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UNOBTRUSIVE SENSING OF PSYCHOPHYSIOLOGICAL PARAMETERS

Some Examples of Non-Invasive Sensing Technologies

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Abstract The quantification of the human perception of experiences can be achieved by the sensing of specific psychophysiological parameters. A growing interest develops for the daily life use of these quantification techniques by unobtrusive and unnoticeable data collection. Remote and non invasive sensing technologies are discussed for the sensing of the following psychophysiological parameters: heart rate variability, and muscle stress. A generic miniature platform for miniature wireless sensing applications is described as an important enabler for unobtrusive and unnoticeable sensing. The technology no longer seems to be a limiting factor for unobtrusive and unnoticeable sensing. Initially the sensors will be worn on the body, but ultimately implantable sensors will become widely accepted, allowing access to new parameters, such as hormone levels and body core temperature.

1 Introduction

Psychophysiological parameters, such as heart rate variability, galvanic skin response, breathing rhythm and muscle stress provide important information on the emotional and cognitive state of a person. Monitoring these parameters thus provides information on the perception of a person in daily life. For Philips the ability of probing the experiences of people is a long standing wish for a multitude of applications. In order to obtain unbiased and accurate sensor data the person under scrutiny needs to be unaware of the sensors. Above all it is important that the presence of the sensors is not disturbing or influencing the wearer.

The ultimate goal for supporting technology in the field of measuring experience is the availability of unnoticeable maintenance free sensors, wirelessly transmitting for example a person's emotional status in a real-time fashion. Clearly some steps are needed to achieve this, so in the short term researchers in the field have to make do with what is available now. This paper does not claim to present an overview of the current hardware capabilities for probing

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experience. It does however try to give the reader a flavour of what is currently being studied at Philips Research in terms of unnoticeable and unobtrusive sensing of psychophysiological parameters.

A distinction can be made between laboratory environment testing and field testing. Field testing requires wearable equipment often capable of maintenance free operation for days or even weeks, whereas laboratory experiments can be operator assisted, hardwired, recorded on video et cetera. In Figure 1 a cartoon shows an exaggerated representation of what the consequence is of the attempt to collect data simultaneously on a large number of parameters: the person under scrutiny clearly is influenced by the presence of the sensor equipment, leading to a distorted outcome of the experiments.

A valuable source of data for the interpretation of the experience of sensations is the measurement of electrical activity in the brain (Electro Encephalogram, EEG). Unfortunately the application of a multitude of EEG probes on the human skull is bound to influence the wearer significantly. Undistorted or minimally distorted sensor data can be obtained only when the sensors are not noticed by the wearer. In most cases this rules out equipment such as high resolution cameras for facial expression interpretation.

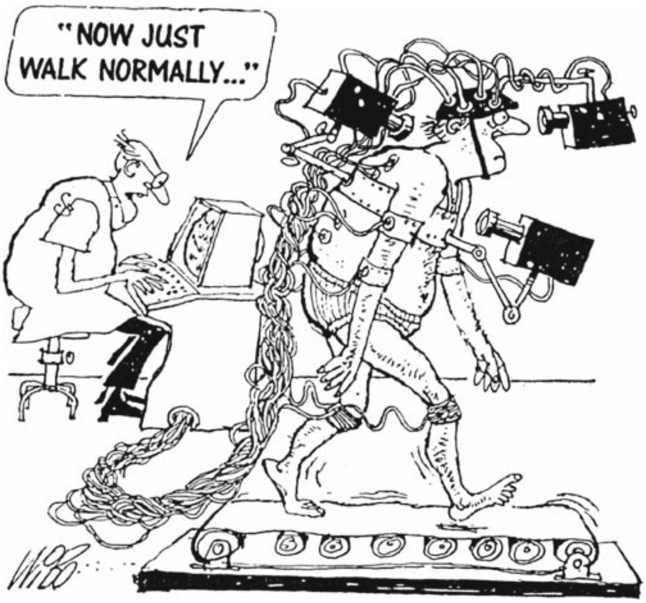

Figure 1. A typical example of obtrusive monitoring (cartoon by Wim Boost, http://nl.wikipedia.org/wiki/Wim_Boost).

An exception may be when the camera can be positioned in a place where the person under test is not capable of detecting its presence, for instance behind a semi transparent mirror.

Special emphasis will be given in this paper on the recording of psychophysiological information in a daily life setting in an unnoticeable and unobtrusive manner.

This paper endeavours to discuss in detail the work on three areas of unobtrusive sensing which are seen as interesting for Philips: heart rate variability sensing from a chair, muscle stress sensing from clothing and a platform for miniature wireless sensors.

2 Contactless sensors for electrophysiological signals

2.1 Electrophysiological signals

All living cells are surrounded by membranes. These membranes are selectively permeable for various ions and may actively transport them through the membrane resulting into a membrane potential. Nerve cells and muscle fibres are depolarised when activated by a certain threshold voltage. The result is the propagation of a depolarisation wave along the nerve and muscle fibre (Basmajian and Luca, 1985; Merletti and Parker, 2004). A depolarisation wave over the muscle fibre is the direct cause of muscular contraction and is subsequently followed by relaxation. The heart consists of muscle fibres which are synchronised when contracting. This results into a deterministic action potential over the heart tissue. For skeletal muscles, the quick combination of contraction and relaxation of a muscle fibre is referred to as "twitch". Since all muscle fibres in a muscle do not twitch simultaneously, the overall observed potential over a muscle is the random summation of multiple single fibre action potentials. For both the heart and skeletal muscles, the electrical waves in the muscle fibres are conducted to the surface of the skin. When measuring the potential on the human skin due to the depolarisation of the heart we speak of the electrocardiogram (ECG), for monitoring skeletal muscle activity on the surface of the skin we speak of surface electromyography (sEMG). In clinical environments, other methods of electromyography are used. For example, by placing metal needles as electrodes in the muscle tissue we obtain a high precision variant of EMG called needle EMG. For unobtrusive sensinf, the use of surface EMG is the most promising.

Figure 2 shows the depolarisation in a skeletal muscle and the conduction into a surface EMG signal. The muscle of the heart depolarises autonomically, but the transfer into skin surface potentials is similar.

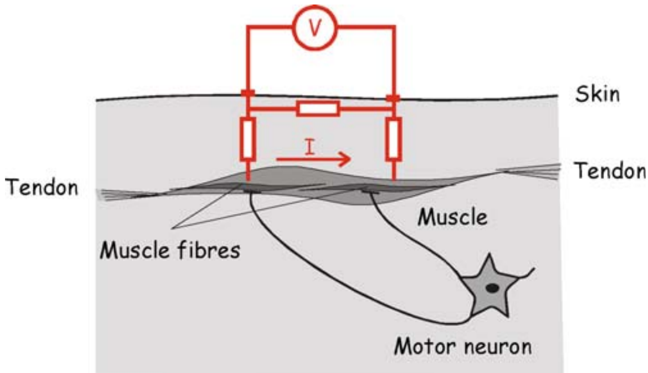

Figure 2. Schematic representation of the transfer of action potentials in a neuron to ionic currents in the muscle fibres and potential distributions on the skin.

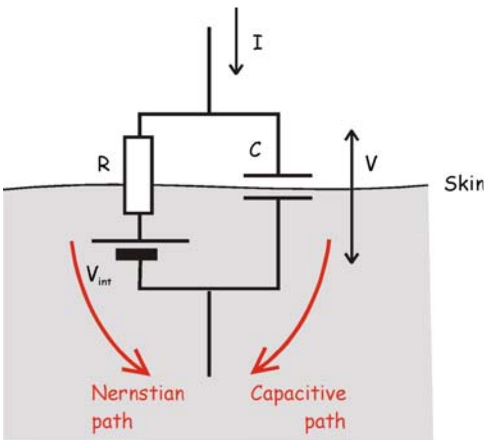

Figure 3. Electrode-tissue interface with components representing the ionic to electronic transition phenomena.

2.2 Contactless sensors

What sEMG and ECG have in common are the electrodes to pick up the skin potentials. A problem with such electrodes is that the interface potential between the skin and the solid electrode is a cause of noise and artefacts. In Figure 3, a closer view of the skin-electrode interface is drawn.

