboration with Philips Research Laboratory (Van Rooijen et al., 2010; Wieringa et al., 2005) mainly addressed for optical pletismography using non contact reflection mode imaging at three wavelengths. Respiration rate as well as heart rate mean values were very well correlated over a 50s window. A prototype of a small battery operated heart rate monitoring camera has been realised and demonstrated on professional swimmers for unrestrained heart rate measurement (Van Rooijen et al., 2010). Nevertheless at the moment further investigation need to be carried out for investigating the influence of factors such as movement artifacts, subcutaneous fat, skin thickness, skin pigmentation and blood pressure have not yet been investigated.

2.4 Other methods

An interesting application for non contact monitoring of heart (and lung) activity, has been demonstrated using magnetic induction (Steffen & Leonhardt, 2007). The physical principle demonstrated is that the mechanical actions of heart, diaphragm and thorax move blood and air through the body; by impedance point of view, this is a movement of well-conducting (blood) and poor conducting (air) matter inside the chest region. Inducing eddy currents into the tissue and measuring the re-inducted magnetic field externally, these impedance changes can be used as signals easily monitored. This is done using a simple arrangement based on one simple coil where excitation and measurement are performed. The proposed system assures a non contact monitoring (even if the coil needs to be placed sufficiently close to the thorax) and could be also integrated into textiles or integrated into bed. An high sensitivity to relative movements between coil and body is reported as the main limit of the technique.
The capacitive coupling method was also investigated for heartbeat detection through the subject's undergarments or an insulator (Kurita K, 2011). The potential of an electrode is measured against the body surface, thus, the capacitive coupling method enables detection of the human heartbeat under *in situ* conditions for a subject who is wearing clothing The capacitance is measured between the human body and an electrode that is isolated from the human body by a dielectric such as clothing. The experimental set-up used to demonstrate the principle is based on an electrode which is placed at a few centimeters from the subject' s chest, and the electrostatic induction current (on the order of pA) flowing through the electrode is measured. In such a scenario, we can express the induced current I flowing through the measurement electrode which is related to the electrostatically induced current generated because of the human heartbeat. The proposed method requires the use of a detection electrode (about 5 cm of radius) placed at 3 cm from the subject's chest. The method appears promising even if the sensitivity to external induction currents (common in non-laboratory and hospitals environments), the necessity to use filters and the very low amplitude of the sensing currents (<pA) appear to be the weak points of this method.
A different approach, reported in literature, is based on the pressure oscillations sensed from a sensing mat placed under a pillow (Zhu et al., 2006; Chen et al., 2005). The method is

here reported because it is a non contact method even if it requires the use of a system (pillow) through which the pulse rate (and the respiration rhythm) is detected. Real-time implementation is conducted using a pressure sensors sensing trough catheters an air-free water-filled vinyl tube under a pillow during sleep. Data reported are very sensitive to body movements and to sensor placement below the pillow, which makes the proposed method needing further improvement before the use on large scale.
A simpler approach is reported for measuring the physiological parameters, using an ultrasonic sensing of heart and respiration rate and movement behavior, developed for elderly people which are in bed. Bed structure oscillations due to the respiration and heart pulses are detected using an ultrasonic transmitter and an ultrasonic receiver installed in both sides of the bed rail (Mukai et al., 2009). When a person is lying in any position, on any bed area, then his/her physiological parameters alter the shape of the mattress and amplitude modulate the received signal. When the person is out of bed, the amplitude does not change, so the system also monitors in/out of bed status, allowing the detecting if the patient has left the bed, which makes it useful for use on subject monitoring (especially for elderly).

