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Non-invasive Electronic Biosensor Circuits and Systems

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1. Introduction

An aging population has lead to increased demand for health-care and an interest in moving health care services from the hospital to the home to reduce the burden on society. One enabling technology is comfortable monitoring and sensing of bio-signals. Sensors can be embedded in objects that people interact with daily such as a computer, chair, bed, toilet, car, telephone or any portable personal electronic device. Moreover, the relatively recent and wide availability of microelectronics that provide the capabilities of embedded software, open access wireless protocols and long battery life has led many research groups to develop wearable, wireless bio-sensor systems that are worn on the body and integrated into clothing. These systems are capable of interaction with other devices that are nowadays commonly in our possession such as a mobile phone, laptop, PDA or smart multifunctional MP3 player. The development of systems for wireless bio-medical long term monitoring is leading to personal monitoring, not just for medical reasons, but also for enhancing personal awareness and monitoring self-performance, as with sports-monitoring for athletes. These developments also provide a foundation for the Brain Computer Interface (BCI) that aims to directly monitor brain signals in order to control or manipulate external objects. This provides a new communication channel to the brain that does not require activation of muscles and nerves. This innovative and exciting research field is in need of reliable and easy to use long term recording systems (EEG).
In particular we highlight the development and broad applications of our own circuits for wearable bio-potential sensor systems enabled by the use of an amplifier circuit with sufficiently high impedance to allow the use of passive dry electrodes which overcome the significant barrier of gel based contacts.

2. Advantages of biomedical signals long term monitoring

Monitoring of patients for long periods during their normal daily activities can be essential for the management of various pathologies. It can reduce hospitalization, improve patients'

quality of life, and help in diagnosis and identification of diseases. Long-term monitoring of activities can also be useful in the management of elderly people. Moreover, the combination of biomedical signals and motion signals allows estimation of energy expenditure (Gargiulo, Bifulco et al. 2008); (Strath, Brage et al. 2005). Hence, it could also enable the monitoring of human performance (e.g. athletes, scuba divers) in particular conditions and/or environments.

To accomplish these tasks the monitoring equipment will have to comply with some specific requirements such as: portability and/or wearability, low power, long lasting electrodes, data integrity and security, and compliance with medical devices regulation (e.g. electrical safety, electromagnetic compatibility) (Lin, Jan et al. 2004).

2.1 Long term of cardiac signals

Cardiology is one branch of medical science that could clearly benefit from long-term monitoring. It is well known that morphological changes or the presence of various arrhythmias in the long term electrocardiogram (ECG) have a strong correlation with heart and coronary artery diseases (Zheng, Croft et al. 2002). Also, the reoccurrence of atrial fibrillations after ablation is not uncommon and these can only be tracked using long term ECG monitoring (Hindricks, Piorkowsky et al. 2005).

Long term ECG monitoring in cardiology is not only useful for follow up of patients where their pathological status is already known, but also for the monitoring of athletes during exercise. The possibility that young, highly trained or even professional athletes may harbor potentially lethal heart disease or be susceptible to sudden death under a variety of circumstances seems counterintuitive. Nevertheless, such sudden cardiac catastrophes continue to occur, usually in the absence of prior symptoms, and they have a considerable emotional and social impact on the community (Basilico 1999). As a result of the ECG screening programs for athletes which are now compulsory in many countries, it is now known that many of these sudden deaths are due to a syndrome called "Athlete's heart". This syndrome may be associated with rhythm and conduction alterations, morphological changes of the QRS complex in the ECG, and re-polarization abnormalities resembling pathological ECG (Fagard 2003). However, it is broadly accepted that the standard 12 lead ambulatory ECG is not reliable enough during movement to clarify the origin of the ECG alteration, especially if this is triggered by the exercise(Kaiser & Findeis 1999). This makes a system that is able to record the ECG during exercise reliably and without interference desirable.

For standard ECG measurements electrodes are attached to the patient's skin after skin preparation, which includes cleaning, shaving, mechanical abrasion to remove dead skin, and moistening. A layer of electrically conductive gel is applied in between the skin and the electrodes to reduce the contact impedance (J. G. Webster 1998). However, in these so-called wet electrodes the electrolytic gel dehydrates over time which reduces the quality of the recorded signals. In addition, the gel might leak, particularly when an athlete is sweating, which could electrically short the recording sites. This is an even larger problem for monitoring athletes immersed in water. Securing the wet electrodes in place is also complicated, since the electrodes cannot directly be glued to the skin due to the presence of the gel. The use of dry or insulating electrodes may avoids or reduce these problems (Searle & Kirkup 2000).

2.3 Physical activity monitoring

There are many techniques to monitor human motion from self-reporting surveys, accelerometers, pedometers to constant video monitoring. Clinically it is interesting to measure gait, posture, rehabilitation from suffers of neurological conditions such as stroke(Uswatte, Foo et al. 2005), tremors associated with Parkinson's' disease and sleep(Mathie, Coster et al. 2004). However the most common aim for long term monitoring is to assess energy expenditure in physical activity due to its positive effects on health, decrease in mortality rates and aid with chronic diseases such as hypertension, diabetes and obesity(Murphy 2009). The gold standard measurement for energy expenditure is doubly labeled water which requires the ingestion of expensive water labeled with a non-radioactive isotope and the expensive and time consuming sampling of fluids such as blood, urine or saliva.

Accelerometry is becoming the widely accepted tool for assessment of human motion in clinical settings and free living environments as it has the following advantages: simple based on a mass spring system, low cost, small, light, unobtrusive, and reliable in the long term and for unsupervised measurements such as in the home. The most commonly used accelerometers for human movement are piezo-electric sensors that measure acceleration due to movement. They are also sensitive to gravitational acceleration which needs to be subtracted. They are normally manufactured using MEMs technology resulting in miniature, low cost and reliable devices. A tri-axial accelerometer can measure acceleration in three orthogonal dimensions and is able to describe movement in three directions. The use of solid state memories enables long term recording with commercial devices able to continuously record 1-minute epochs for longer than a year(Murphy 2009).

Home use is preferred to clinic studies to reflect normal functional ability of the subject. Activity monitoring with tri-axial accelerometers in a free living environment has been shown to correlate well with the gold standard(Hoos, Plasqui et al. 2003). Accelerometers also show little variation over time (drift) and can be easily recalibrated by tilting in gravitational field. They respond quickly to frequency and intensity of movement and are found to be better than pedometers which are attenuated by impact or tilt(Mathie, Coster et al. 2004).

Their main disadvantage is position dependence when whole body movement is desired. The common approach is to locate the sensor at centre of gravity (such as the waist or pelvis of a human subject) or for improved accuracy, locate many sensors in various positions over the body. Accelerometers are also sensitive to static position changes and movement. Most human movements are in the frequency band of 0.3 to 3.5Hz so most systems use a high pass filter with cut-off of 0.1 to 0.5 Hz to separate static orientation and body movement. As with any free-environment measurement, compliance is an issue as the data may not be used if the subject chooses or forgets to wear the sensor(Mathie, Coster et al. 2004). Some accelerometers measure the stationary tremor of the human body while others use skin conductance to detect when the sensor is being worn. It is also common to use signal processing to estimate compliance and remove non-compliant data segments (Murphy 2009).

To determine metabolic activity from accelerometer measurements various empirical models are have been proposed some of which rely on measurement of other variables such as mass, sex, age. However good correlations have been found with consumed oxygen in various populations with a model based solely on accelerometer counts (Pate, Almeida et al.

2006). Different activities such as running, walking, up-down stairs, cycling can be determined from accelerometer measurements but there is variability with accuracy ranging from 0.89 in house bound subjects to 0.59 for those in a free-living environment (Mathie, Coster et al. 2004). New technologies, including the combination of accelerometry with the measurement of physiological parameters, have great potential for the increased accuracy of physical-activity assessment (Corder, Brage et al. 2007).

2.2 Long term of brain signals

Long term monitoring of brain signals is used in neurology, cognitive science and psycho-physiological research. Its use in clinical EEG recordings improves diagnostic value by up to 90% (Logar, Walzl et al. 1994) One of its extended uses is in neuro-feedback or brain computer interfaces where the brain signals are interpreted as controls for a computer system (Gargiulo, Bifulco et al. 2008).

The clinical motivation to record brain signals long term has traditionally been to observe the Electroencephalogram (EEG) to aid in epilepsy or sleep studies. Epilepsy is an underlying tendency of the brain to produce sudden bursts of electrical activity that disrupt other brain function or a seizure. It is estimated that 10% will experience a seizure during their lifetime with 1% diagnosed with epilepsy. Seizures are variable in severity, frequency and the affected region of the brain and so difficult to diagnose (Waterhouse 2003).

Standard electrodes are small Ag/AgCl discs applied with conductive paste or gel to improve the conductivity of the contact. Collodium glue is often used in long term recording to ensure contact. The international 10-20 EEG electrode placement uses 30 electrodes arranged approximately in two concentric rings around the head and bio-potentials are recorded differentially using high gain, high input impedance FET input amplifiers usually arranged in an instrumentation amplifier circuit (J. G. Webster 1998). In modern systems the differential recording is converted using a high resolution analogue to digital converter so that difference voltages can be selected in computer software. This flexibility is important as clinicians tend to view the recordings as bipolar differences between sets of electrodes or differences from the reference or average of all electrodes. These are commonly referred to as montages(Waterhouse 2003).

EEG potentials recorded from electrodes placed on the scalp represent the collective summation of changes in the extracellular potentials of pyramidal cells. These are the most prevalent and largest cells in the cerebral cortex and are arranged in columns causing their activation currents to add. The resulting voltage is attenuated by about 10x by volume conduction through the tissues of the head: cerebrospinal fluid, skull, scalp and skin(J. G. Webster 1998).

The EEG signal normally ranges from 10 to 150uV and are commonly categorized in frequency bands which can indicate brain states and pathology depending on where they are recorded on the scalp: Delta 0.1-3.5 Hz, Theta 4-7.5 Hz, Alpha 8-13 Hz and Beta 14-22 Hz((Ed.) 2006). Exemplificative interpretations of such waveforms are: epilepsy seizure on-set (increased presence of 3 Hz spikes), the alpha wave replacement phenomenon, evoked potentials and the Mu-rhythm commonly used for BCI (Brain Computer Interface) applications.