The electrical potentials in the human skin are induced by ionic currents, while electrical currents in the electrodes are the result of moving electrons.

This means that somewhere there is an ion-electron transition which results into a so-called overpotential (Webster, 1992).

There are two parallel paths by which the signal can be transferred from the skin to the electrode. First, there is the Nernstian path, characterised by a transfer resistance and an interface potential. The resistance is the result of dead cells in the stratum corneum and can therefore be reduced by sanding the skin or wetting the electrode. The interface potential is the result of an electrochemical transition described by the Nernst equation. Because the ionic strengths in the skin may differ in time and place, this interface potential is not stable. This is the reason that in conventional electrodes sometimes an AgCl load is included in the gel. This will form a well defined interface potential with the chloride ions in the skin.

So with conventional electrodes, wetting, sanding and AgCl loading is used to stabilise the Nernstian path. Another approach could be to benefit from the other path. This is the path in which signals are coupled directly from the skin to the electrode capacitively. To do this, we simply apply an insulating layer to the electrode surface and take care there is a nice parallel plate configuration of the electrode to the skin.

This was first demonstrated by Richardson of the US Airforce in 1967 (Richardson, 1967) and improved at the Case Western Reserve University in the early 1970's (Ko et al., 1970). It took over thirty years before some academic groups managed to employ modern semiconductor technology to realise small interface amplifiers. Groups structurally working on contactless ECG can be found at the University of Sussex (Harland et al., 2003) and the Seoul National University (Kim and Park, 2005). An American company Quantum Applied Science and Research Inc. (QUASAR) has developed contactless sensors for heart rate monitoring in 2001 as a DARPA project. The results are some patents in the field of capacitive sensors and electrical circuitry to stabilise the input signal (Quasar patents and patent applications).

2.3 Advantages and disadvantages of contactless sensors

The electrical consequence of a capacitive transducer is the high input impedance needed to track the ECG and sEMG signals which have components below 1 Hz. Due to this high impedance, the electrode acts as an antenna for environmental noise. To minimise the noise, active electronics is needed directly on top of the electrode. So complete textile integration will not be possible, there will always be a small module attached to the electrode. Design proposals for such a module are described in section 2.4. Such a module will hamper the options for washability of optional clothing with integrated contactless EMG/ECG electrodes.

Another disadvantage is that the capacitive transducer is more sensitive to motion artefacts than fixed electrodes. The electrical operation of a capacitor is described by the equation

$$i = C\frac{du}{dt} + u\frac{dC}{dt}$$

with i the current through the capacitor, u the voltage over the capacitor and C the instantaneous value of the capacitor. Because the capacitance C changes in time due to motions, the second term *u.cdC/dt* is introduced. So the effect of motion *dC/dt* can electrically not be distinguished from heart and muscle signals *dU/dt*. To minimise this effect, we should take care of proper positioning of the sensor or introduce compensation of the change in the capacitor value.

Note that the problems with motion artefacts are fundamental, but can be solved in principle. Options can be found in digital signal processing techniques, which monitor the capacitance C and compensate for the fluctuations due to motion. So on the long term we expect even measurements with a clinical quality and robustness.

Nevertheless, a sensor which is not in galvanic contact to the skin has some essential advantages. It is uncomfortable to tape conventional contact electrodes onto the skin. The possibility to avoid direct skin contact reduces skin irritation problems. Contactless sensors can be implemented in the environment, for example a chair or wearable clothing. We can measure through clothes and hair and even measure on persons with burn wounds or irritated skin. In addition, the constraints with respect to safety are less critical when there is no galvanic contact.

So despite some concessions to robustness, contactless sensors may open options for new applications for monitoring electrophysiological signals. Especially monitoring over a long time in the natural environment is facilitated.

2.4 Contactless ECG

Figure 4 shows a photo of the capacitive sensors as developed at Philips Research. The size of the sensor is about 32 by 32 mm^2. The diameter of the capacitive coupling area is 14 mm, which is a trade-off between the signal strength (capacitance) and spatial resolution. In contrast to other capacitive sensors (Harland, 2003; Kim and Park, 2005; Ko et al., 1970; Quasar patent and patent applications; Richardson, 1967), care has been taken to realise an embodiment that is flexible. The rationale behind the flexible aspect of the sensor device is that enables unobtrusive measurements. The flexible

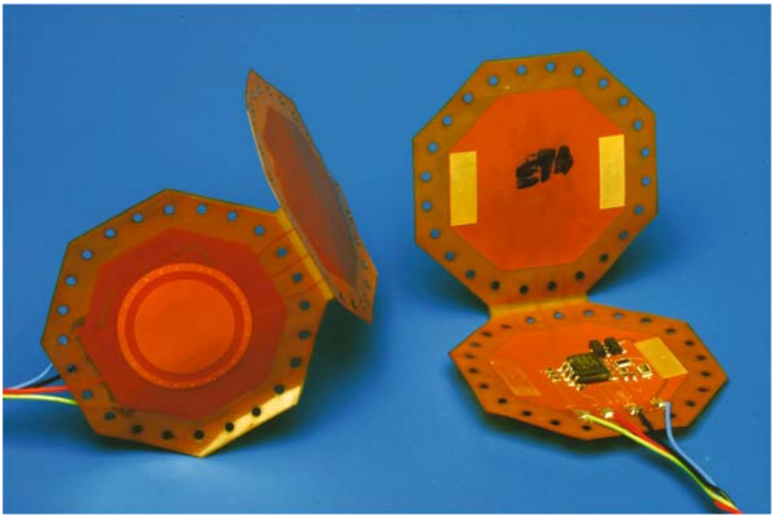

Figure 4. Philips Research flex foil capacitive sensor for electrophysiological signals.

foil is designed to be easily woven into fabric such that it is present in an unobtrusive way. Moreover, the flexing capability ensures that optimum capacitive coupling is maintained when the underlying human body surface has a curvature, which is often the case. A flex foil capacitive sensor e.g. woven into the clothing will then still continue to have good capacitive coupling with the body. A rigid shape would prevent flexing, so the curvatures of the human body will not be followed, leading to poor capacitive coupling, which deteriorates the signal strength.

The sensor can be fitted to regular EMG/ECG/EEG equipment without problems. The only difference is that the amplifier needs a small +7.5V/–7.5 V DC power supply. Presently this is because of portability insights provided by four flat rechargeable Li-polymer batteries.

An illustration of a controlled ECG measurement with the capacitive flex foil ECG sensors is shown in Figure 5. A comparison has been made between the conventional gel electrode attached to the skin and a flex foil sensor on top of a cotton shirt.

It can be observed that the flex foil sensors are capable of reproducing a signal comparable to the gel electrode measurements, however they are more prone to distortions. The latter effect is clearly seen between 69 and 70 seconds, where after 70 seconds the signal recovers to the gel electrode signal.

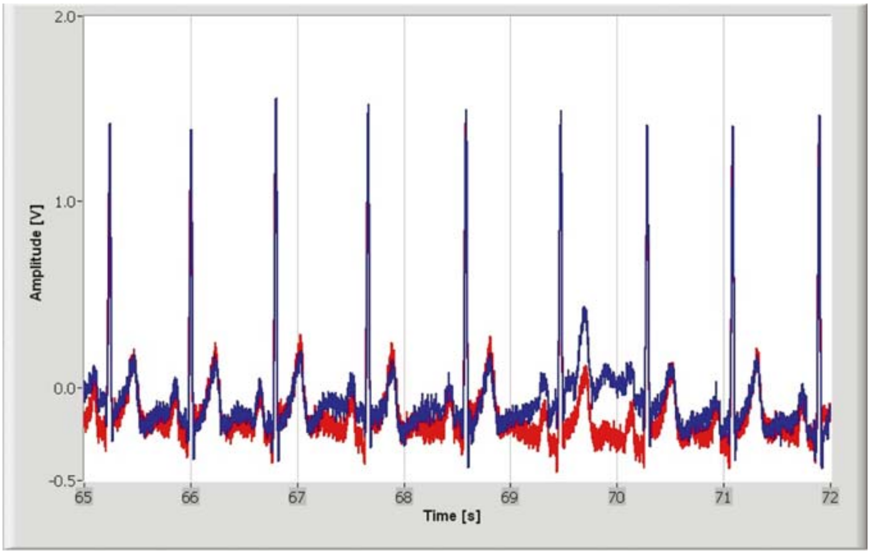

Figure 5. Comparison of a gel electrode measurement (red line) with sensors on the skin and the use of a capacitive flex foil sensor (blue line), where the sensors are put on top of a cotton shirt.