3. Final remarks

People are not used to wearing sensors in contact with their body in everyday life and for prolonged period of time, thus, in general, the compliance with sensors, electrodes and devices is very low. Nevertheless, there is a growing interest in continuous monitoring of the physiological activity of patients and non patients in order to gain information on their health status. The ideal solution would consist of using sensing systems that do not require physical contact with the user: the subject is not touched by a sensor at all. Accordingly, the sensing process could be unobtrusive to the user. Heart rate or respiration rate, for example, could be continuously monitored and data transmitted for remote signal analysis and feature extraction, allowing recognition of possible anomalous behavior or worsening of vital signs.
As an important part of the pervasive health care systems are the sensing units, the computation units and the communication units that are directly related with the implementation costs of these kind of systems, but also with the easy acceptance by the users. Referring to the sensing units, the non-invasive and the unobtrusive characteristics are important requirements taking into account the acceptance from the user. In the past, different solutions have been proposed for improving the acceptance of such systems: the use of non conductivity electrodes for ECG recording, where the electrodes are embedded on the chair, ballistocardiography sensing, where the sensing part is embedded in bed, chair or wheelchair, in furniture and photoplethismography using oxymeters. Sensing units, as part of ubiquitous systems, which work without mechanical and electrical contact with the subject represent a big challenge.
A possible field of application of new non contact method for heart rate monitoring is the heart rate variability (HRV) which has been shown to be a predictor of mortality after myocardial infarction and it is also associated with congestive heart failure, diabetic neuropathy, depression post-cardiac transplant, susceptibility to SIDS and poor survival in premature babies. HRV studies have also been used to examine autonomic function in the context of bodily pain. In recent years, the stress evaluation technique using the heart rate variability (HRV) has been recognized widely as an advanced diagnostic tool to prevent the

stress syndrome, for example, sleep sign while driving in a car, etc. HRV is typically carried out by means of long ECG analysis (more then 5/6 min).The HRV is obtained calculating RR intervals, however, evaluation of stress by ECG is not well suited for long-time monitoring due to the fact that several electrodes must be attached directly to a human body to acquire the ECG data. For this application non contact heart rate monitoring technique appear to be extremely interesting, especially for out of the hospital applications where application of electrodes are not feasible.

Another interesting field of application is represented by the need of HR monitoring of subjects during MRI; in such case, in fact, it is not recommended the presence of any metallic object which could create a serious risk for the patient (due to the presence of very high RF field). In such a case the laser-based as well as the image-based method reported are very encouraging avoiding heart rate (as well as respiration rate) monitoring without the need of metallic or conductive electrodes and reducing the possible interference effects.

A second important field of application where all the non contact heart rate monitoring technique could find important success is when the patient (or subject) can't be easily accesses (subject buried under building ruins), can't be safely accessed (contaminated subject) or it represents a risk for the operators to apply traditional monitoring procedures (such as the case of suspected victims). Remote sensing opportunities offered by laser-based, image based or electromagnetic-base techniques are extremely promising for those applications and have been demonstrated (at list in laboratory conditions) able to operate in such difficult conditions.

Nowadays, the electromagnetic-based systems are more advanced respect to the other non contact methods; the developments in the area of Doppler radar permit to develop unobtrusive cardiac and respiratory activity measurements that satisfied the above mentioned requirements. Doppler radars in continuous wave (CW) and pulsed radar (UWB) have, recently, been introduced to various applications, including home monitoring, research and rescue operations. As final remarks, it can be observed as the RF or MW based monitoring techniques appear to be among the most close to be ready for commercial distribution (first systems are already available, as seen in paragraph 2.1). Very interesting are the possibility to monitor from distance and to allow the monitoring also trough layers of different materials (tissues, clothes, wall, etc) which solve the problem of a direct access to the subject skin as it is for the laser-based methods. The aspects related to the exposition to electromagnetic fields are resolved considering the very low power density used for the cited applications (typically $< 1mW/cm^2$). Nevertheless, among the many papers published in this field, it appears necessary to act in a direction of a deep analysis of the phenomenon (interaction between RF wave and target, scattering from the target and the other environment objects, effect on the signal of the beam and of the antenna characteristics, etc.) using numerical simulation or developing specific codes in order to allow an optimal design of the RF monitoring system based not on empirical base, but on model based data of the many possible situations.