Alpha wave replacement phenomenon is easy to elicit. As one of the most studied elicited mental states, it is also commonly used in clinical practice to ensure EEG setup validity.

It is well known that the closing of both eyelids in a relaxed subject is followed by alpha wave replacement in the EEG. In awake relaxed subjects the phenomenon presents a visible increase in the magnitude of alpha waves (8-13 Hz) that starts after the closing of both eyelids and stops with the opening of the eyes (J. G. Webster 1998).

The phenomenon is more apparent in the frequency domain as shown Figure 1. Observe the difference in the spectrum around 9 Hz between the eyes open (bold) and eyes closed cases (grey) as recorded by two electrodes placed on the scalp of a volunteer subject.
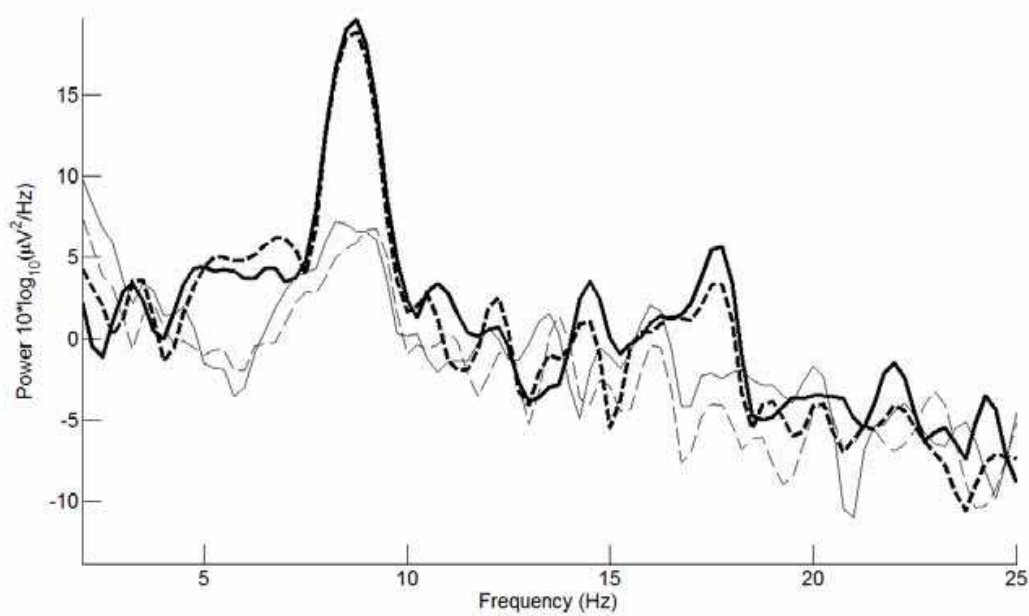
Fig. 1. Power spectral density showing alpha wave replacement.

Long term monitoring of EEG signals might also provide advantages for Brain Computer Interfaces (BCIs). An EEG based Brain-Computer Interface system seeks direct interaction between the human brain and machines, aiming to augment human capabilities by enabling people (especially those who are disabled) to communicate and control devices by merely "thinking" or expressing intent. Therefore, it is possible to say that the main aim of BCI researchers is to build a new communication pathway for the human brain that does not depends from its standard output channels such as nerves and muscles. (Millan, Renkens et al. 2004; Pfurtscheller, Brunner et al. 2006)

Such systems can be realized in two ways: in externally (stimulus)-paced mode (synchronous BCI) or in an internally paced mode (asynchronous BCI). Synchronous BCI requires that the subject achieves a specific mental state in response to an external event, within a predefined time window, whereas in asynchronous BCI is not required any time window constraint so the subject is free to intend a mental state or a specific thought. (Pfurtsheller & Neuper 2001) However, it is possible to say that both methodologies make use of classified EEG signals epochs,. Synchronous BCIs make use of oscillatory EEG activity (Pfurtscheller, Brunner et al. 2006) and slow cortical potential shifts (Hinterbergera, Küblera et al. 2003), while for asynchronous BCIs, various types of event-related potentials are used (Millán 2003).

Focusing on synchronous BCI, two types of oscillation seems to be the more usable: the Rolandic mu rhythm in the range 7–13 Hz and the central beta rhythm above 13 Hz, both

originating in the sensorimotor cortex area, these phenomenon (known from the early '50 (Chatrian, Petersen et al. 1959)) are not only linked to the voluntary motor intentions, but as many recent studies confirm, they are linked to the mental imagination of movements. It has been shown that motor imagination involves similar brain regions/functions which are involved in programming and preparing such movement. (Jeannerod 1995)

Routine clinical EEG recordings are brief, typically a 20 minute recording, and unlikely to catch a sporadic seizure. They are also set in artificial environments such as a clinic. So ambulatory EEG: continuous recording over 72 hours has arisen with availability of solid state memories for easily storing data(Waterhouse 2003).

These allow long term brain signal recordings in a free- environment such as home use which is more convenient, less expensive, familiar environment. The extended recording time increases the chance to catch a seizure and recording time length correlates with number of seizures measured(Logar, Walzl et al. 1994). These have been found to be clinically useful in 75% of subjects with low false positives (0.7% in asymptomatic adults), they can detect normally up to 63% additional seizures over conventional recordings and can be used to validate therapy by showing a change from abnormal to normal EEG with treatment (Waterhouse 2003). They have been adopted for sleep monitoring and combined with EMG, ECG, respiration, oximetry to measure other sleep signals of interest.

3. Passive dry electrodes

As should be clear from what has been explained until now, it is urgent to find a solution to the number of issues raised by the use of conventional gel-based electrodes and skin preparations the so called 'wet electrodes' in biomedical signal recording. As mentioned a possible solution could be the use of the so called 'passive dry electrodes',(Gargiulo, Bifulco et al. 2008; Gargiulo, Bifulco et al. 2008; Bifulco, Fratini et al. 2009). These are distinct from active dry electrodes that require local active electronics, power supply over cables, additional manufacturing and hermetic enclosures. Passive dry electrodes have no local active electronics so it is possible to integrate them in garments or clothes resulting less obstruction in daily life activities.

A material with several particular attributes for dry electrodes is conductive rubber. Electrodes made from such material are durable, washable and re-usable, the carbon and silicon materials (commonly used in conductive rubber) are biocompatible, they provide a smoother and more uniform contact surface with the skin, they also can be thin, flexible and easily applied to a variety of substrates (Muhlsteff & Such 2004; Chang, Ryu et al. 2005). However, they also present much higher impedance with respect to conventional wet electrodes (Baba & Burke 2008).

Alternatively, textile electrodes, which can be embedded in clothes, can also be used as dry electrodes. These offer a high degree of patient autonomy and freedom of movement and are suitable for long term monitoring. Textile electrodes are typically made of synthetic materials which endure abrasion very well, do not irritate the skin, and are lightweight and washable. A major drawback is the high contact impedance of these electrodes, typically 1-5 MΩ/cm^2, compared to a 10 kΩ/cm^2 impedance for the disposable Ag/AgCl electrodes (Catrysse, Puers et al. 2003).

The high electrode impedance is a common issue with all types of dry electrodes. Therefore, to use these electrodes they must be connected to instrumentation amplifiers with extremely high input impedances (Prutchi & Norris 2005).

3.1 Active dry electrodes

Another possible solution to the entire above mentioned problem largely described in literature for many years is to buffer the electrode signal directly at the electrode to provide impedance conversion (Taheri, Knight et al. 1994; Harland, Clark et al. 2002; Valchinov & Pallikarakis 2004); this type of assembly is referred to as an active electrode. However, this approach exhibits a number of weaknesses. Firstly it is possible to observe that some rigid circuitry is fixed on the electrode, increasing the size of the electrode and resulting in an inflexible electrode that cannot be embedded in clothes e.g. Textile electrodes. Moreover, despite the various methodologies attempted to apply the electrode to the skin, particularly in EEG applications, dry electrodes signals are affected from large movement artifacts due to skin/electrodes stretch (Talhouet & Webster 1996) and electrode skidding that contributes to increases in the contact impedance imbalance. In "wet" electrodes this is mitigated by skin preparation and conductive paste interposition.

3.2 Physical activity monitoring

To monitor physical activity during daily life is not a trivial problem to solve. Current clinical assessments of physical activity monitoring and body energy expenditure are based on the evaluation of several parameters such as ECG, body temperature and oxygen consumptions. However, new perspective in physical monitoring are offered by novel M.E.M.S. (Micro Electro Machined Sensors) accelerometer sensors. Such kind of sensor is based on the changes in the value of a capacitor given geometrical variations. Often one plate of the capacitor is kept fixed whilst the other one is free to vary with the stimuli (in this case the acceleration). Nowadays such sensors are tiny, cheap, light weight and more important they are offered tri-axial and sensitive to the statically gravitational acceleration, so they are useful for orientation and position assessment in the gravitational field. Their use in daily life devices (such as gaming console or mobile phone and daily home-care (Freescale Semiconductor 2005)) is increasing and there are many examples in the available literature of their use as physical activity monitoring device.

Experimental evidence has shown that there are clear different pattern recordable by accelerometer sensors during different tasks and movements e.g. it is possible to distinguish between slow normal or fast walking simply looking at the signal magnitude end or its second derivate (Mühlsteff, Such et al. 2004). Moreover, rapid gradient changes and fast transients in the signal are useful for posture assessment and free fall recognition that are clearly useful in tele-medicine and home-care (Strath, Brage et al. 2005; Bifulco, Gargiulo et al. 2007; Giansanti 2007).

Measuring the acceleration, theoretically would be possible to calculate the velocity and then the position as function of the time, however, in order to calculate velocity and time two initial condition are needed. Furthermore, more than one sensor is needed to monitor anatomical segments such as arms and legs.

