A Flexible Wearable Sensor Network for Bio-signals and Human Activity Monitoring

Ruben Dias
INESC TEC (formely INESC Porto)
Porto, Portugal
Email: rtdias@inescporto.pt

José Machado da Silva
INESC TEC (formely INESC Porto)
Faculdade de Engenharia, Universidade do Porto
Porto, Portugal
Email: jms@fe.up.pt

***Abstract*—The work presented herein addresses the development, implementation, and evaluation of a new wearable system for monitoring bio-signals and physical human activity, namely for gait analysis and cardiovascular surveillance. It consists of a wearable textile substrate (pantyhose and/or T-shirt) with embedded conductive yarns interconnecting custom electronic devices, in a mesh or other network type, that acquire bio-signals and/or inertial data. All data are aggregated in a central processing module from where they are sent via a wireless link to a mobile phone or personal computer for final processing. The network topology, sensor nodes architecture and results obtained with first prototypes are presented.**

***Keywords*-Body sensor network; EMG, ECG; wearable.**

I. Introduction

In order the increased life expectancy indicators that have been witnessed [1] can be assured with adequate quality of life, it is imperative to find economic and efficient healthcare procedures that allow for greater patients' autonomy in outpatient care.

Wearable systems built on textile platforms represent a convenient solution for outpatient care, provided these are comfortable, non-invasive and require no technological skills [2]. One type of information that commonly needs to be monitored is patients' physical activity together with correlated bio-signals that provide feedback on the respective health condition, namely Surface Electromyography (sEMG) and electrocardiogram (ECG) signals.

Both wireless and wired networks can be found in WeaSN applications [3]. Wireless solutions are often preferred for their flexibility in terms of sensor nodes deployment and scalability, but these imply higher power consumption, which is a critical aspect on a system where the number of batteries and bulky components is to be kept to a minimum [4]. Wired networks provide higher speed, better reliability and allow for communications with lower power consumption [5].

A flexible wearable sensor network (WeaSN) that can be used for gait analysis and cardiovascular surveillance is being developed (Fig. 1a). It is meant to be implemented on a fabric substrate with the purpose of capturing data in a practical and non-invasive way, even for people with strong impairments or disabilities. Two applications are being targeted: an e-legging to capture human locomotion

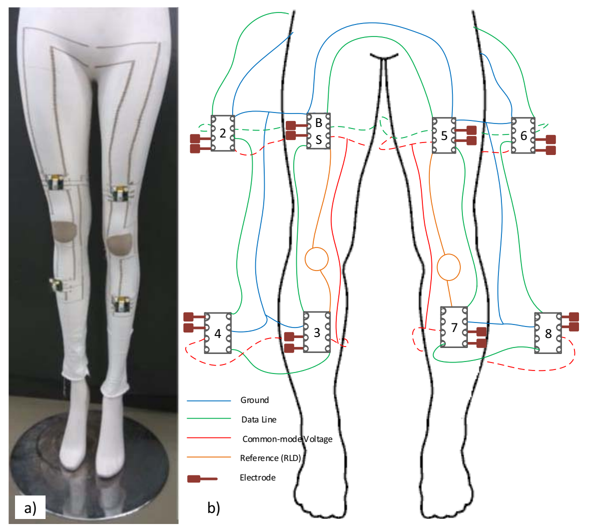


Figure 1. e-Legging: (a) Photograph of the prototype; (b) Diagram of interconnections.

parameters and a T-shirt to capture ECG, respiratory rhythm, and pressure in the abdominal aorta after an endovascular repair procedure. Textile conductors are embroidered in the fabrics to provide the electrodes to capture EMG and ECG signals, as well as the conductors which interconnect the Sensor Node (SN) in a wired mesh network.

Section II describes the network topology and main functional characteristics. Section III presents the hardware architecture and section IV the network performance when capturing data. Experimental results relative to the capture of data for gait analysis and for cardiorespiratory monitoring are presented in section V and section VI highlights the main conclusions.

II. WeaSN for gait analysis

A. e-Leggings

The system is based on a leggings with elastic properties, which allows the correct positioning of the textile electrodes and electronics. The legging is meant to be comfortable, implying an adequate combination of materials, compression effect, and electronic components incorporated in the textile and interconnected with data and power tracks made with

978-1-4799-6136-8/14 $31.00 © 2014 IEEE
DOI 10.1109/BSN.Workshops.2014.20

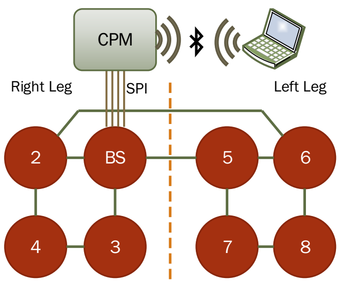


Figure 2. System data connections.

textile conductors. This piece of garment is meant to be easy to dress, reusable, and cleaned and maintained with traditional methods.