There are three benefits comparing MW sensing with ECG recording system. First, the subject will feel comfortable and relaxed without any physical and psychological burden, which the waveforms of the heartbeat will more objective. Second, the heartbeat can be detected at a distance of several meters by penetrating the clothing, thus the microwave sensor can be attached to the ceiling of a room or somewhere can not be seen by the subject. The procedure of the setup will be easier than the 3-lead ECG recording system, which needs the professional person to operate it. Finally, the microwave sensor can be easily integrated in unobtrusive healthcare systems.

However, two aspects limit electromagnetic-based systems the current use in ambulatory recording: e.m systems are susceptible to significant motion artifacts and the propagation loss in free space. Advanced signal processing represents a possible solution for the reduction of motion artifacts overlapped with the heartbeat signal, which might enable the sensors to be effectively used for long-term recording in mobile patients. Another aspect is that the propagation loss of the microwave sensor in free space should be taken into consideration during the recording and when the distance is increased, the transmitted power should be increased as well. Finally with the development of the integrated circuits, high gain and narrow side-lobe antenna in tiny size, the electromagnetic sensors can be made very small in size.

In reality, however, detecting human heartbeat at standoff via surface vibration monitoring poses a number of major technical hurdles. First of all, while physicians routinely monitor patient's heartbeat by accessing specific areas of the human body like the chest, carotid, and temples where surface displacement caused by heartbeat is most profound, it is to be noted that such accessibility can be very difficult, if not impossible, to achieve in situations where manipulation of the human subject's body is next to impossible for e.m. based systems, while feasible for laser-based methods. These situations include the inspection and evaluation of wounded, immobilized war fighters in the battlefield, real-time, standoff monitoring of astronauts operating inside space suits, as well as the monitoring of burn patients. The second technical challenge is the fact that while human body conducts acoustic vibrations relatively well, significant attenuation in the vibration strength excited by the heartbeat can be caused by the clothing, equipment, and other gears that the human subject might be wearing. As such, the surface vibrations that the human subject exhibits at the body parts accessible by the standoff interrogation technology can be greatly diminished, leading to the requirement of highly sensitive detection technologies in order to decipher such minute surface vibrations. Another technical hurdle relates specifically to laser based technologies which employ laser beams to conduct standoff monitoring of human subjects. Just like most of the objects we encounter in our daily lives, the surface quality of the human subject is far from optically flat, leading to the presence of optical speckles in the back-scattered light beam. Such optical speckles cause conventional coherent detection sensors to fail due to the random nature of phase fluctuations in the optical speckles, leading to erroneous output readings and mis-information. As human subject's body surface could be contaminated with sweat, grease, dirt, and even blood, it is safe to assume that abundant speckles exist in the laser light beam back-scattered from human subject's surface. Thus, whatever laser based technology is to be used to monitor the surface vibration of human subjects, it is vital that they be highly tolerant to the presence of speckles in the light beams so as to minimize the output of erroneous information in analyzing the cardiac functionality of the subject under standoff interrogation.

Image-based and the other methods reported appear to be promising for the simplicity of the hardware, the high degree of integration and the (probably) final cost, even if they need to be more deeply investigated in order to demonstrate they feasibility.

Finally it is important to evaluate the possible scenarios for the future use of noncontact electromagnetic system for the detection of heart rate. With several potential medical applications and ready-to-launch UWB radar based monitoring products, the medical device industry is likely to observe significant changes. One of the biggest advantages for the UWB based medical sensing products is the low cost associated with them (this is not still to be proven for the laser-based systems as well as fort the others techniques here proposed). With budgetary concerns being a perennial challenge for both the European and US medical device markets, these devices are likely to find ready acceptance. However, whether these technologies shall gradually cannibalize the conventional vital signs monitoring devices market is yet to be ascertained. Most of These technologies are likely to target unconventional end user groups (including home care, low acuity areas in hospitals and the military sector) instead of the conventional high-end hospital market and this seems to prospect the possibility to create a niche market segment for itself instead of penetrating into the conventional markets. Amongst all the UWB based medical devices, it is likely that the vital signs monitoring device to hit the market first. The progress of this niche market segment shall be observed carefully by other global vendors selling vital signs and patient monitoring devices. It is likely that many of the new entrants could get acquired by the leading patient monitoring vendors who do not have a significant presence in the home care or low acuity care market segments. This will help them diversify to new markets and ensure that their existing market shares are sustained, if not increased.