3.3 State of the art wearable biosensor systems

Many groups worldwide are developing wearable wireless biosensor systems for applications ranging from health care, athlete monitoring and vital signs in high risk environments. Most report on systems aimed at a particular application and these systems usually measure one or two physiological signals only (Pandian, Mohanavelu et al. 2008). Other groups are studying optimal network methods for body sensor networks (BSNs),

body area networks or personal area networks (Hao & Foster 2008). The aims here are to develop a miniature, low power nodes, several of which are worn on the body and each capable of sampling, processing and communicating physiological or environmental measurements. In wireless BSNs the nodes wirelessly communicate to a central hub, removing the need for cables between sensors. While this improves long term monitoring from a comfort and practicality perspective, it does introduce challenging power and size requirements on the nodes.

A more immediate concern for long term monitoring of surface bio-potentials is the current need to use conductive gel. The "smart vest" introduced in (Pandian, Mohanavelu et al. 2008) is capable of monitoring a number of vital parameters in a single device, including ECG without the use of gel and cuff-less non-invasive blood pressure derived from the ECG and PPG. The vest is made of cotton or lycra with sensors embedded as to make good contact with the skin. The ECG sensor is custom made consisting of silicon rubber with pure silver filings and worn in the form of belts to acquire standard lead-II ECG. We would consider this design as a state of the art long term monitoring system, however it has not been combined with a physical activity monitor or accelerometer which is important in applications such as athlete monitoring.

4. A new wireless system based on passive dry electrodes

Our present research is focused on the use of passive dry electrodes mainly realized with conductive rubber. Hence, in order to successfully use such material to build electrodes we propose a bio-potential amplifier with an ultra-high input impedance. This high impedance allows our electrodes to overcome the barrier of the increased contact impedance and contact impedance imbalance that it is usually mitigated by the use of conductive gel or paste. The design, as reported later in this section, uses only commercial ICs and can thus be readily replicated by other researchers in the field. Laboratory and clinical tests demonstrate that the system is able to acquire the ECG and EEG of subjects as well as clinical ECG and EEG devices (0.95 correlation) and improved monitoring can be performed for at least 24 hours. The device (for ECG purposes), is shown to work even when the subject is fully immersed in water.

However, as mentioned, dry electrodes still suffer of many problems such as increased artifact sensitivity and high contact impedance value, which solutions are still the object of active international research.

4.1 The bio-amplifier hardware

We designed the biomedical front-end to provide a very high input impedance, and a selectable gain feature from ~1000 V/V for EKG/EMG up to ~70 kV/V for EEG applications. The bandwidth of the front-end is adjustable (from 0.05 or 0.5 Hz up to 40 Hz for EEG and EKG applications, or from 5 Hz up to 400 Hz for EMG applications). Bench testing shows that the front end current consumption is less than 1 mA when the circuit is powered by 4.5 V.

Figure 2 shows the implementation of a single analogue channel. The Burr-Brown INA-116 instrumentation amplifier is used for its extremely high input impedance. A virtual signal ground is derived from the battery (single power supply) using a voltage divider (R5/R6). The virtual ground is buffered to provide an optional driven ground connection for the right ear lobe (EEG) or for the right leg driver connection (EKG) via a calibrated coupling impedance Rcouple.
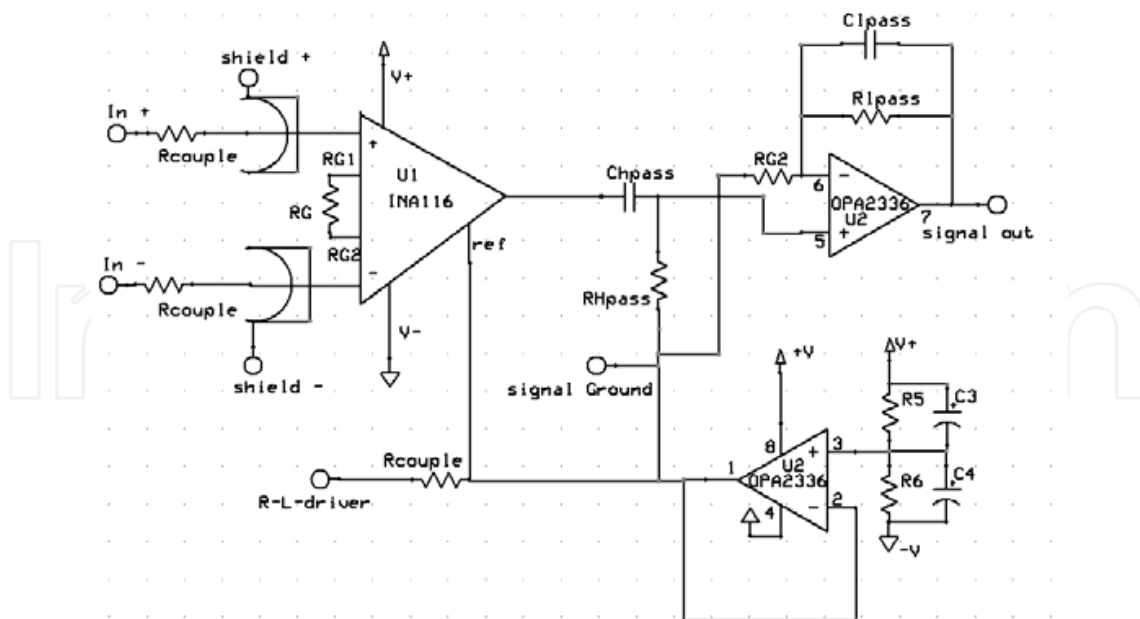
Fig. 2. Schematics of a single input channel

The INA116 is designed to work with a dual 9 V power supply, but because of the very low bandwidth requirement and the small amplitude of the biomedical signals, we were able to use it with a single voltage supply of only 2 V.

The cut-off frequency for the high pass filter is tuneable by changing the value of CHpass while RHpass is kept fixed at 390 kΩ. The second stage of amplification and filtering provides enough gain and high frequency suppression to directly feed the ADC. The cut-off frequency of the low pass filter is regulated by tuning CLpass while keeping RLpass fixed at 1 MΩ. The second stage amplification and the driven ground are implemented using the low-power, precision operational amplifier OPA2336 from Burr-Brown. The measured input referred noise for the proposed circuit (included electrodes) was less than 2 µVpp in the bandwidth up to 10 Hz. (Gargiulo, Bifulco et al. 2008; Gargiulo, Bifulco et al. 2008). A peculiarity of the amplifier is that it offers driven active guard shields (*shield+ and shield- in figure 2)* on each input to minimize capacitive signal leakage due to the PCB or cables. This ensures low input bias current and high input impedance which allow the recording of biopotentials from high contact impedance, dry electrodes connected by cables with driven shields. .

4.2 The passive dry electrode

Our electrodes are made with commercially available 1.5 mm thick silicone conductive rubber shaped in discs of 8 mm diameter. Figure 3 shows an illustrative diagram of the dry electrode. The active side of the electrode is capacitive coupled through a layer of insulating silicon rubber with a metal shield wired to the active guard shield. The impedance of the realized electrodes at 100 Hz is greater than 20 MΩ with a parasitic capacitance no greater than 2 pF. Laboratory tests demonstrate that even a tolerance of 20% for the electrode's impedance is acceptable and does not influence the quality of the measurement, even in a multi-channel montage. In order to avoid any accidental contact with the electrode's shield or with the optional grounded cable sleeve, a final layer of insulation rubber (not showed in Figure 3), is poured to cover them.(Gargiulo, Bifulco et al. 2008; Gargiulo, Bifulco et al. 2008)
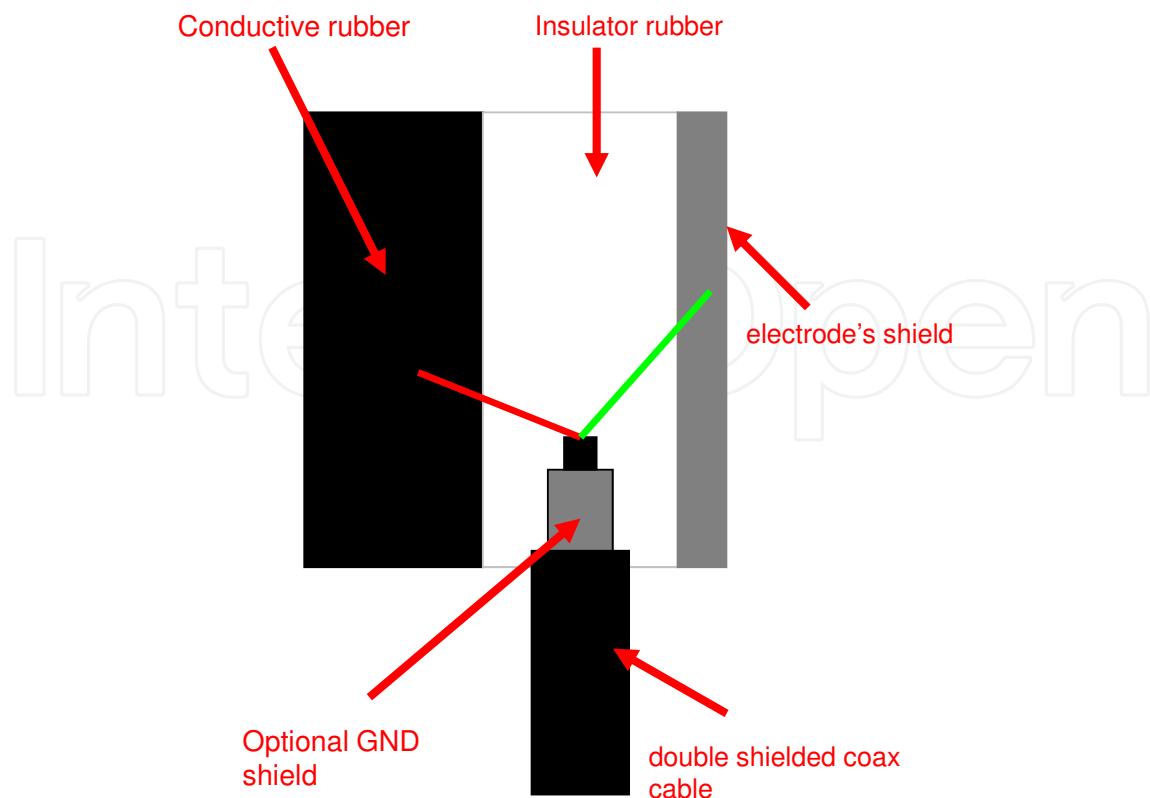
Fig. 3. Dry electrode illustrative diagram

The quality of the measurement it is not influenced by contact impedance imbalance between electrodes. It is possible to mix the electrodes – some dry and some wet – in our system.