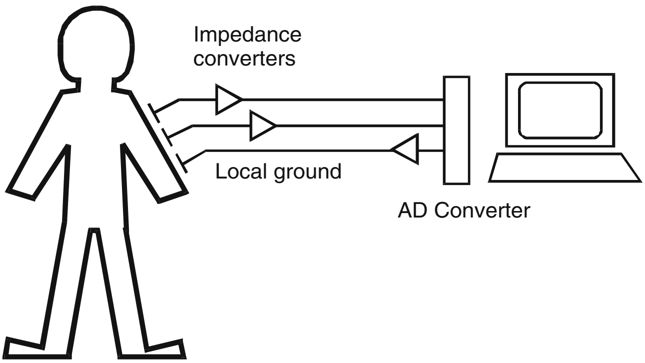

Figure 6. Set-up for contactless sensing of EMG signals.

2.5 Contactless sEMG

Figure 6 shows the bipolar set-up with two contactless electrodes with which the first EMG experiments are successfully performed. The electrodes are the same as used in the previous paragraph.

Figure 7 shows the measured EMG signal on the biceps while lifting a weight of 2.5 kg with a 90 degrees bended arm. The contactless sensors have

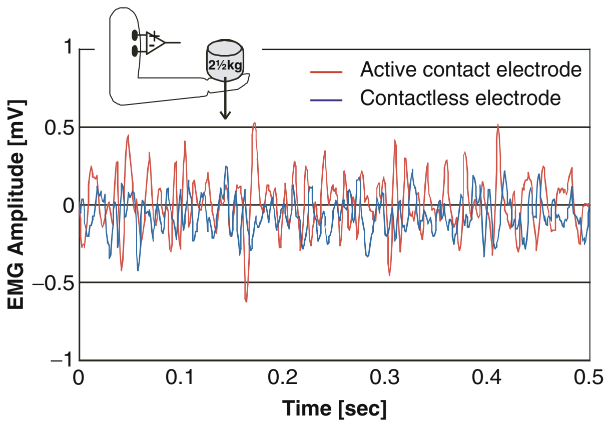

Figure 7. Recorded EMG signals with both an active- and contactless electrode.

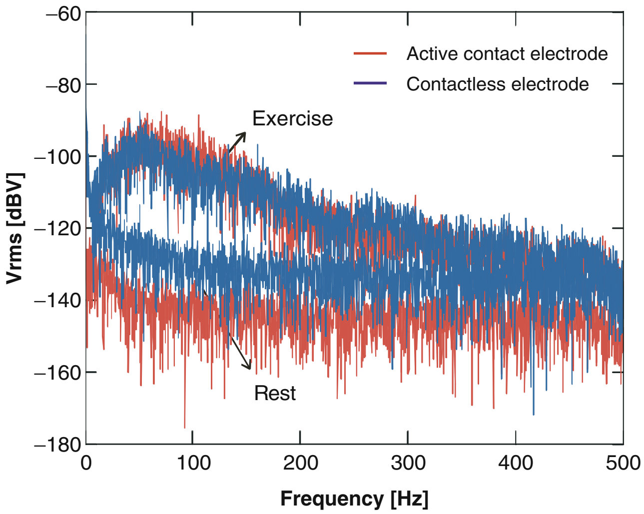

Figure 8. Spectra of the recorded EMG signals with both an active- and contactless electrode.

an electrode spacing of 37 mm and a gain of 11. It is compared to a commercial active sEMG electrode (B&L Engineering type BL-AE-N) having a spacing of 20.6 mm and a gain of 346. In Figure 7, the signals are normalised by the gains to give the skin surface voltage. Note that it is not possible to perform the two recordings simultaneously.

In Figure 8, the spectra are shown for the two measurement methods using the same data set as in Figure 7. As a reference, the spectra during rest are plotted as well. We can see that the contactless electrodes and the commercial active sEMG electrodes provide similar signal levels and shapes. Only the bottom noise during rest is a little bit higher. The bandwidth of the contactless electrode set-up is adequate and comparable to the reference measurement.

2.6 Conclusion

We have shown the feasibility of contactless sensors for monitoring surface EMG and ECG. Contactless sensors will open a whole new world of application areas because unobtrusive monitoring in our daily environment is enabled. In the next sections, we will see examples of applications.

3 Heart rate variability sensing from furniture: Sense chair

3.1 Heart rate variability sensing

The heart rate signal is known to not only give an indication of the physical state of the person, but also of his/her mental state. Well-being, attention and cognitive capabilities reside at least partially in the brain and from there they have an impact on the physiology of the body. Specifically, they influence the heart signal by varying the interval between the heart peaks in the ECG. In a study by Stephen H. Fairclough (2005), the authors monitored the EEG, ECG, EOG and respiration of thirty subjects over a learning period of 64 minutes. A multiple regression analysis revealed that specific psychophysiological variables predicted learning at different stages on the learning curve. The performance of two groups of children on a selective attention task was described in Althaus et al. (2005). EEG and cardiac activity were continuously recorded during the tasks. Decreases in specific spectra of heart rate variability were found to correlate with the degree of extraversion and task performance and with the children's temperament. It was observed that changes in attention are reflected in changes in the relative involvement of the sympathetic and the parasympathetic nervous systems on the heart. The sympathetic nervous system is known to increase heart rate, whereas the parasympathetic nervous system decreases heart rate. As a consequence, the R-R peak interval changes when the level of attention shifts. Attention is basic to and needed in many human activities. The results of the previously mentioned studies, however, were derived only after the collection and post-collection analysis of vast amounts of physiological data. It would be desirable to be able to produce indications of attention changes in real time and with minimal analytical complexity.

Probably the first application of a so-called Cepstrum technique to link with mental state is thought up by Dan Winter (http://www.soulinvitation.com/clinicalintro). He invented a real time measurement device called the Heart Tuner, which is said to measure and display moments of empathy and bliss.

The same technique could be used as a sensing algorithm capable of on-the-fly detection of attention shifts or lapses within two seconds of occurrence. By taking the Fourier transform of the last five-to-eight seconds output of ECG sensors a power spectrum can be obtained. A regular pattern of peaks occurs in this spectrum when a person has undivided attention (i.e. attention for just one subject) in this time period. Taking an inverse Fourier transform of the logarithm of the magnitude of the Fourier transform, often called Cepstrum (Bogert et al., 1963), yields in that case a strong peak at the R-R peak interval. Attention shifts or attention lapses break the short-term regularity of the R-R intervals, causing the “cepstrum peak” to (almost) disappear within two seconds.

3.2 Sense chair application

In the Ambient Intelligence (AmI) vision we anticipate a world in which technology is integrated into the fabric of everyday life (Aarts and Marzano, 2004). The present-day availability of powerful, low-energy sensors with small dimensions enables us to embed sensor networks into furniture. In this study we investigate concepts for the design of a smart chair for the purpose of sensing psychophysiological information, this information will be ultimately used to interact actively with changes in the environment.

3.2.1 Technology The aim of the Emotion-sensing Chair is to sense human physiological parameters, such as heart rate (ECG) or muscle strain (EMG) in a non-invasive manner. Small flat sensors on a flex foil have been developed which capture the ECG and EMG signals by means of capacitive coupling. The method of sensing ECG via capacitive coupling allows for non-skin contacted measurements and as such the sensors are placed behind the textile fabric of the upholstery of the chair. Capacitive coupling thus makes it possible to measure physiological parameters (e.g. ECG) in an unobtrusive and unnoticeable way. The flex foil capacitive sensors are invisibly present in the back of the chair. The high sensitivity of the sensors allows the measurement of the heart and muscle signals through both the textile of the chair and the clothing of the person. As soon as a person sits down, sensing is activated and heart and muscle activity data can be extracted and analyzed by appropriate algorithms. These algorithms e.g. link the heart rate via a wireless link to the light colour, illumination level, or the type of music that is playing.