4. References

Abowd G., Bobick A., Essa I., Mynatt E., and Rogers W (2002). The Aware Home: Developing Technologies for Successful Aging. In *AAAI Workshop and Automation as Care giver*.

Alexander CM, Teller LE, Gross JB (1989). Principles of pulse oximetry: theoretical and practical considerations. *Anesth Analg*.;68:368-376.

Aubert AE, Welkenhuysen L, Montald J, de Wolf L, Geivers H, Minten J, Kesteloot H, Geest H (1984). Laser method for recording displacement of the heart and chest wall, *J Biom Eng*, (2), 134-40.

Azevedo SG, McEwan TE (1996), Micropower Impulse Radar. *Science and Technology Review*, UCRL-52000-96-1/2.

Bocconcelli P, Pierantozzi A, Scalise L, Morbiducci U, De Melis M, Corbucci G (2006). Una nuova tecnologia senza per la valutazione della meccanica cardiaca nei pazienti scompensati: la vibrometria laser. 6° Congresso Nazionale Associazione Italiana Aritmologia e Cardiostimolazione (AIAC), 30 Marzo - 1 Aprile 2006.

Castellini P, Martarelli M, Tomasini EP (2006). Laser Doppler Vibrometry: Development of advanced solutions answering to technology's needs. *Mech Syst Sig Proc*, 20(6):1265-1285.

Chan KH, Lin JC. (1987). Microprocessor-based cardiopulmonary rate monitor. *Medical and Biological Engineering and Computing*, 25(1):41-44.

Chen M, O'Sullivan JA, Singla N, Sirevaag EJ, Kristjansson SD, Lai PH, Kaplan AD, Rohrbaugh JW (2010). Laser Doppler vibrometry measures of physiological function: Evaluation of biometric capabilities. *IEEE Transactions on Information Forensics and Security*, 5(3):449-60,.

Chen W, Zhu X, Nemoto T, Kanemitsu Y, Kitamura K, Yamakoshi K (2005). Unconstrained detection of respiration rhythm and pulse rate with one under-pillow sensor during sleep. *Med Biol Eng Comp*, 43(2):306–312.

Chia MY, Leong SW, Sim CK, Chan KM (2005). Through-wall UWB radar operating within FCC's mask for sensing heart beat and breathing rate. *IEEE European Microwave Conf. 2005*, 3(14).

Da Costa (1995). Optical remote sensing of heartbeats. *Optics Communications*, 117(5-6):395-98.

Daisuke Nagae and Atsushi Mase(2010). Measurement of heart rate variability and stress evaluation by using microwave reflectometric vital signal sensing. *Rev Sci Instrum* 81, 094301.

De Melis M, Morbiducci U, Scalise L, Tomasini EP, Delbeke D, Baets R, Van Bortel LM, Segers P (2008). A non contact approach for the evaluation of large artery stiffness: a preliminary study., *Am J Hyp*, 21:1280-83.

Dressier W (1970)., *Clinical aids in cardiac diagnosis* . Grune & Stratton.

Droitcour A, Lubecke V, Lin J, Boric-Lubecke O(1995). A microwave radio for Doppler radar sensing of vital signs, *IEEE MTT-S Int. Microw. Symp. Dig.*, 175.

Durand LG, Pibarot P (1995). Digital signal processing of the phonocardiogram: review of the most recent advancements. *Crit Rev Biomed Eng*.

Einthoven W (1902). Galvanometrische registratie van het menselijk electrocardiogram [Galvanometric registration of the human electrogram]. In: Herinneringsbundel Prof. S.S. Rosenstern Leiden, the Netherlands: Eduard Ijdo; 1902; 101–106.

Fowler NO (1962). *Physical diagnosis of heart disease*. Macmillan, New York.

Garbey M, Sun N, Merla A and Pavlidis I (2007). Contact-free measurement of cardiac pulse based on the analysis of thermal imagery, *IEEE Trans. Biomed. Eng*. 54(8), 1418–1426.