4.3 Wearable personal monitor system

Wireless connectivity using a low-power ADC equipped with a Bluetooth module (currently approved for medical devices), was added in order to complete the system. The module is capable of acquiring up to eight analogue channels with a total sample rate of 4000 Hz transmitting data up to 25 m open field. The measured current consumption of the module operating in full mode (maximum sample rate available) is rated 35 mA when powered from a 4.5 V battery pack and 6.5 mA when operated in SNIFF mode. SNIFF mode is a standard Bluetooth low power modality that on the one hand reduces the power consumption, but also limits the data throughput of the device down to 500 Hz maximum sample rate. The general architecture of the realized device is depicted in Figure 4.

It is possible to observe data are acquired from a 3-axial accelerometer (for body or body parts motion sensing) and a number of bio-medical front ends in according with the application and the sample rate constraints.

In summary the prototype of our wearable personal monitor has the following specifications:

- Dimension: 43x60x15 [mm]
- Weight: 200 gr (battery excluded)
- Power consumptions: 40 mA @ 4.5 V reduced to 10 mA @ 5 V when SNIFF mode is used (M. Catrysse & H. van Egmondc 2004)
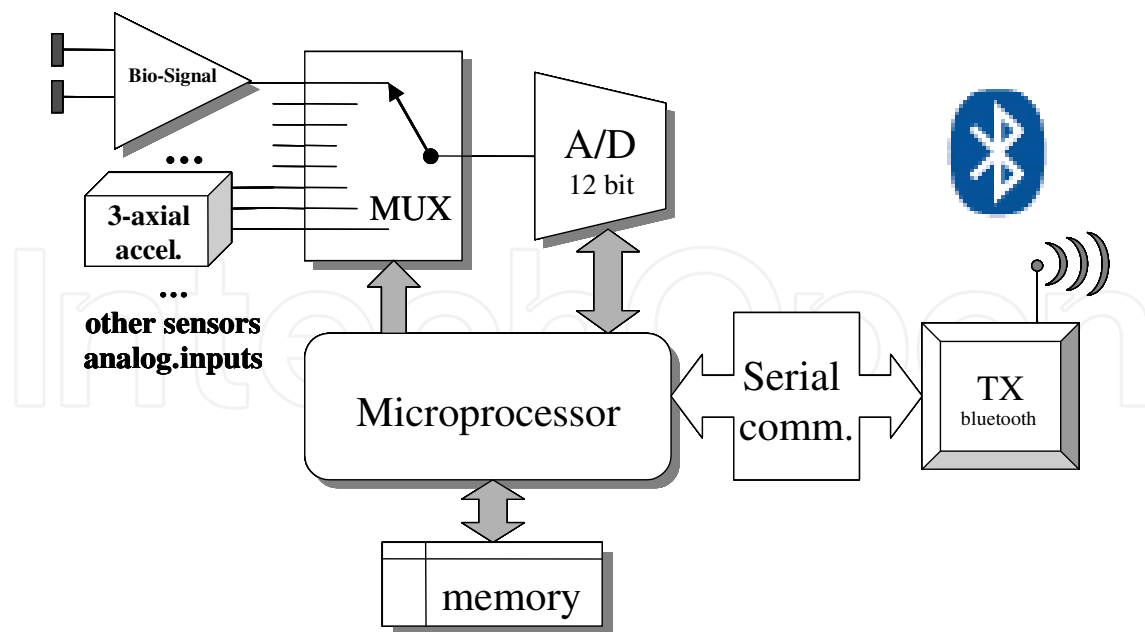
Fig. 4. System architecture

- AA batteries can provide up to 3000mAH, operation time is up to 300 hrs in SNIFF mode.
- Up to four configurable bio-front-ends
- Body motion detection with 3-axial MEMS accelerometer: 0.8 V per g (g=9.81 m/s^2)
- Customizable sample rate up to 4 KHz (500 Hz in SNIFF mode)
- Data security (Bluetooth 2001)
- Standard and wearable textile electrodes suitable.

4.3.1 ECG application

Acquiring data from a single Bio-medical front end configured as ECG (standard bandwidth of 0.5÷150 [Hz], sampling the data at 500 Hz or lower, it is possible to realize a very low power ECG monitor. Figure 5 shows an excerpt of data as they are recorded from a volunteer subject.

Then the qualitative performance of ECG system is assessed in parallel with standard commercially available ECG devices. In this comparative test, dry electrodes belonging to our system are placed on the chest of a volunteer subject as close as possible to standard wet electrode belonging to the control system. Skin preparation was used for wet electrodes but not performed for dry electrodes.

As shown in Figure 6, the signals were almost identical in freshly installed electrodes (fresh montage), with the correlation coefficient of the two signals scores 0.95. Correlation is calculated on the recorded workspace of ten minutes. Small differences in recorded signals come from the different filter orders and slightly different bandwidths (due to the variance of the components) and of course by the slight difference in electrode position. However, as shown in Figure 7, after 24 hours the differences between the two signals become evident. In particular it is possible to observe that the standard system (bottom panel) suffers from signal distortions from a loose contact between electrode and skin. Gel desiccation and adhesive problems do not affect our dry system since it does not rely on a full contact with the skin. (Gargiulo, Bifulco et al. 2008)
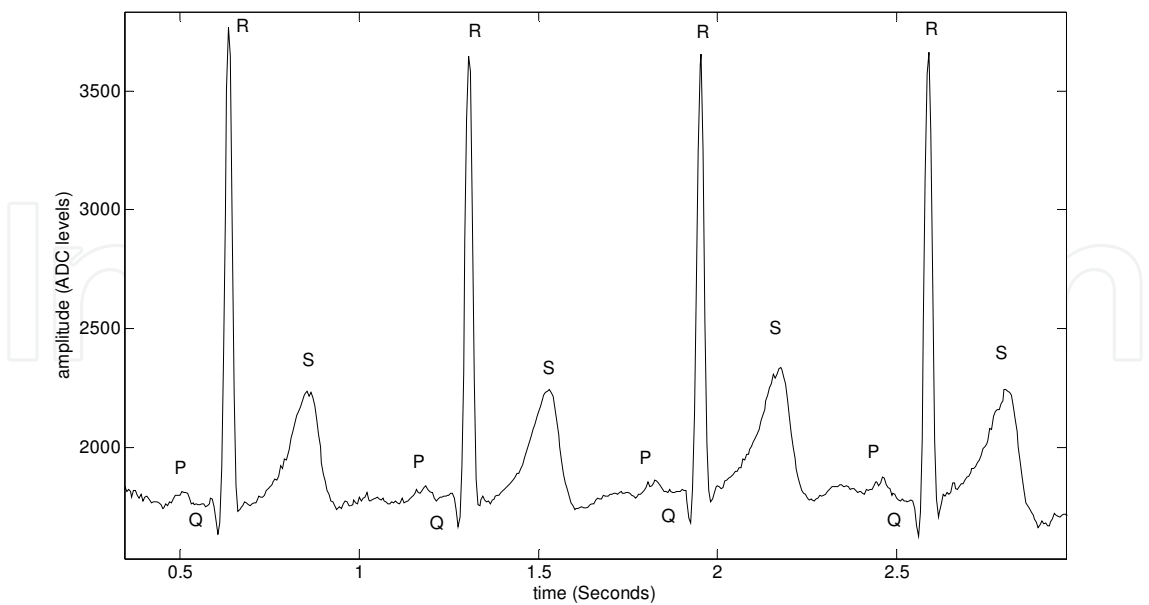
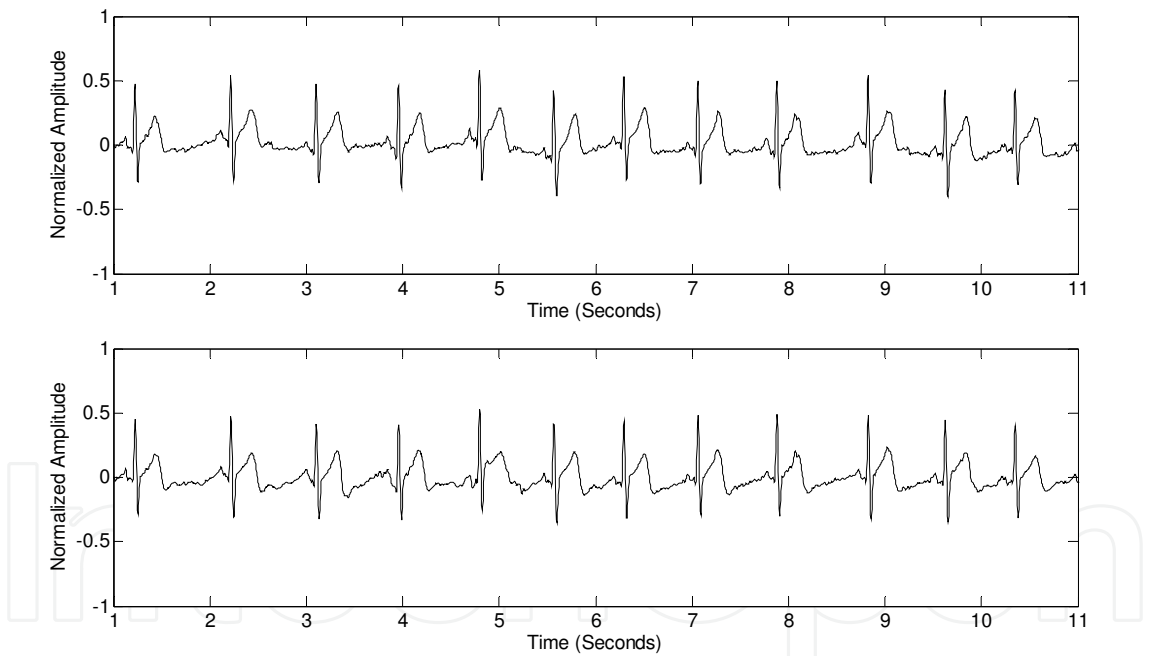Fig. 5. ECG raw data acquired from a resting subject