Figure 9. Preparing the sense chair. Amongst the sensor the capacitive groundplane is clearly visible.

3.2.2 Feasability A first prototype of the Emotion-sensing Chair was built using the Philips Research flexfoil capacitive sensors in a standard chair; see Figure 9. The chair contained sensors in the back and the bottom of the seat as well as in the armrests. A pad in the bottom was applied to ground the signals collected by the sensors.

The sensor output was collected using standard computer equipment. The idea is that this high sensitivity combined with smart noise reduction measures results in a high reliability in capturing the ECG and EMG signals.

A good example of obtained sensor data is shown in Figure 10, where the person is sitting in the chair wearing a cotton shirt with the sensors placed behind the upholstery of the chair. The peaks corresponding to the heart beat are clearly discovered in this graph, although the signal is only about 30–40 mV.

Additional digital signal processing can be used to obtain extra information. For instance, the onset of fatigue can be interpreted by analyzing muscle firing frequencies, or even emotional or mood changes can be derived from values such as the heart rate variability (HRV). The sensitivity of the HRV to emotional changes is obvious, but to ensure a correct and accurate interpretation requires careful study with users in a realistic context. A promising application domain for the Emotion sensing Chair is that of “Active Relaxation”. By means of biofeedback, users can for instance be supported in achieving a more relaxed state.

One project demonstrator uses the capability of the chair to sense the heart rate of a person sitting in the chair and make the beating of the heart audible. Additionally, music is played with the same rhythm, as an alternative form of

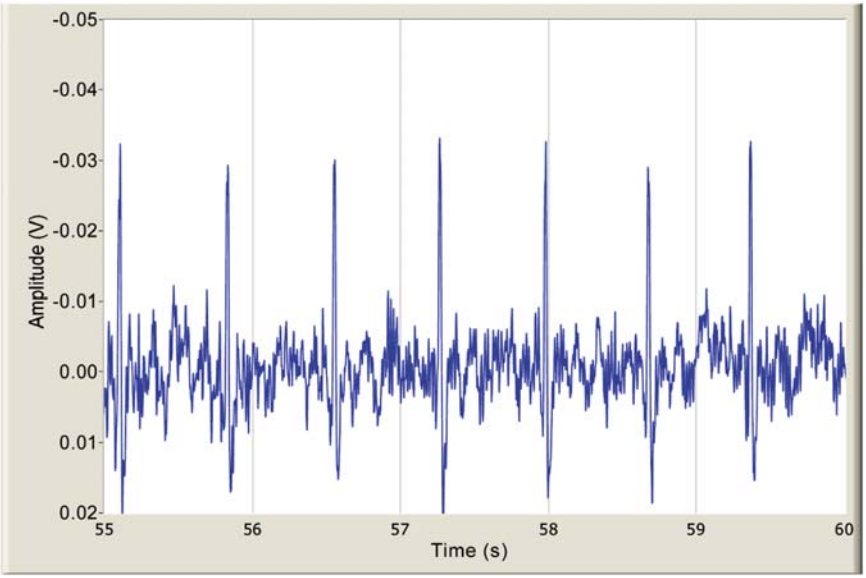

Figure 10. Sense chair sensor output with sensors behind heart and under lower right arm, using the capacitive ground plane below the legs.

biofeedback. It is envisioned that slowly decreasing the tempo of the music can help people to relax both physically and mentally. With the project demonstrator in Home Lab, the effectiveness of explicit and implicit biofeedback, such as music, can be investigated.

Feasibility studies in a realistic environment, as provided by Home Lab, showed that apart from the electrical signals emanating from the person sitting in the chair a multitude of other signals are detected by the system. The most prominent is the 50 Hz signal caused by the mains power lines. These signals are actually impinged onto the human body of the person sitting in the sense chair from elsewhere in the room, causing the sensors to couple into them through capacitors. Notch filtering eliminates most of this noise problem. A different problem that is much harder to tackle is interference from other people present in the same room. Motions of other persons cause charging and discharging effects in the person sitting in the chair, which in turn are picked up by the capacitive sensors. The resulting signals turned out to be of high amplitude and are almost impossible to filter out. As a result the signal quality of the ECG and EMG signals reduced severely. Another manifestation of the induced charging/discharging effects due to motion of people present in the same environment is that the signals could get so strong that the 50 Hz mains signal blocking notch filters were turned off-sync, because the signal went out o the operational range of the amplifier. The only way to solve the problem of induced charging/discharging of moving persons and objects in the vicinity of the person sitting in the sense chair is when the static charge of the moving persons is reduced by counter-measures, such as ion showers. However, the latter can be regarded as unacceptable from a user point of view.

3.3 Concluding remarks

The studies with the Emotion sensing Chair showed that it was feasable to build a chair that could remotely sense biometrical information of an individual, such as his ECG and EMG. However, putting the chair into the real-life environment of Home Lab, where there are many interfering signals resulting from the presence of electronic equipment as well as other persons, buries the captured body signals in noise to a degree that they become immeasurable. This calls for the development of sophisticated noise reduction techniques. After this problem has been resolved, the question will remain how people will react to the feedback given by the chair and how they will use this information in their daily lives.

4 Textile integration and the ConText project

An excellent embodiment of a health monitoring device is a wearable system, for example a health awareness system in cloths. Cloths are natural possessions and are part of the processes and routines in our daily life. The technological drive is to integrate sensors and electronics in textiles in such a way that the usage and advantages of cloths are maintained. A high level of textile integration has to be combined with aspects of reliability, comfort and wash resistance.

The ConText project (http://www.context-project.org/) was initiated to develop a wearable vest with sEMG sensors for constant monitoring of muscle activity. It is based on contactless sensors integrated into textile, hence the name “ConText”. The vest measures muscle activity in order to derive the psychological stress level of a person. The first contactless sensors were the ones developed in the heart monitoring projects as described in section 2.4. By exploiting the research in a European project, we could benefit from the project partners to facilitate integration into textiles.

This chapter first describes the market of medical monitoring systems for home use. Next, a short introduction of the topic of stress is given after which the technical challenges of textile integration are summarised. The chapter ends with some measurements using textile integrated sEMG sensors.

4.1 Personalised health and musculoskeletal disorders

The healthcare market is currently subject to a structural change. Healthcare expenditures are currently focused on professional consultancy after an individual has become ill. This is an expensive system due to the high costs of medical examinations, interventions and hospital beds. One solution to reduce the healthcare costs is to put our efforts on the pre-intervention chain from health management, prevention, self-diagnosis to home monitoring. The technologies to enable this are gathered under the personalised health and electronic health monitoring concepts: pHealth and eHealth. The health monitoring tools are no longer solely prescribed by the professional but more and more the individual patient can buy health monitoring equipment in the shop. The World Health Organisation requires that every country develops a pHealth strategy within four years (Jean-Claude Healy eHealth technologies for the citizen centred health system).

A health monitoring tool like the ConText vest is an answer to this call. The initial class of diseases to be addressed are the musculoskeletal disorders (MSD). MSD occur when there is a mismatch between the physical requirements and the physical capacity of the human body. The term MSD refers to a group of disorders with similar characteristics like myalgia and tendonitis (Armstrong, 1993). MSD are caused by a combination of factors such as repetitive motion, force exertion, psychological and physiological stress, vibration and bad posture. For example, high physical and mental job demands can cause work-related musculoskeletal disorders (WMSD). WMSD have both personal consequences, such as discomfort, pain, malfunctioning and disability, and socio-economical consequences such as reduced productivity, reduced performance and absenteeism (http://wmsd.org). Over 40 million workers in all sectors are affected. The most common occurences are Repetitive Strain Injury (RSI) and lower back pain. They are responsible for forty to fifty percent of all work-related ill-health and lead to losses of 0.5 to 2% of GNP per year. The problem is noticed by the European Commission and reported in two memo's (European Commission press release IP/04/1358; http://agency.osha.eu.int/publications/reports/201/en/index.htm). The EU Advisory Committee on Safety, Hygiene and Health at work emphasizes that a number of measures should be taken to enable successful prevention of WMSD.