Gorgas, DL (2004). *Vital signs and patient monitoring techniques*, Clinical Procedures in Emergency Medicine 4th ed (JR Roberts and JR Hedges, Eds) Sauders, Philadelphia, USA.

Gorgels APM (2007). *Chapter 3 Electrocardiography*. In: Willerson JT, Cohn JN, Wellens HJJ, Holmes DR Jr, eds. Cardiovascular Medicine, 3d ed. Springer, London, UK.

Harland CJ, Clark TD and Prance RJ (2002). Electric potential probes-new directions in the remote sensing of the human body, *Meas. Sci. Technol.*, 13:163–169.

Hein JJ, Wellens MD, Gorgels AP (2004). The Electrocardiogram 102 Years After Einthoven. *Circulation*, 109:562-64.

Hong H and Fox M (1997). Noninvasive detection of cardiovascular pulsations by optical doppler techniques. *J of Biomedical Optics*, 2(4):382-390.
http://lifewaveinc.com/
http://www.microimpulse.no
http://www.wireless2000.com

Immoreev I, Tao TH (2008). UWB radar for patient monitoring, *Aerospace and Electronic Systems Magazine*, 11(18).

Ishijima M (1993). Monitoring of electrocardiograms in bed without utilizing body surface electrodes, *IEEE Trans. Biomed. Eng.*, 40(6):593–594.

Kim HJ, Kim KH, Hong YS and Choi Jj (2007). Measurement of human heartbeat and respiration signals using phase detection radar, *Rev Sci Instrum*, 78, 04703-1.

Kurita K (2011). Note: Human heartbeat measurement on the basis of current generated by electrostatic induction. *Rev Sci Inst* 82, 026105.

Lim YG, Kim KK and Park KS (2006). ECG measurement on a chair without conductive contact", *IEEE Trans. Biomed. Eng* ,53:956.

Lim YG, Ko Keun Park, Kwang Suk (2007).ECG Recording on a Bed During Sleep Without Direct Skin-Contact, *IEEE Transactions on Biomedical Engineering*, 54(4):718.

Lin, J. C. (1975). Noninvasive microwave measurement of respiration. *Proc. IEEE*, p. 1430.

Lin J C, Kiernicki J, Kiernicki M, Wollschlaeger PB (1979). Microwave Apexcardiography. *IEEE Transactions on Microwave Theory and Techniques*, 27(6): 618-620.

Lu G, Yang F, Tian Y, Jing X, Wang J (2009). Contact-free Measurement of Heart Rate Variability via a Microwave Sensor. *Sensors.* 9(12):9572-9581.

Matsuda, Toshiyuki, Makikawa, Masaaki (2008). ECG monitoring of a car driver using capacitively-coupled electrodes, *Engineering in Medicine and Biology Society, 2008. EMBS 2008. 30th Annual International Conference of the IEEE,*: 1315 - 1318, 20-25 Aug. 2008

Matsui T, Hagisawa K, Ishizuka T et al (2004) A novel method to prevent secondary exposure of medical and rescue personnel to toxic materials under biochemical hazard conditions using microwave radar and infrared thermography. *IEEE Trans Biomed Eng* 51:2184–2188.

Matsui T, Ishizuka T, Takase B et al (2004) Non contact determination of vital sign alterations in hypovolemic states induced by massive hemorrhage: an experimental attempt to monitor the condition of injured persons behind barriers or under disaster rubble. *Med Biol Eng Comput* 42:807–811

Mei C, O'Sullivan JA, Singla N., Sirevaag EJ, Kristjansson SD, Po-Hsiang L, Kaplan AD, Rohrbaugh JW(2010). Laser Doppler Vibrometry Measures of Physiological Function: Evaluation of Biometric Capabilities. *IEEE Transactions on Information Forensics and Security archive*, 5(3).

Morbiducci U, Scalise L, De Melis M, Grigioni M (2006). Optical vibrocardiography: a novel tool for optical monitoring of cardiac activity. *Annals of Biomedical Engineering* 35(1):45-58.