Fig. 6. Parallel ECG recording in fresh montage (Top panel: dry electrodes system)

4.3.2 EEG application

The realized system is been tested for EEG application during a BCI task; performance of the dry electrode system are assessed based on a parallel recording similar to the one performed for the ECG capabilities assessment. In this configuration, our system is able to acquire up to eight electrodes sampled at the maximum frequency of 128 Hz. Moreover, reference electrode circuitry was built without the use of additional circuitry, the shielding terminal of each electrode (referred as *shield+* and *shield-* in Figure 2) gives a replica of the sensed signal
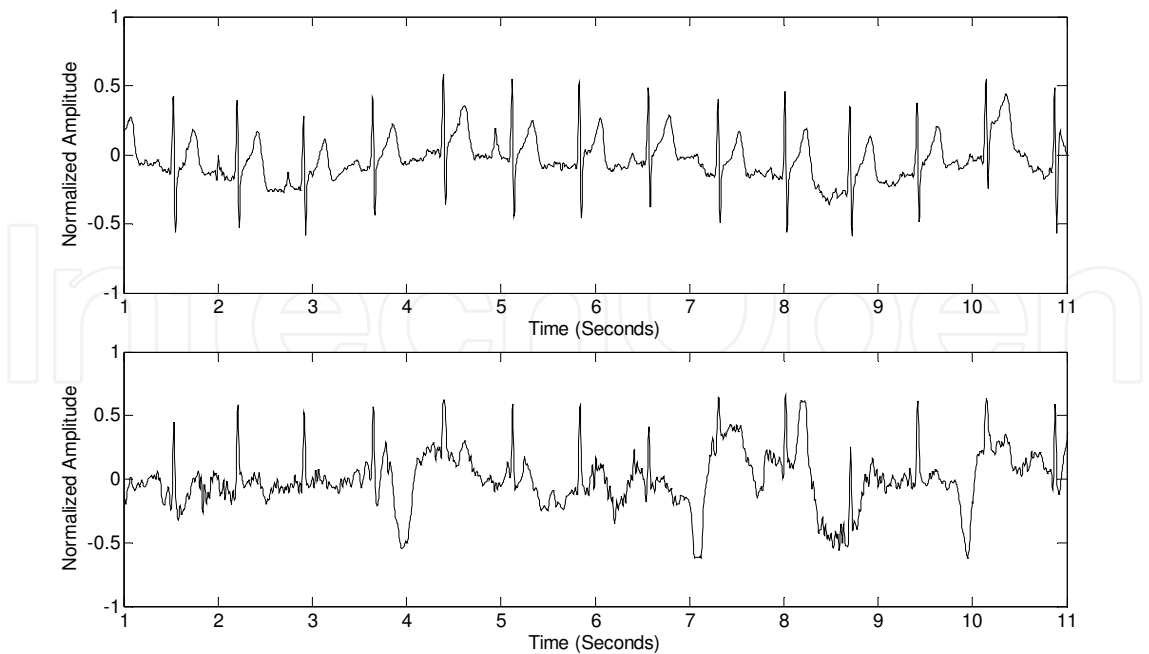
Fig. 7. Parallel ECG recording after 24 hours (Top panel: dry electrode system)

at the electrode (Horowitz & Hill 2002; Bifulco, Gargiulo et al. 2007; Gargiulo, Bifulco et al. 2008; Gargiulo, Bifulco et al. 2008), therefore, this terminal can be used to replicate the reference signal as depicted in Figure 8. To simplify the drawing, in Figure 8 is depicted only the pre-amplifier section of the first two channels.
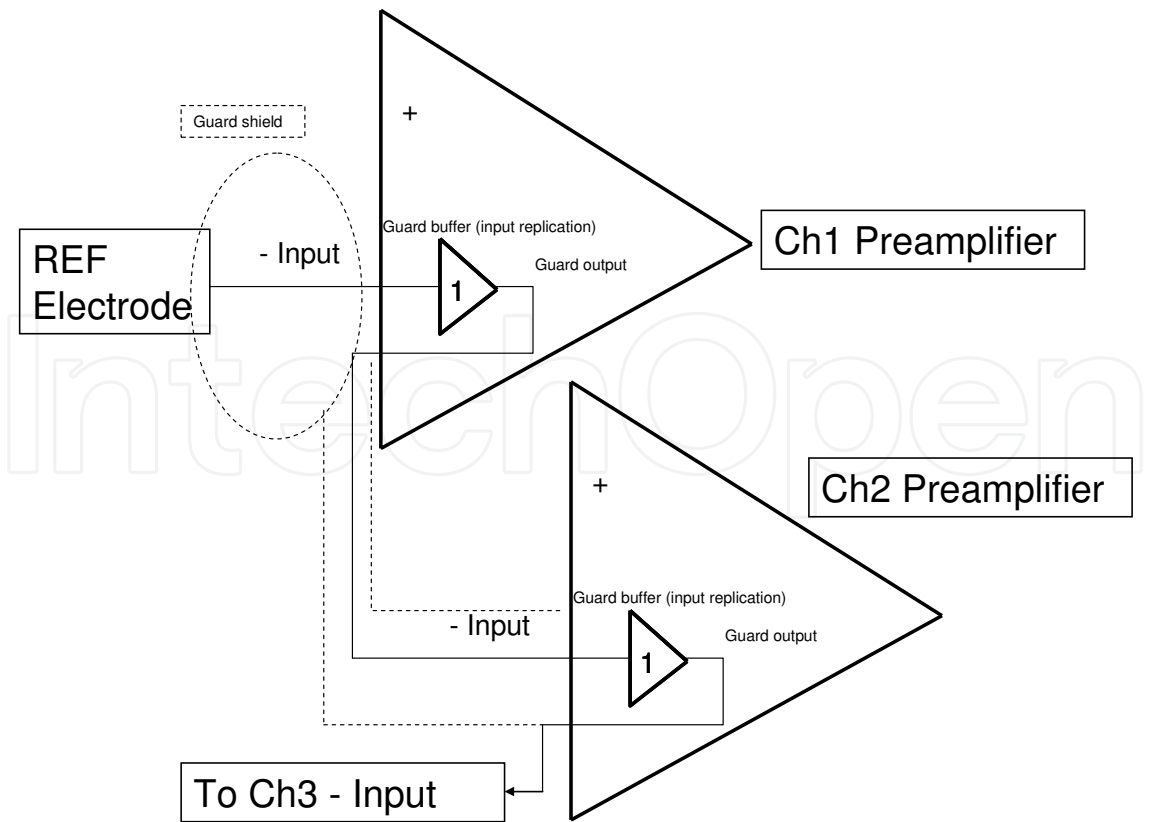

Fig. 8. Multi-channels wiring

A number of untrained subjects were asked to perform a BCI mono-dimensional cursor control task (left-right movement) with the following protocol consisting of three steps.

1) Familiarization trial (approximately 3 minutes): in this task the subjects were asked to manually depress a button with their right or left hand when a target appeared on the respective side;

2) Pre-BCI trial of 3-6 minutes, where the subjects were asked to imagine pressing the button when a target appeared on the respective side;

3) BCI L-R control tasks (until the subject got tired), where a cursor was moved based on the EEG signals recorded.

During the experiment EEG signals were recorded in parallel by both machines using the following montage: dry electrodes were placed at C3, C4, and Cz (also position used from the BCI classifier) and were surrounded by wet electrodes (belonging to the control machine) at Cp3, Cp4, Cpz, C1, C2, C5, C6, Fc3, Fc4 and Fcz.

Since we are interested in evaluating how the experimental burden is reduced using the new system, the time required to prepare the subjects was recorded, in particular, the preparation time per electrode was recorded separately for dry and wet electrodes. For the wet electrodes, full skin preparation, and contact impedance checking are required. The wet system required 2-3 minutes set-up time per electrode; wet electrode are kept in position using conductive adhesive paste and a bandage soaked in standard collodion. The dry system only required ten seconds per electrode, which was the time needed to dry the collodion applied directly to the surface of the electrode, Figure 9 shows the electrode montage, arrows indicate the dry electrodes.
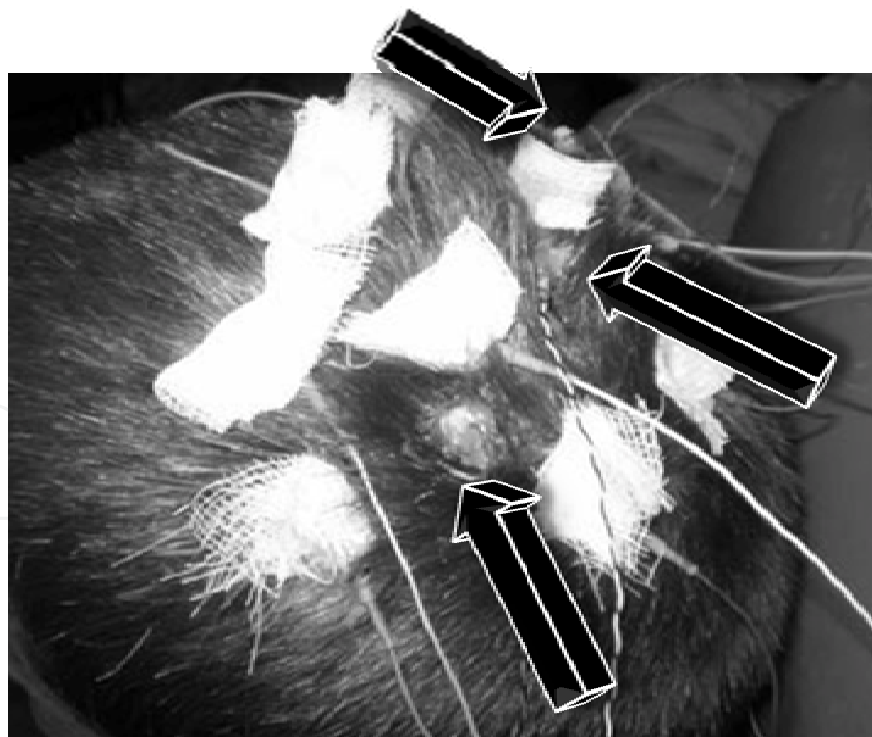

Fig. 9. Electrode montage, arrows indicate the dry electrode

In order to minimize the differences in the acquired signal due to hardware differences, the data were equalized in bandwidth to 0.5-35 Hz using a band pass filter (50th order FIR) and a 50 Hz IIR notch filter was applied to the recorded signals.