A tool to measure the pathomechanisms of MSD could be very useful. Wearable electronics can assist appropriate medical management by early diagnosis of symptoms and detection of MSD for enabling prompt treatment and proper rehabilitation.

4.2 Measuring stress

The scope of the ConText project is to investigate the influences of fatigue and stress on muscle activity. The human body is experiencing stress, whether this stress is pure physiological or psychophysiological, as a threatening situation. A strong hormonal reaction is provoked: the level of cortisol and norepinephrine in the body is augmented. This increase from the stress hormones brings the body in a condition with increased alertness where the surviving instinct of the body is augmented: increased heart rate, increased blood pressure, and increased muscle tension. This reaction is the so-called ‘fight or flight’-reaction of the body. The secondary needs (emotions and thinking) are turned of by the body as they are not needed. Recent research did establish a relationship between stress and increase in muscle activity. It has been shown by Westgaard et al (1987) that an additional complex mental task to a postural load increases the muscle activity significantly in the M. Trapezius pars ascendens. Note that it is difficult to distinguish the muscle activity from postural load and the increased muscle activity from stress. Interpretation of the signals may be assisted by electrocardiography (ECG) signals and the output of and accelerometers.

The development of a stress algorithm would enable a bio-feedback signal that allows the reduction of stress at work or in other applications like sports, and revalidation specifically or WMSD in general.

4.3 Technical challenges

The textile industry in Europe sees electronic and functional textiles as a good opportunity for new growth, in view of the competition from the Far East. The textile industry is very focused on innovation as a means of increasing turnover and profit. However, one should realise that garment manufacturers are still very conservative. They are not interested in participating in the development process, although they are interested in incorporating the electronic devices once the development is completed and proven successful.

Apart from electronic textiles there is also a lot of development in passive, functional textiles. These textiles have special properties that give them extra functionality, for example textiles with antibacterial coatings that decrease the chance of infection in hospital or textiles with very high strength used in construction. A combination between active electronic components and passive functional textiles can be very interesting.

The link between personalised health and wearable electronics defines the technical requirements on electronic textiles. The most important are:

- Higher level of **textile integration**. Many products and demonstrators still contain metal wiring and big rigid blocks with the electronics. Further textile integration will make the devices more comfortable for the user and more “ambient intelligent”.

- For wearable devices the **power supply** should be wireless and wearable. At the moment, batteries that are used are too heavy and too large. Flexible solar cells are an option, but are not so commonly available yet and are technically not good enough yet.

- **Connections** between flexible textile and rigid electronic components. Electronic textiles are now a combination of rigid ICs and other electronic components combined on a textile substrate. The connection between the two is one of the major points of failure. Consider the ConText project in which contactless electrical bio signal monitoring electrodes are used. A direct consequence of capacitive contactless sensors is that an electronic impedance converter must be placed as close as possible to the electrode. Therefore we must solve the interconnect problem because we can no longer put the active electronics in a removable box on the belt. The STELLA project focuses on technology for flexible substrates and stiff electronics.

- Overall robustness of the system, especially the **washability**. Mechanical stability upon stretching and flexing.

These technical challenges will in the end result into electronic functions integrated in our daily environment. These electronic functions, both sensing and actuating, will change our daily live by making professional medical and psychological guidance available. So, besides developing technology, it is extremely important to do structured research on what continuous and unobtrusive monitoring can do. An example of a project where the application development is balanced with technology research is described in the next paper.

4.4 Textile EMG sensors

The sensor as shown in Figure 11, was developed with the Fraunhofer Institute (IZM) from Berlin (Linz et al., 2007). It consists of an embroidered electrode on a woven substrate. The electrode is electrically shielded by a metal

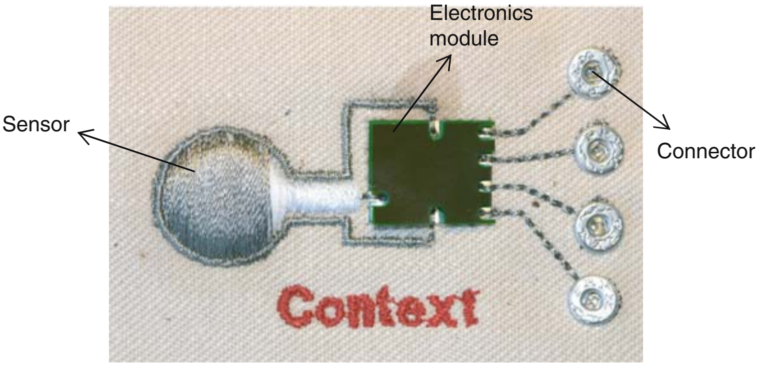

Figure 11. Embroidered Sensor with interconnection to the electronic module and snap fasteners as interface to the computer (Fraunhofer IZM Berlin).

cup which is embroidered as well. The module with electronics is electrically connected in the same embroidery process.

To evaluate the embroidered capacitive transducers, without having the problems of motion artefacts and noise of the human body, an artificial muscle model was used. Figure 12 shows the hardware model. The muscle itself is emulated by a strip of moderately resistive paper. By two aluminium beams, an electrical current can be forced through this muscle. On top of the muscle, a leather chamois is used which mimics the human skin. On the chamois we can put several types of textile on which in its turn the sensor is placed. So, the model does imitate the

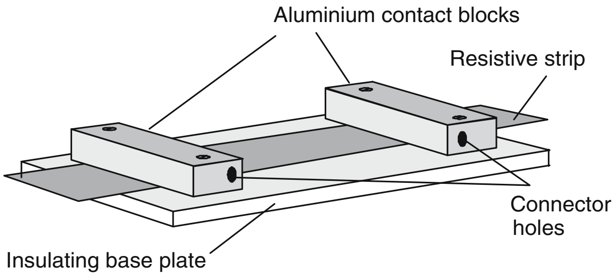

Figure 12. Artificial muscle model.

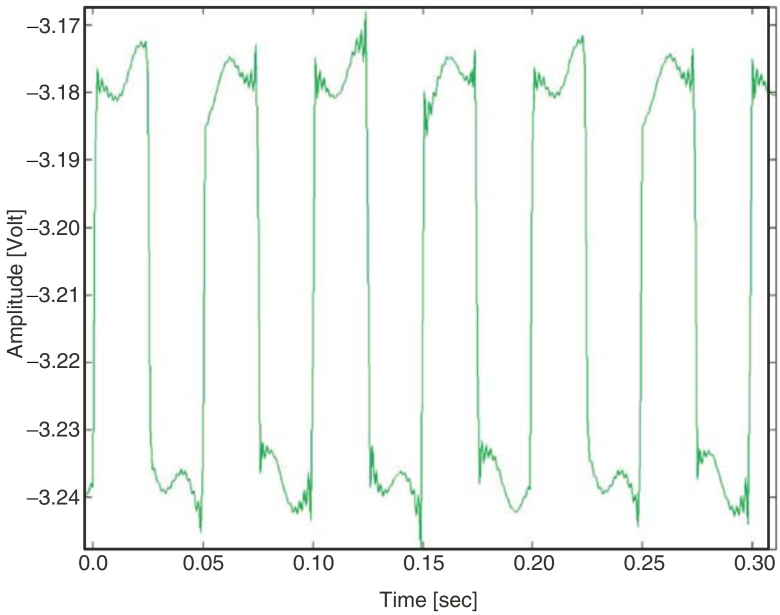

Figure 13. Square wave recorded in capacitive manner.

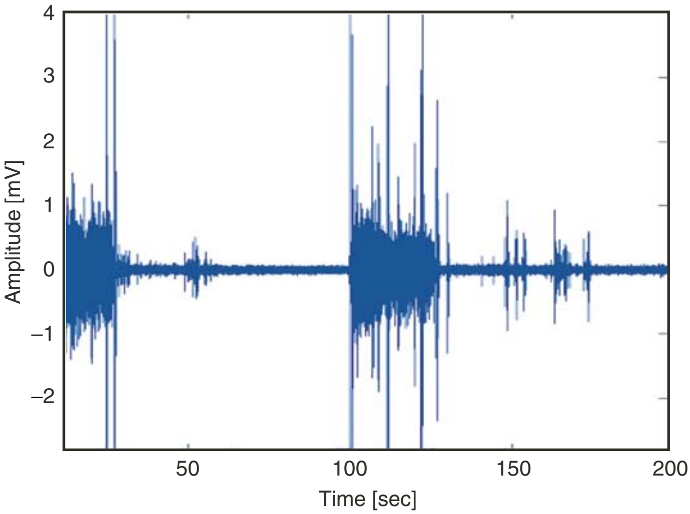

Figure 14. Bipolar EMG on a human biceps recorded in a capacitive manner.

contactless behaviour and the distributed shape of a buried muscle, but does not include the human tissue volume conductor properties.