Morris, G. M. (1954). A review of ballistocardiography. *Journal of the American Medical Womens Association* 1972, 9(12), 391-393.

Mukai K, Yonezawa Y, Ogawa H, Maki H and Caldwell MW (2009). A remote monitor of bed patient cardiac vibration, respiration and movement. *31st Ann Int Conf IEEE EMBS* Minneapolis, Minnesota, USA, September 2-6.

Parra, JE, Da Costa, G (2001). Optical remote sensing of heartbeats. *Proceedings of SPIE - The International Society for Optical Engineering*, 4368, 113-121.

Paulson CN, Chang JT, Romero CE, Watson J, Pearce FJ, Levin N (2005). Ultra-wideband Radar Methods and Techniques of Medical Sensing and Imaging. *SPIE International Symposium on Optics East* Boston, USA, Oct. 23-25.

Poh, MZ, McDuff, DJ, Picard, RW (2011). Advancements in Non contact, Multiparameter Physiological Measurements Using a Webcam, *IEEE Transactions on Biomedical Engineering*, 58(1):7-11.

Poh, M.Z., McDuff, D.J., Picard, R.W. (2010). Non contact, Automated Cardiac Pulse Measurements Using Video Imaging and Blind Source Separation, *Optics Express*, 18(10):10762-10774.

Poliac MO, Zanetti J, Salerno DM, Wilcox GL (1991). Seismocardiogram (SCG) interpretation using neural networks. *4th IEEE Symp Computer-based Medical Systems*.

Pollock, P (1957). Ballistocardiography: A Clinical Review. *Canadian Medical Association Journal*, 76(9), 778-783.

Ramachandran G, Singh M (1989). Three-dimensional reconstruction of cardiac displacement patterns on the chest wall during the P, QRS and T-segments of the ECG by laser speckle interferometry. *Med Biol Eng* 27(5).

Rohrbaugh JW et al. (2007). The Physiology of Threat: Remote Assessment Using Laser Doppler Vibrometry . *The Journal of Credibility Assessment, and Witness Psychology*, 7(2):17-19.

Salerno DM, Zanetti J (1990). Seismocardiography: A new technique for recording cardiac vibration. Concept, method and initial observations. *J Cardiov Tech*, 9:2.

Sandham W et al. (1998). Multiresolution wavelet decomposition of the seismocardiogram. *IEEE Tr Signal Processing*, 46:9.

Scalise L. et al. (2005a). Optical monitoring of heart rate variability. *Laser in Medical Scien*ce, 20, s25.

Scalise L et al. (2005b). Optical monitoring of heart beat. In *Modelling in Medicine and Biology*, WIT press, 181-190.

Scalise L et al. (2006)A laser Doppler approach to cardiac motion monitoring: Effects of surface and measurement position. *Proc SPIE 7th Int Conf on Vibration Measurements by Laser Techniques: Advances and Applications*, Ancona.

Scalise L , Morbiducci U (2008). Non contact monitoring from carotid artery using optical vibrocardiography. *Med Eng & Phy*, 30(4): 490-97.

Scalise L, Ercoli I, Marchionni P, Tomasini E P (2011). Measurement of Respiration Rate in Preterm Infants by Laser Doppler Vibrometry. *Proc. 2011 IEEE International Symposium on Medical Measurements and Applications (MeMeA 2011)*, 418-422.

Staderini E M. (2002). UWB radars in medicine. *IEEE Aerosp. Electron. Syst. Mag.*, p.13.

Steffen M, Aleksandrowicz A, Leonhardt S (2007). Mobile Noncontact Monitoring of Heart and Lung Activity. *IEEE Transactions on Biomedical Circuits and Systems*, 4(1):250-57.

Takano C, Y Ohta (2007). Heart rate measurement based on a time-lapse image. *Med Eng & Phy*, 29(8):853-857.