It is possible to measure bio-electric potentials using conductive fibers or yarns instead of conventional (dry and wet) electrodes [6]. To successfully produce fabrics with textile conductors, and particularly to build surface EMG electrodes, yarns made with twisted filaments, each one a polymeric filament covered with silver with linear resistances of about 30-40 Ω/m, where used. A suitable but complex routing of the tracks is necessary to knit the yarns without them crossing each other. This is where the capabilities of the production machines assume a critical importance, justifying the seamless technology that was adopted [6]. The machine is capable of drawing a pattern where the conductive yarn is meant to be placed, thus ensuring that the tracks will not intersect.

B. Interconnections

Given the vulnerability of textile conductors and the number of sensors used, a mesh topology was adopted to organize the SN. With this topology higher reliability is obtained due to the presence of redundant links. Fig. 1b) shows all interconnections implemented in the e-legging prototype. Each SN has at least two paths with a neighbor SN through which collected data can be routed. Each knee acts as a sEMG common-mode reference point (Right Leg Drive (RLD)) for the circuits placed in that leg. The common-mode voltage of the four circuits present in one leg is averaged in the RLD bias circuit of one of these SNs. In total, each leg shows a textile conductor to interconnect all common-mode voltages, a ground connection, the data lines and one RLD (with a second redundant option).

The mesh topology allows a fully distributed network configuration. However, as it is not possible to cross conductive yarns using a single fabrics layer, the number of possible interconnections that can be knit in the legging is limited. The solution reached for the SN interconnections is shown in Fig. 2.

One critical aspect is power management. The SNs may be powered by small batteries or by energy distributed from the Central Processing Module (CPM) through the conductive yarns. In the first prototype, each SN has its own power battery, but the final objective is to have the SN harvesting power from the mesh network, being power line communication used to convey data. For that purpose a specific transceiver has been developed [7].

C. Routing Protocol

The system should work for long periods of time, especially during prolonged monitoring. Thus, an energy-efficient Source based Routing for Minimum Cost Forwarding (SRMCF) routing protocol, described in [8], was specifically developed and adopted for the network data layer. In this protocol there is a base node (BS) that collects all the captured data from the SNs. These use the minimum cost forwarding method as the routing algorithm for sending acquired data to the BS. Communication from the BS to a specific SN over a minimum cost path typically requires intermediate nodes to have stored information about the minimum cost routing paths. With the SRMCF protocol the routing path is carried in the packets header, avoiding the need for the routing information to be stored in intermediate nodes.

In the proposed WeaSN the communication energy budget is 0.68 nJ/bit, while for the Bluetooth link it is 231 nJ/bit. As data sent from the different sensors is gathered in the CPM and sent in bursts using a single Bluetooth transceiver, energy is saved as the number of bits transmitted by each SN is lower. Notice that each Bluetooth frame comprises an access code and a 54-bit header which in this case are needed only for the CPM.

D. Time Synchronization

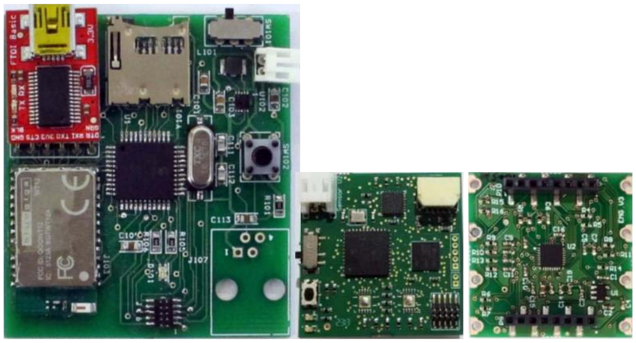
To properly synchronize the data of different SNs a time stamping technique is used. This consists in recording the event time using a marker synchronized with a central clock source. The method used here is based on getting clock information from the node which is in the minimum cost path to the BS node. That is, each node sends a request to the aforementioned neighbor node and receives a reply with a message including the real time clock. The main clock source is in the BS node. When a SN wants to synchronize with the network it sends a quarry message to BS and this replies with its time [9].