Time and frequency domain evaluation was performed on the data. In the time domain, in order to minimize the effect of clock misalignment and different ADC jitter in the two recording systems, an analysis based on the maximum of the correlation between signals recorded with the two systems was used. Using a one second long (256 samples) moving window, we calculate the correlation between electrode signals. We found that the maximum correlation of a 3 minute recording, i.e., when the two series are time aligned, was 0.90 when comparing electrodes from the same machine. The average of the maximum correlation between a dry electrode and the mean signal from its surrounding wet electrodes was 0.76.

We believe that the difference is mainly caused by the presence of artifacts that introduce high amplitude noise. Since each machine has slightly different recovery times and filter responses, these artifacts reduce overall correlation between the two systems. Figure 10 shows an excerpt of data where the subject was asked to strongly contract the jaw muscles, due to the absence of the gel able to mitigate the effect of the electrode bumping that follow the large artifacts, the response of the dry electrodes (represented in bold) is clearly different.
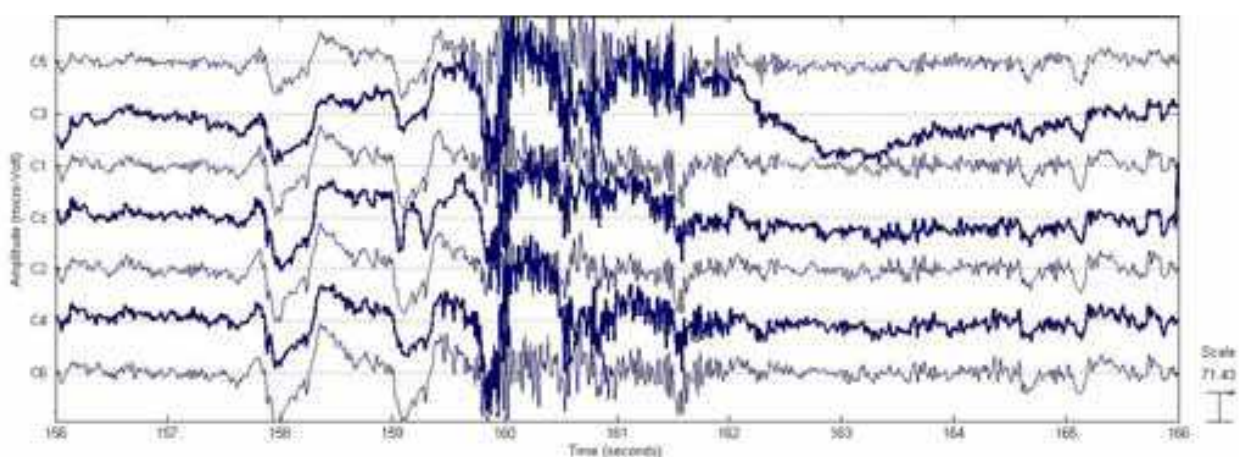

Fig. 10. Ten seconds EEG recording showing clear differences to artifacts between dry (bold line) and standard wet electrodes

Our hypothesis that the above artifacts account for the main differences between dry and wet recordings was tested in further analysis where the signals were visually inspected by a neurologist who was unaware which signals had been recorded by which system. They were asked to identify periods within the trials that contained artifacts and these were removed. Following the removal of artifacts, the average of the maximum correlation between a dry electrode and the mean signal from its surrounding wet electrodes increased from 0.76 to 0.94 (Gargiulo, Bifulco et al. 2008). Further tests show that the combined use of the amplifier and the passive dry electrode allows us to mix in the same montage dry and wet electrode. It is possible to record EEG using as reference a standard golden brass electrode applied as usual (conductive paste and collodion soaked bandage), while the active or exploring electrodes are dry passive ones. This montage superimposes a contact impedance imbalance of at least $100\,k\Omega$ (wet prepared electrodes usually offers a contact impedance smaller than $5\,k\Omega$, while the passive dry electrodes contact impedance was always greater than $100\,k\Omega$ or even off the scale of the instruments).

4.2.3 Athlete monitoring

Configuring our system to acquire the tri-axial accelerometer and a single EMG channel at the sample rate of 500 Hz (that is the minimal sample rate for EMG applications (Ives & Wigglesworth 2003)), it is possible to realize a body part monitor. For example the accelerometer was installed on an athlete's wrist using an elastic band. Another band, containing two wearable electrodes for EMG recording, was installed on the biceps. The athlete executed several cycles of a standing biceps curl with supination. During these slow movements, the dominant acceleration is gravity, so that information of the orientation of the accelerometer can be extracted. A remote coach was able to gather information about the athlete performance from the acquired signals as follows:

- Rhythm and speed of the cycle's executions was determined by looking at the shape and time duration of the accelerometer's waveforms.
- As depicted in Figure 11 a single repetition cycle of the exercise can be divided into four sections. During section 1 (curling) the forearm is being raised and the acceleration on the red axis decreases, while the acceleration on the blue axis increases as the sensor is being tilted towards the horizontal. Next supination occurs when the lower arm turns, causing acceleration on the green axis to increase, while decreasing acceleration on the blue axis (section 2). This is followed by voluntary peak contraction (section 3) and relaxation that includes gradual pronation to return in the initial position (section 4).
- The presence of cheating, and the amount of cheating can be evaluated by abnormal acceleration just before a new curl is executed. This acceleration indicates that the arm is being swung prior to curling in order to increase the momentum to lift up the weight.
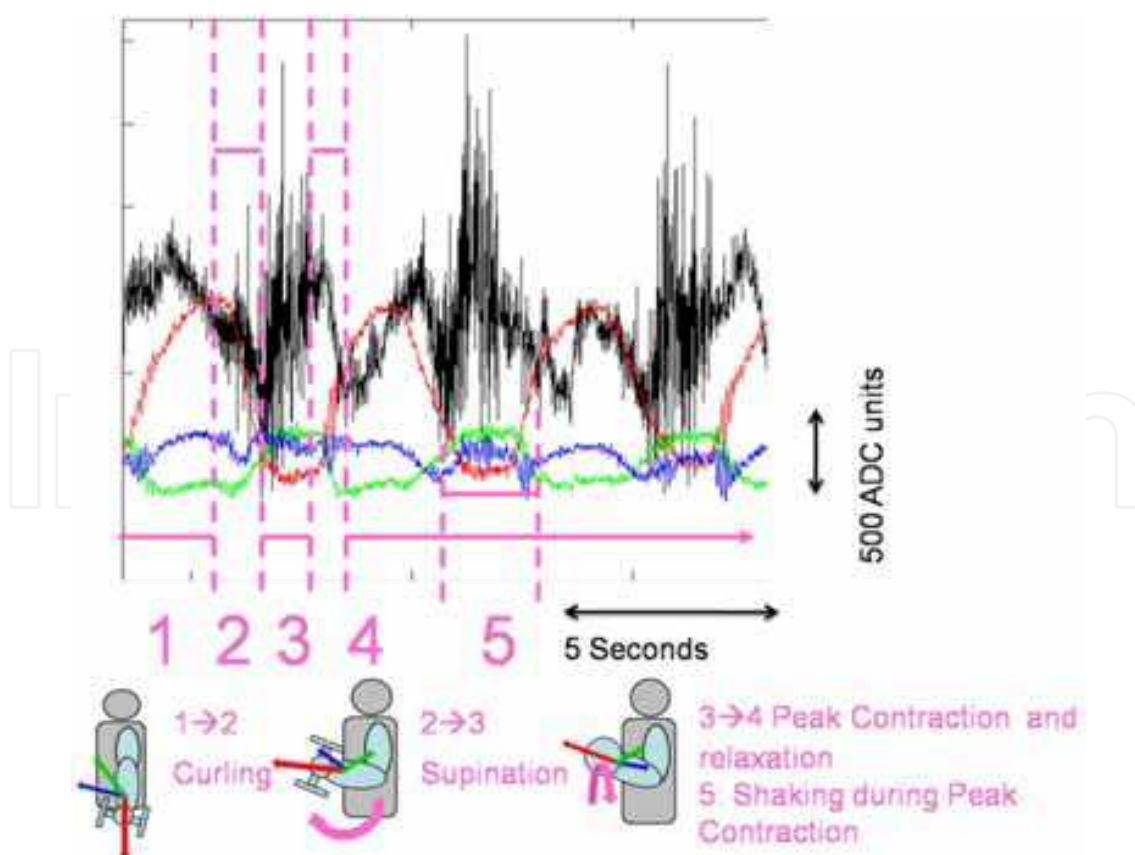

Fig. 11. Remote standing biceps curl assessment

- The voluntary peak contraction, denoted by the forearm shaking pattern (as it is evident in Figure 11, blue trace, section 5);
- The evaluation of the EMG signal (in particular its RMS value) will give information about muscular stress and fatigue across the cycles.

Configuring the hardware to acquire ECG and the tri-axial accelerometer, it is possible to obtain a full body exercises remote athlete management system or a general full body patient monitoring system.