Tanaka S, Matsumoto Y, Wakimoto K (2002) Unconstrained and non-invasive measurement of heart-beat and respiration periods using a phonocardiographic sensor. *Med Biol Eng Comput* 40:246–252

Tateishi N., Mase A., Bruskin L., Kogi Y., Ito N., Shirakata T., and Yoshida S. (2007). Microwave Measurement of Heart Beat and Analysis Using Wavelet Transform. *Proceedings of the 2007 Asia Pacific Microwave Conference*, Bangkok, 2007, pp. 2151–2153.

Tomasini EP et al. (1998). Carotid artery pulse wave measured by a laser vibrometer. *Proc SPIE of the 3rd Int Conf on Vibration Measurements by Laser Techniques: Advances and Applications (3411*), Ancona, 1998;: 611-616.

Van Rooijen V, de Voogd-Claessen L, Lauche K, Jeanne V, van der Vliet R (2010). Development of a new product for unrestrained heart rate measurement in swimming: a user centered design approach. *Procedia Engineering* , vol. 2, no. 2, pp. 2693-2699.

Wahr JA, Tremper KK, Diab M (1995). Pulse oximetry. *Respiratory Care Clinics of North America*.1:77-105.

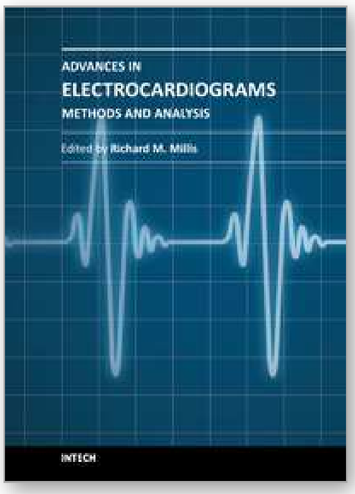
Wang C,Trivedi S, Rodriguez P, Jin F,Kutcher S, Swaminathan V, Nagaraj S, Quoraishee S, Prasad N (2010). Non contact Human Cardiac Functionality Assessment Using a High Sensitivity Pulsed Laser Vibrometer. Proceedings of *27th Annual Army Science Conference*, Orlando, Fla., Nov. 29 - Dec. 2. , 2010.

Webster J.B. (1988). *Encyclopedia of medical devices and instrumentation*, Vol. 4, Wiley, New York, USA.

Welch JP, DeCesare R, Hess D (1990). Pulse oximetry: instrumentation and clinical applications. *Respiratory Care*. 35:584-601.

Wieringa FP, Mastik F and van der Steen AFW (2005). Contactless Multiple Wavelength Photoplethysmographic Imaging: A First Step Toward SpO2 Camera Technology. *Annals of Biomedical Engineering* 33(8), 1034-1041.

Zhu X, Chen W, Nemoto T, Kanemitsu Y, Kitamura K, Yamakoshi K and Wei D (2006). Real-Time Monitoring of Respiration Rhythm and Pulse Rate During Sleep. *IEEE Transactions on Biomedical Engineering*, 53(12):2553-2564.



Advances in Electrocardiograms - Methods and Analysis

Edited by PhD. Richard Millis

ISBN 978-953-307-923-3

Hard cover, 390 pages

Publisher InTech

Published online 25, January, 2012

Published in print edition January, 2012

Electrocardiograms are one of the most widely used methods for evaluating the structure-function relationships of the heart in health and disease. This book is the first of two volumes which reviews recent advancements in electrocardiography. This volume lays the groundwork for understanding the technical aspects of these advancements. The five sections of this volume, Cardiac Anatomy, ECG Technique, ECG Features, Heart Rate Variability and ECG Data Management, provide comprehensive reviews of advancements in the technical and analytical methods for interpreting and evaluating electrocardiograms. This volume is complemented with anatomical diagrams, electrocardiogram recordings, flow diagrams and algorithms which demonstrate the most modern principles of electrocardiography. The chapters which form this volume describe how the technical impediments inherent to instrument-patient interfacing, recording and interpreting variations in electrocardiogram time intervals and morphologies, as well as electrocardiogram data sharing have been effectively overcome. The advent of novel detection, filtering and testing devices are described. Foremost, among these devices are innovative algorithms for automating the evaluation of electrocardiograms.