A waveform generator was connected to the artificial muscle. A square wave of 1 Volt peak to peak with a frequency of 20 Hz was generated. Figure 13 shows the recorded signal by using a single embroidered sensor on the artificial muscle. Note that the envelope of the recorded square wave shows a 50 Hz noise signal. This is the result of the single-sensor approach. By using a single sensor with respect to a grounded reference, we will see 50 Hz noise as picked up capacitively from the environment. This will be cancelled when using the set-up of Figure 6.

In Figure 14, the human EMG is measured on the biceps using two electrodes of the type of Figure 11. At t = 0 sec and t = 100 sec, a contraction of the biceps was applied. We can see that muscular activity is clearly detected by the textile embroidered sensors. Some motion artefacts are visible as spikes on the signal.

Further technology to be developed is a suitable method to integrate wiring for power and signal lines. Options can be found in the printing or weaving of conducting wires (Gimpel et al., 2003; Gimpel et al., 2004).

4.5 Conclusions

In the European project ConText, the expertise and skills of partners are combined to develop a muscle activity monitoring vest. The vest anticipates to a set of work related diseases, which are an increasing financial burden to our society. To exploit the natural usage of clothes, contactless EMG sensors are chosen. A direct consequence is that pre-amplifiers have to be mounted directly on the textile sensors. This results in the need for advanced interconnect technologies for textile integration.

5 Miniature wireless sensors

A typical way to achieve unobtrusive sensing is to make the sensor devices small, thin and wireless.

Research on the subject of small autonomous network devices at Philips Research finds its origin in a lecture on Ambient Intelligence (Aarts and Marzano, 2004) given by Emile Aarts for the Integrated Device Technologies group in March 2002. In this lecture a picture of ubiquitous computing was sketched with the example of swarms of embedded micro devices forming an electronics ambient, which senses people and reacts to people. The original thoughts on such a concept came from Mark Weiser from Xerox published in Scientific American (Weiser, 1991). The article starts with the sentence:

"The most profound technologies are those that disappear. They weave themselves into the fabric of everyday life until they are indistinguishable from it."

Although this reference is from 1991, the vision has until now not become reality. Miniature wireless sensor devices as small as a few cubic millimetres have been shown to be feasible, mass market introduction is still lacking. In general a severe reduction in size leads to a severe limitation in device functionality. In designing miniature wireless devices requirements such as fast reaction times, bi-directional wireless connectivity at medium data rates, long operational lifetimes, lead to relatively complex devices and moderate power levels. For instance most devices that are being commercialized use a wireless standard, such as 802.15.4, with security overhead and other features. Energy scavenging schemes, such as vibration energy scavenging, and thermal energy scavenging have been shown to have a poor power density and poor miniaturization possibilities, rendering them unsuitable for powering miniature wireless sensor devices except for some small exceptional niche applications. Photo voltaic energy scavenging is applicable in a wider range of applications, but miniaturized photo voltaic systems are still far from mature. In order to enable a long enough operational lifetime the battery volume needs to be sizeable. The energy density of Lithium rechargeable batteries (i.e. the batteries with the highest gravimetric energy density) determines in that case the size of the device. As an example the energy density of a LiR2430 80 mAh/3.6 V battery is 200 Wh/dm^3. Even if the battery volume is allowed to take 50% of the volume of the device, the volume reaches a value close to 3 cubic centimetres, far removed from several cubic millimetres.

In the literature several attempts have been made in order to have a sound comparison of the state-of-the-art wireless sensor systems (Anliker et al., 2004; Beutel, 2006; Hill, et al., 2004; Römer and Mattern, 2004). Various aspects have to be taken into account to make such a comparison, if possible at all. Aspects influencing the size of wireless sensor devices are the requirements and restrictions imposed by production. It can be concluded that if the manufacturability by present day production facilities is required the design needs to fulfil a host of requirements, almost all leading to size constraints.

As such one of the first problems in making a good comparison is the difference between packaged and unpackaged sensor devices. Many of the wireless sensor devices are offered in an unpackaged fashion, some even without battery holder or antenna. A product incorporating all these aspects as well as being small has yet to evolve. In Figure 15 we present an overview of some of the existing solutions for wireless sensor devices. The choice has been made to compare the Philips Research miniature wireless sensor device solution with three unpackaged solutions and three packaged versions that are commercially available and that are often cited in the wireless sensor devices community. The reason to take as well unpackaged solutions is because of the absence of a variety of packaged solutions let alone smaller. So benchmarking the level miniaturization and integration from electronics point of view should be done on the basis of comparison with unpackaged

Figure 15. Comparison of the Philips solutions with different devices often cited in the wireless communications community.

Table 1. Clarification of the functionality of the sensor devices shown in Figure 15.

Philips Button	**Philips Cylindrical**
Processor CoolfluxDSP Wireless 2.4 GHz IEEE802.15.4 3D accelerometer, 3D magnetometer Antenna, battery, package included 28 mm diameter, 10 mm height	Processor CoolfluxDSP/PCH7970 Wireless 2.4 GHz IEEE802.15.4 3D accelerometer, Antenna, battery, package included 14 mm diameter, 14 mm height
Intel iMote	**IMEC (1st generation)**
Processor ARM7TDMI Wireless 2.4 GHz Bluetooth Sensors excluded Battery excluded 30×30 mm (W×L)	Processor MSP430F149 Wireless 2.4 GHZ, Nordic Temperature sensor Battery included 14×14×12 mm
Moteiv	**Xsens**
Processor MSP430F1611 Wireless 2.4 GHz IEEE802.15.4, Zigbee compliant 2D accelerometer, temperature sensor, light sensor, microphone, loudspeaker battery included 50×94×22 mm (W×L×H)	wired system to wireless hub 3D gyroscope, 3D accelerometer, 3D magnetometer No battery 58×58×22 mm (W×L×H)
Crossbow MICAz	**Crossbow MICA2DOT**
Processor Atmel Atmega 128 L Wireless 2.4 GHz IEEE802.15.4, Zigbee compliant Light, temperature, sound, 2D accelerometer, 2D magnetometer Battery included (W×L×H)	Processor Atmel Atmega 128 L Wireless 868/916, 433 MHz or 315 MHz Temperature sensor No battery/antenna included 25 mm diameter, 10 mm height

solutions. The different devices depicted in Figure 15 are plotted such that the scale puts them in good perspective with each other.

From Figure 15 it is clear that the Philips devices offer a very good solution, especially because it also contains a package. The latter property makes it a valuable tool to allow for testing even in an outdoor environment. Obviously, in order to have a sound comparison we need the additional information about what is inside the package. In Table 1 we shortly list the functionality of the different devices in order to validate our current approach.

5.1 Design considerations

In designing a miniature wireless sensor device a number of parameters need to be considered. The application sets constraints and requirements for the device. Especially for wearable sensors the volume, shape and weight are important factors. These factors will be discussed in relation to a number of applications in the area of personal wearable wireless sensing in the next sections.

For a given application aimed at sensing psychophysiological parameters related to experiences the use case needs to be considered in terms of operational lifetime, duty cycle, accuracy and stability of sensor output. In operator assisted laboratory tests short operational lifetimes combined with full time sampling, high accuracy will prevail. In attempts to record these parameters in a daily life situation care must be taken to render the devices wearable. This use case dictates long operational lifetimes, small duty cycles and high sensor stability, sacrificing sampling rate and accuracy. The most challenging use case scenario is 24 hour round the clock monitoring, where even the incorporation of sensors into clothing may fall short of the goal (since some people like to sleep naked), and implanted or skin mounted devices are necessary.