A good example of athlete management is depicted in Figure 12, in this task the subject executes a squatting exercise. In this kind of exercise, the athlete usually carries a heavy weight on the shoulders and a loss of equilibrium over the horizontal axis (depicted in Figure 12 in turquoise) could be very dangerous. In addition, since during squatting the chest will bend forward, there is a high risk of overcharging the back muscles during return to the standing position. A remote coach can judge the quality of the exercise by evaluating several parameters such as:

- Recovery time between cycles from duration of the plateau on the vertical accelerometer axis (red trace).
- Lateral equilibrium loss denoted by the presence of peaks in the accelerometer's horizontal axis (turquoise trace)
- Recovery time in squatting position (length of simultaneous plateaus on the turquoise and green axes)
- Potentially dangerous charge of the back muscles during standing up. A hyperextension of the back will result in a negative deflection of the accelerometer's green axis.
- Additional information could also be gathered from the EKG such as heart rate and heart rate variability within cycles as well as the respiration rhythm.
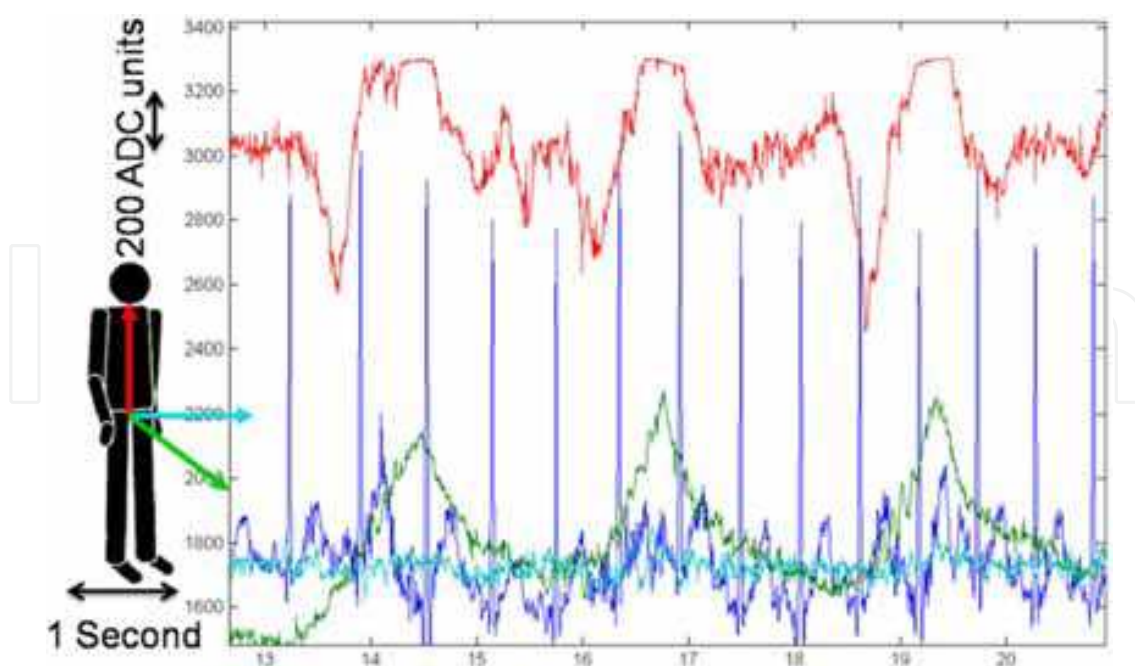

Fig. 12. Remote squatting exercise assessment

Even when the subject is not performing any particular task, since the accelerometer used is sensitive to gravity it is possible to transform the recorded acceleration along the three axes

into tilt angles of each axis with respect to the direction of the gravity vector, for static positions, as well as during slow movements. This allows us to assess body posture in the gravitational field.

Figure 13 shows an example of posture assessment. In this trial, the signal recorded from the accelerometer positioned in the subject's belt and oriented as depicted, was translated in axis tilt with the gravity direction. Figure 13 shows the following body positions:

A: Standing up
B: Lying horizontally (face up)
C: Lying horizontally on the left side
D: Lying horizontally on the right side
E: Lying horizontally (face down)
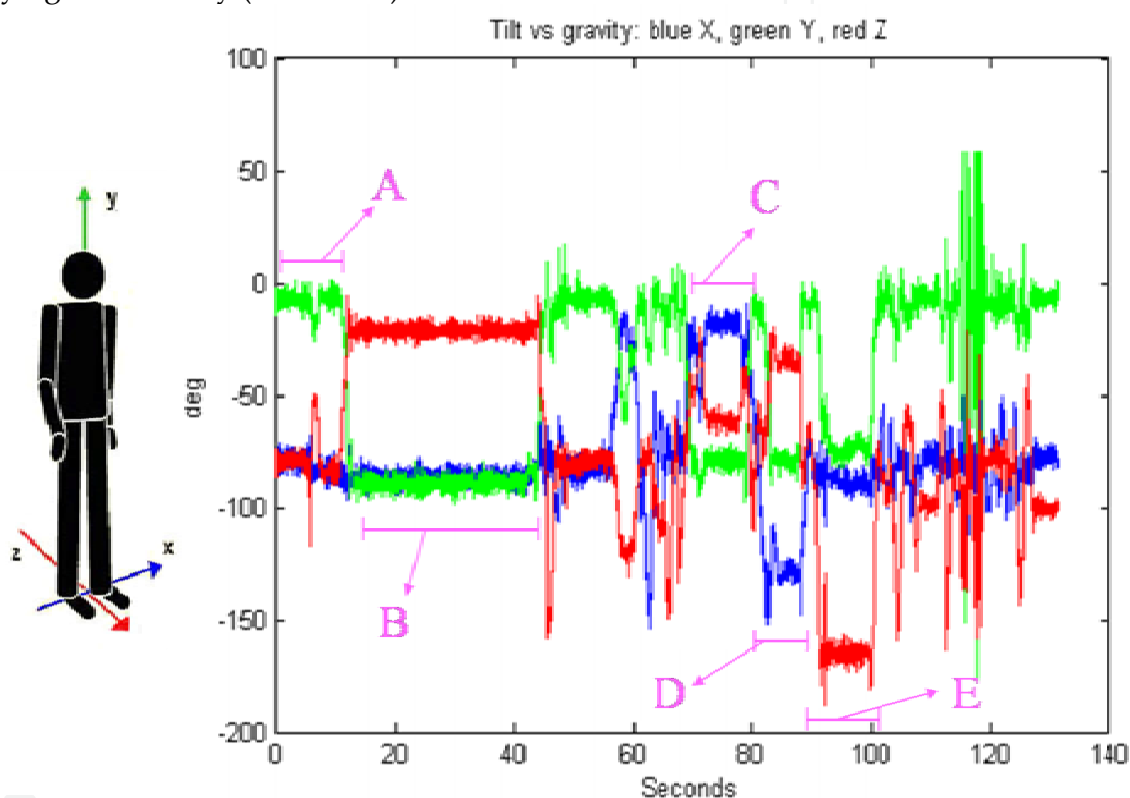
Fig. 13. Posture assessment

We found that that such a system is able to assess human performances and it may be useful in military and high risk zone operator monitoring and management. Moreover, it is possible to combine more than one system configured for different signals on the same subject to obtain more detailed information on the subject' performance.

5. Conclusion

Non-invasive biosensor systems are increasing in demand and many useful applications exist, particularly in long term monitoring One bottleneck is the current need for obtrusive 'wet electrodes' that fail to work in long term recordings. We have described a long term monitoring, wearable personal monitoring system that is wireless, low power and uses convenient dry electrodes. Its use for ECG and athlete monitoring has been demonstrated. This would be useful for those at risk of heart failure or health and activity monitoring in

our increasingly sedentary and overweight society. We are currently assessing the system for EEG recordings, in particular for as a BCI device that would greatly assist the severely disabled and it may also be of use in epilepsy monitoring being able to track movement and record EEG in a comfortable environment.

6. Future researches

Currently our researches are focused on exploring all the possible uses of the proposed biomedical sensing system particularly in athlete and long term patient monitoring and BCIs.

6.1 Physical activity monitoring

Currently our research in physical activity monitoring is still focused on clinical assessment of human performance for long term monitoring particularly for full body assessment. It is well known that rapid changes in body orientation, such as during a free fall, may be identified from the information gathered by the accelerometer. Figure 14 shows an example using data recorded using our device. Moreover, being able to detect rapid changes in body orientation provides useful information for syncope detection, geriatric care and sport science.

In this evaluation the prototype was attached to the subject's chest using an elastic band with embedded dry electrodes. Our device was configured to acquire one EKG channel, a signal from the light reflected PPG (Photo PletysmoGraphic), unit, and skin temperature (not shown). The top section of Figure 14 shows the posture assessment gathered from the accelerometer during a passage from a lying down (face up) to a standing position. The lower section of Figure 11 shows the event related biological signals, i.e., ECG (1st lead, top trace) and PPG signals (bottom trace).

The passage from a lying position to a standing will cause a large blood pressure gradient inside the body (a vasovagal reaction) and this could be a cause for a syncope attack (Benditt, Ferguson et al. 1996).
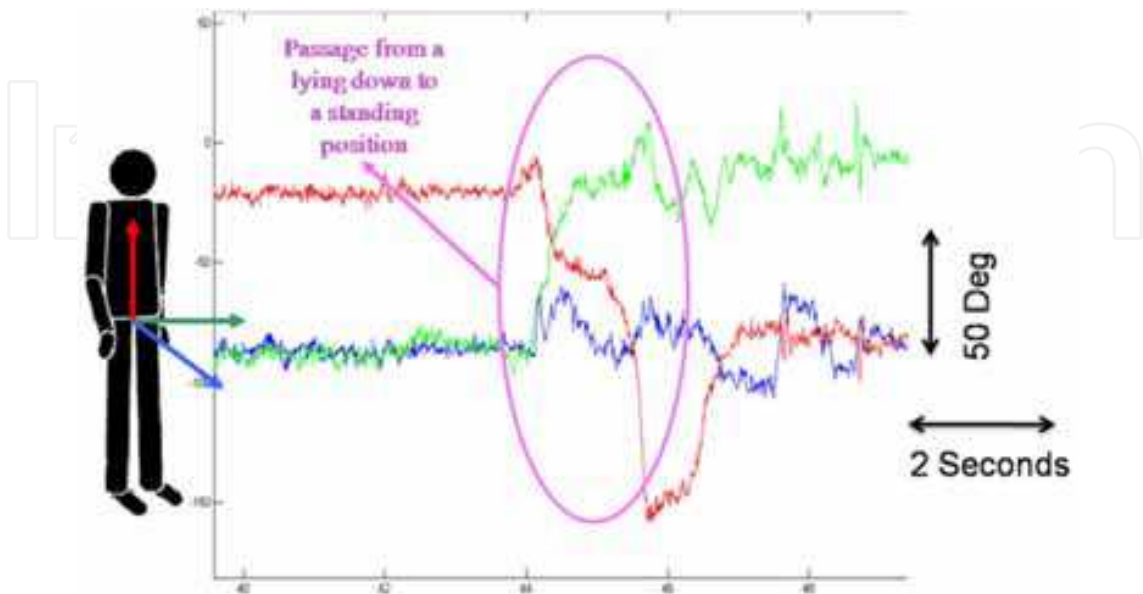

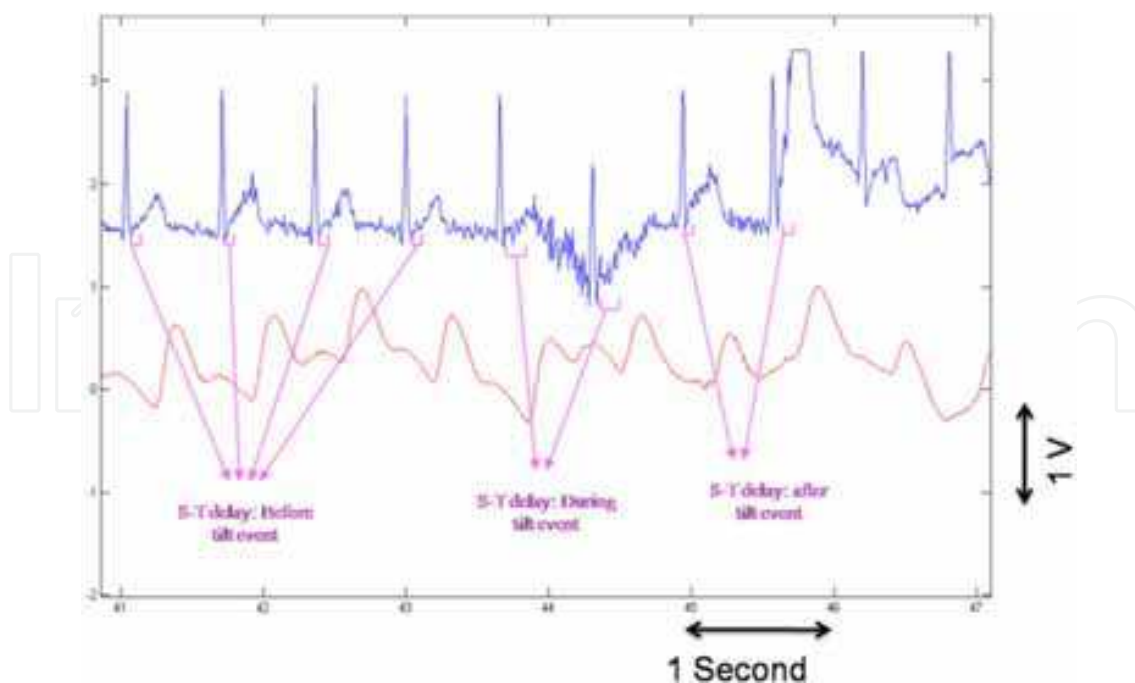Fig. 14. Posture assessment, accelerometer signals

Fig. 15. Posture assessment, ECG and PPG signals

By wearing our device it will be possible to extract important information about the subject's health from the data recorded continuously each time that the subject changes from a resting position to an upright position. Evaluating the EKG signal (shape of a heart beat and delay between beats) during these events could improve therapies of at risk or elderly patients. Currently we are developing algorithms for the automated extraction of this information from long term monitoring periods (24hr or more). Recalling the well known Newton's formula that allows given the mass and the acceleration to calculate the force (F) as:

$$F = ma$$

Theoretically is possible to calculate the power (and then the calories expenditure) for a given exercise in a given time for a subject of known mass. It is worth to highlight that the calculation is not that trivial because it is obvious that the acceleration information that is possible to retrieve from the single posture sensor does not result enough to assess such estimation. However, further experiments using professional athlete in known tasks are scheduled to measure the error when comparing the calories expenditure calculated with the accelerometric sensors with the one calculated using standard equipments.

Moreover, an interesting link to the EEG, long term brain monitoring is the uses of accelerometers to detect seizure movements, as subjects usually have repeats of the same type of seizure the accelerometer could be placed on the known limb.. This might be able to serve as a proxy for the video used in clinical EEG, to correlate movements with spikes or the prediction of spikes.

6.2 ECG application

Our new focus in ECG application is the continuous monitoring of swimmers and divers. Usually this application requires water proofing of the electrodes because the water can short recording sites, moreover, water resistant glue must to be applied to keep the electrode in position.
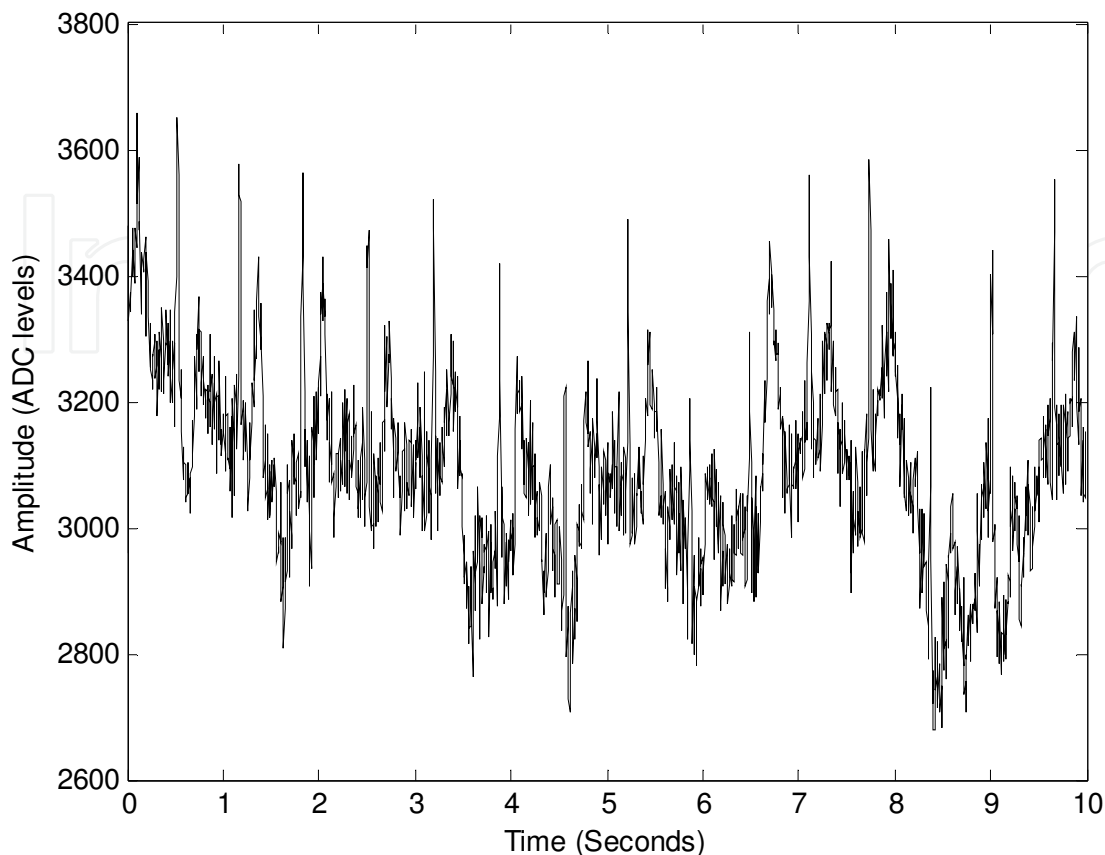
Fig. 16. Underwater ECG recording

The use of the proposed monitoring system opens a new monitoring scenario in this field as well. Even though our system is designed to operate in a dry environment, it can also be used in a wet environment it will even work when submerged in water. Figure 16 shows an excerpt of the data (raw) recorded from a subject totally submerged in fresh water, electrodes are placed on the chest. No special skin preparation was used and no waterproofing was performed at the electrode level. As it is possible to observe from the trace, the ECG signal is clearly recognizable, the baseline variation and the EMG artifacts clearly affecting the signal are due to the chest's muscles that the subject was using keep himself totally submerged (Gargiulo, Bifulco et al. 2008).

6.3 Long term of brain signals

Dry electrodes are obviously more convenient for long term EEG studies as gel melts as it heats up with body temperature, it smears shorting electrodes and is not convenient. Moreover EEG based BCI systems that ideally are to be worn as "plug and play" machine would have a great advantage from a system that result easy to install and remove, or even stable and reliable particularly when the subject is learning the BCI control. BCI training experiments can result tiredness and often the subject preparation takes longer that the experiment (in dense EEG montages). Therefore, beside the quest in finding a dry electrodes holding system able to work as good as the collodion glue (Gargiulo, Bifulco et al. 2008), without the mess caused from its repeated use, our current investigations are focused on the

role played from the feedback in BCI. Typically (but not always (Hinterberger, Neumann et al. 2004)) visual feedback is given to the user; however, it is broadly recognized that feedback plays an important role when subjects are learning to control their brain signals.

Moreover, it is worth highlighting that long term EEG monitoring could be part of a system that detects seizures and initiates automatic therapy (vagal nerve stimulator, deep brain stimulator or antiepileptic drugs) There is now even evidence that EEGs might predict seizures with inter-cranial electrodes (Waterhouse 2003).

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Intelligent and Biosensors

Edited by Vernon S. Somerset

The use of intelligent sensors have revolutionized the way in which we gather data from the world around us, how we extract useful information from that data, and the manner in which we use the newly obtained information for various operations and decision making. This book is an attempt to highlight the current research in the field of Intelligent and Biosensors, thereby describing state-of-the-art techniques in the field and emerging new technologies, also showcasing some examples and applications.

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Gaetano Gargiulo, Paolo Bifulco, Rafael A. Calvo, Mario Cesarelli, Craig Jin, Alistair McEwan and André van Schaik (2010). Non-invasive Electronic Biosensor Circuits and Systems, Intelligent and Biosensors, Vernon S. Somerset (Ed.), ISBN: 978-953-7619-58-9, InTech, Available from:
http://www.intechopen.com/books/intelligent-and-biosensors/non-invasive-electronic-biosensor-circuits-and-systems

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