Daily life monitoring falls within the scope of this paper, which is about unobtrusive and unnoticeable sensing. Incorporation of sensors into furniture, such as chairs and beds are an attractive option, but for full time monitoring they are obviously unusable. The authors are of the opinion that implantable devices have a level of invasiveness exceeding current mainstream acceptance. The necessity of skin worn devices or skin contacted devices is unavoidable in some cases, such as when measuring the galvanic skin response. The devices embedded into clothing certainly are the champions of unobtrusiveness, as long as weight, shape and volume stay within limits. These limitations are the subject of the following section.

5.1.1 Device volume In the device volume of the electronics, the package and the battery are the main contributors. Let's postulate a 5 cubic centimetre device volume as the maximum volume allowable for unobtrusive use. Relevant literature concerning this topic seems to be lacking. The volume of a modern (women's) watch is therefore seen as a reasonable assumption. The battery volume of such a hypothetical device is linked to the average power consumption and the desired operational life. In Figure 16 the relation of these parameters is shown for a 2 cubic centimetre rechargeable Li ion battery, taking up 40% of the available volume.

The average power to run the device for a day, week or year would be 15 mW, 2 mW or 40 μW respectively. Needless to say the choice of required operational life may pose severe restrictions on the choice of electronic components, sampling rate and duty cycle. With current state-of-the-art technology the collection of simple sensor data such as temperature or resistance may be feasible at reasonable sample rates to allow a device to be made to function for a year or even more. In a totally different field, such as automotive tire pressure monitoring such miniature wireless sensor devices are already a common product, operating up to 5 years (not continuously, because

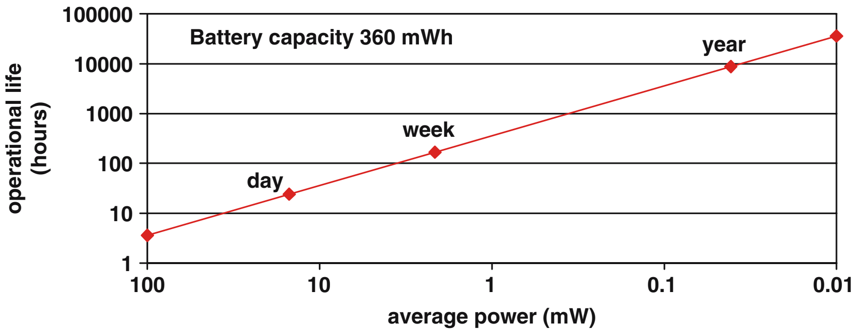

Figure 16. Operational lifetime as a function of average power for a 360 mWh battery capacity.

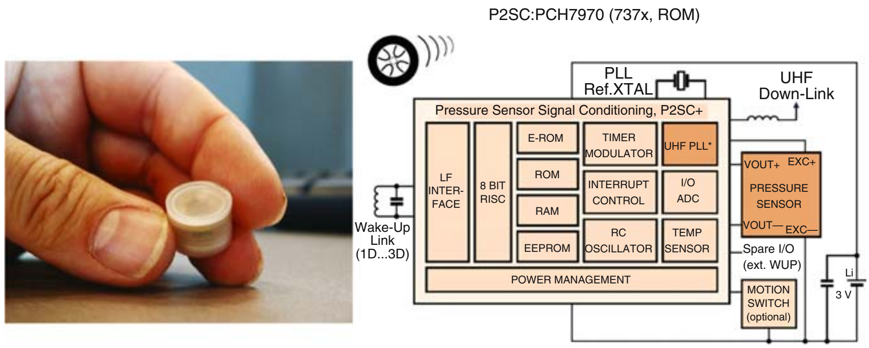

Figure 17. Miniature wireless temperature and pressure sensor based on NXP tire pressure sensor chip PCH7970.

a car is driven intermittently) on a single battery. Reuse of this hardware for body temperature and galvanic skin response monitoring is a distinct possibility. Philips Semiconductors (now NXP) developed a tire pressure monitoring product based on the microcontroller PCH7970 (http://www.nxp.com/acrobat_download/literature/9397/75015738.pdf). The block diagram of this chip is shown in Figure 17.

A tiny wireless temperature and pressure sensor for body sensing purposes has been made by Philips Research and is also shown in Figure 17 Since the pressure sensor as well as the high precision temperature sensor both are resistance monitors this device can be easily be used for galvanic skin response purposes.

The volume of the device is about 1 cubic centimetre and it runs continuously for 3 weeks at a sampling rate of 1 Hz on a simple 3 V Li CR1220 battery.

5.1.2 Shape considerations For miniature wireless sensor device which are to be worn on the body or in the clothing a number of shapes can be considered. Most desirable is a thin and highly flexible device, built nearly indistinguishable into the fabric of the clothing. The EMG devices discussed in the previous chapters are coming close to this form. A format which is mostly used for identification purposes is the smartcard. Already now electronics, displays and batteries are built into these cards (Chan et al., 2004). A possible next step is the incorporation of sensors. Activity and posture monitoring sensors and temperature sensors are candidates for this. The card may function as hub in a body area network, having a wireless link to other body worn sensors, fulfilling a function in data fusion, data processing and storage. Using a display it may even function as user interface. Other shapes to be considered are the button shape, the cylindrical shape and the pencil shape. Shapes with sharp edges, such as a box shape or a cube shape fail to meet the unobtrusiveness criterion due to the increased chance of being noticed by the wearer.

The pencil shape may double as a writing utensil, which most people tend to wear anyway. The high aspect ratio is a limiting factor for the capability of integrating electronically components. Some components, such as the antenna, or an AAA-type battery fit very well in such an elongated shape. The Button and cylindrical shapes have actually been made into prototypes and these will be discussed in more detail in the next section. In Figure 18 the considered shapes for unnoticeable and unobtrusive sensing of psychophysiological parameters for experience assessment are shown. The shapes are shown with the same volume, indicating aspect ratio differences.

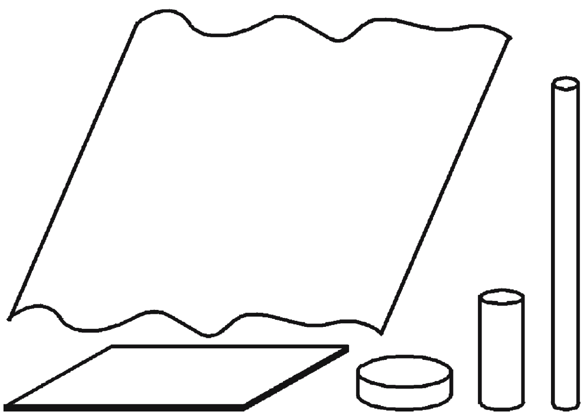

Figure 18. Various shapes for miniature wireless sensor device embodiments: fabric, smartcard, button, cylinder and pencil shape. The shapes are shown with the same volume, indicating aspect ratio differences.

Table 2. Various shapes which are suitable for miniature wireless sensor devices compared with the same volume

Shape	Dimensions for a 5 cm^3 volume
Fabric	0.2 mm × 160 mm × 160 mm
Smartcard	1.1 mm × 85 mm × 54 mm
Button	7.1 mm height × 30 mm diameter
Cylinder	34 mm height × 14 mm diameter
Pencil	100 mm height × 8 mm diameter

If a 5 cm^2 miniature wireless sensor device volume is taken as a starting point the aforementioned shapes can be translated into sizes that are tabulated in Table 2.

When taking a closer look to the values as given in Table 2 it can be argued that the different shapes all correspond to a particular use case. The size for *Fabric* is indeed such that it allows for unobtrusive use in clothing, the *Smartcard* dimension is identical to the currently used identification badges used by various companies and as such it has been proven to be an acceptable size. The *Button* shape is according to the authors suitable for application development testing of particular sensor combinations for 24–48 hours experiments in e.g. the medical and healthcare domain. The latter shape also corresponds nicely to the requirements for gaming applications. The *Cylinder* shape has the advantage that a specific electronics module can be used for each of the main components, but turns out to be the least wearable solution. For implantable and swallowable devices however the advantages are obvious. The *Pencil* shape may be easily integrated in writing utensil and the length could even be increased for this application. However, from an implantable and swallowable point of view the length is problematic, while the diameter is satisfactory.

5.1.3 Weight considerations As is the case with the optimal wearable volume without being obtrusive or noticeable, the optimal wearable weight of a miniature wireless sensor device is somewhat hard to define. Again looking at devices people tend to carry around without trouble, such as jewellery, timepieces, mobile phones the weight of a watch is seen as a reasonable criterion. Although watches may weigh as much as 100 grams a small sized (woman's) watch weighs about 10 grams. For unnoticeable and unobtrusive this is subsequently taken as the maximum allowable weight. In the case the devices are to be worn as jewellery there are deviating criteria.

The button shaped miniature wireless sensor device that recently have been realized within Philips is shown in Figure 19.

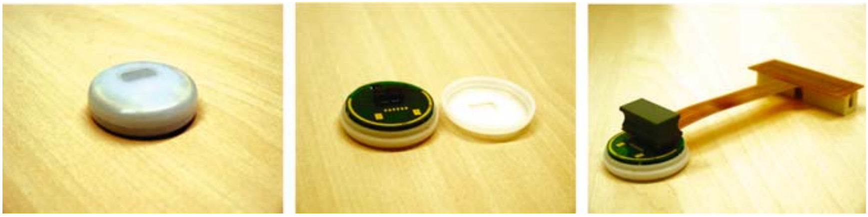

Figure 19. Button shaped miniature wireless sensor, showing top connector for additional modules or software upload/data download.

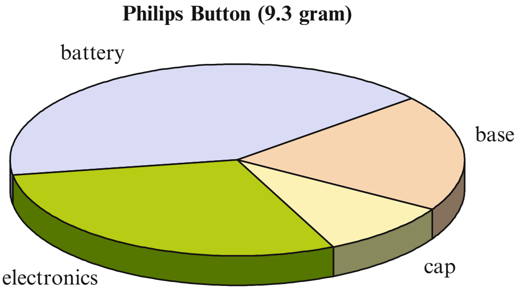

Figure 20. Weight Breakdown Phillips Button device with a diameter of 30 mm and a height of 9.7 mm.

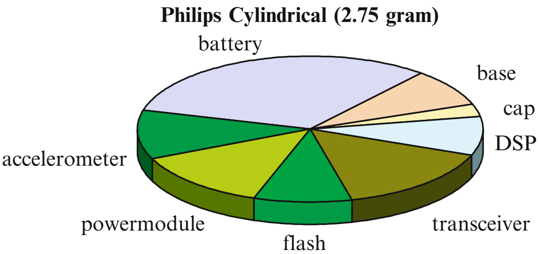

Figure 21. Weight breakdown for Philips Cylindrical device, with a diameter of 14 mm and a height of 20 mm.

In Figure 20 and Figure 21 a pie chart diagram is shown where a weight breakdown for the two unobtrusive miniature wireless sensor devices. From these figures it is clearly observed that indeed for both devices around 30% of the weight is taken by the batteries (even a bit more for the Philips Button device). Regarding the Philips Button solution, around 30% is taken by the package which is because the packaged of this device has been designed with a focus on robustness. For the Philips Cylindrical device (Figure 17) a larger part of the weight is taken by the electronics, due to an innovative stackable packaging concept.

For both the cylinder and the button device the electronics are the same in the weight breakdown comparison: in both cases the device is a motion sensor.

With respect to the impact of device weight on unnoticeability it can be remarked that due to inertia the maximum allowable weight will be less for parts of the human body that are prone to move at high accelerations. Also the ration of the device weight to the body part weight is important: Consider for example heart rate variability sensing using a photoplethysmograph: if the device is to be used as ear mounted jewelry this poses severe restrictions to the weight.

5.2 Special requirements for miniature wireless devices

This paragraph deals with some so-called special requirements that improve the quality of the final sensor devices when taken into account. Because of the fact that we deal with body worn sensor devices, skin contact is quite often the case. When sensors are skin contacted, irritation of the skin can be an unwanted result, or even an allergic reaction to the sensor devices. This is exactly one of the reasons why capacitive contactless ECG/EMG sensors have been developed. Another issue that strongly relates to sensing the ECG/EMG is the skin resistance, conductive gels are to be used in order to decrease the skin resistance and to have it a more or less constant value. Moreover the devices should be electrically safe, i.e. without problems regarding wireless emissions or external electrical fields. The devices should obey the rules regarding current IEC601–1 standard on human safety (Modi, 1997; http://www.tnocertificationmedical.nl/).

Another aspect is the battery, the choice for rechargeable batteries or primary batteries. The good thing of rechargeable batteries is the environmentally more friendly aspect. The battery is not immediately thrown away. This battery aspect is especially important when the considered sensor devices are only used for 24–48 hours experiments. However, when the application combines a long service life scenario and low power consumption simple primary coin cells such as the zinc-air type, offer a much larger energy density. These batteries are not capable of handling a high discharge rate. Recharging of the batteries is closely related to the so-called *Ease-of-Use* of the sensor devices. The word *Ease-of-Use* is used in the sense of a common denominator in what follows. One of the preferred ways of recharging the batteries is when the device is simply plugged on top of a connector. But this would mean that the device has an opening for the connector which out rules the use in water. The other option would be wireless recharging in a cradle, by means of induction coils, which is compatible with a completely closed package to be used in a watery environment as well.

The use in or close to a watery environment (i.e. the human body) in the sensing of psychophysiologal parameters offers an important requirement for the transceiver. The attenuation of electromagnetic radiation by water quickly rises when frequencies exceed several hundred megahertz. Especially for implantable miniature wireless sensors, or sensors which have to transmit through a human body, the choice of frequency is limited to bands below 1 GHz, such as the 402 MHz band for implantables, or for other devices the IST bands at 433, 868 and 916 MHz.

Another aspect regarding recharging of the batteries in close relation to *Ease-of-Use* is related to the topology of the wireless sensors network. Imagine the wireless sensors to be used in a large network. It is clear that the batteries should be rechargeable, and moreover, an energy scavenging option should be included because there is no possibility to change all the batteries one-by-one. Apart from the recharging options which influence the packaging of the devices the *Ease-Of-Use* also translates back in terms of possible outdoor use. Regarding sensors that are integrated in the clothing *Ease-Of-Use* obviously means that washability is the key requirement, and when met, this would most probably mean that outdoor use of the particular clothing is possible as well.

Furthermore *Ease-Of-Use* regarding software upgrades in a large wireless sensor network it is a common understanding that the possibility of using a wireless link to achieve the software upgrade should be included. When the wireless sensors are not used in a large sensor network structure, but for application development, software upgrades could be realized via a connector which should easily accessible from the outside. An example of such a connector can be seen in Figure 19.

6 Concluding remarks

The research topics on psychophysiological sensing and miniature wireless sensor devices described in this paper show that within Philips there is a growing interest in the measurement of experiences, emotions and moods. The application areas suitable for such information are in the domain of wellness, motivation, persuasion, illness prevention and personal healthcare.

The interpreted sensor output can be used as such: offering information about oneself or relatives, friends. The value may be in how the information is conveyed or used: lighting or audio/video content may adapt to a person's mood or even attempt to improve upon it.

The technology no longer seems to be a limiting factor for unobtrusive and unnoticeable sensing. Initially the sensors will be worn on the body, but ultimately implantable sensors will become widely accepted, allowing access to new parameters, such as hormone levels and body core temperature.

The increased computing capabilities allow real time interpretation of complex data, such as facial expressions, data fusion of multi parameter sensor inputs.

The information on experiences, emotions, and moods will be used not only for a single person, but increasingly in interpersonal relationships. Ultimately hard to fathom aspects such as group feelings that occur in a highly coherent group may come into reach.

In interpersonal relationships uncertainties about how others react to you, how popular are you can be resolved, enabling feedback to improve in case of shortcomings.

Personal health indicators addressing issues such as chronic stress and mental health are becoming available for preventive healthcare in the case of 24 hours daily life monitoring of psychophysiological indicators.

All these fine future prospects have one major requirement: the obtaining of unbiased true psychophysiological sensor data relies on true unobtrusive and unnoticeable monitoring.

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