Each data packet generated by a SN includes the time stamping information of the first sample. In the final data processing this information is then used for time synchronization, extrapolating the time of the remaining captured samples according to the known sampling frequency.

III. Hardware

A. Sensor Nodes and Central Processing Module

A preliminary e-legging prototype was designed for a WeaSN devoted to capture sEMG signals and kinematic information from the lower limbs. The network comprises



(a) (b) (c)

Figure 3. Hardware prototypes: (a) CPM; (b) *SensorV2*; (c) *EMGV3*.

eight (four per leg) SNs and one CPM. All these modules were custom developed to present the functional characteristics specifically required for this WeaSN.

The CPM module (Fig. 3a) receives data from the SNs and sends them to an external computer via a Bluetooth connection. This module includes a 16-bit microcontroller (μC, PIC24FJ64GA104), is equipped with a USB port and a MicroSD card to record data whenever wireless communication is unavailable, and shows a 50 x 55 mm^2 area. This version of the CPM does not has the hardware needed to perform the SRMCF routing protocol, leading to the need of the BS functions to be executed in a SN. In a final version that capability will be available also in the CPM, allowing to replace the current serial peripheral interface (SPI) connection with the same type of SN-to-SN interconnection, improving thus the network reliability.

Each SN comprises two modules with an area of 30 x 30 mm^2: *SensorV2* (Fig. 3b) and *EMGV3* (Fig. 3c), fitted on top of one another. *SensorV2* includes a μC, a FPGA (Actel AGLN250), an inertial measurement unit (IMU, the InvenSense 16-bit MPU-6050 accelerometer and gyroscope) and a three-channel shunt voltage monitor. The *EMGV3* module is the sEMG acquisition circuit based on the low power 24-bit Texas Instruments ADS1291 chip. Vertical pin connectors are used to fasten the *SensorV2* board on top of the *EMGV3* board.

The oscillator of the μC (16 MHz) sources also the FPGA clock. The μC implements the networking and application layers, and acquires the sEMG and inertial data. The physical and MAC layers of the data routing protocol are implemented in the FPGA (Fig. 4). The power supply is a small (5.7 x 28 mm^2) LiPo battery of 3.7 V and 110 mAh. A DC-to-DC converter generates the 3.3 V power supply voltage for the μC, IMU and *EMGV3* module. To supply the FPGA and line driver circuits a 1.5 V linear regulator is used.

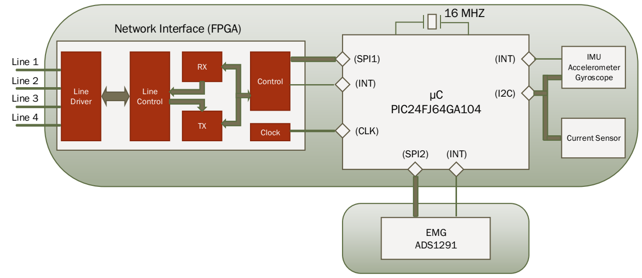


Figure 4. Sensor node architecture.

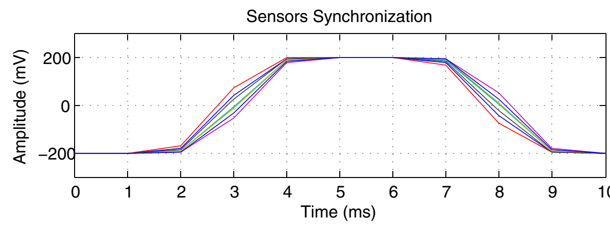


Figure 5. Sensor nodes synchronization.

IV. Data Acquisition

A. Network Capability

By default the system configuration is 1 ksps for sEMG acquisition in eight SNs and 50 sps for the IMU in four SNs (the four IMU units in the back of the legs have off, due to information redundancy). Different configurations are possible, being the data communication restriction limit (1 Mbps) set by the serial communication with the Bluetooth link. The baud-rate between SNs is 4 Mbps but can be increased to a maximum of 9 Mbps, even with the higher impedance textile conductors [8]. The SPI communication between the BS and CPM can be performed at 8 Mbps.

B. Sensors Synchronization

The synchronization among SN is crucial as the respective information has to be correlated. The accelerometer and the gyroscope are intrinsically synchronized as they are integrated into the same chip. To synchronize the sEMG signal with the IMU data, as well as the information among SNs, the time stamping from the routing protocol is used as a reference. The time stamping resolution is 1 ms, but can be increased if necessary.

To test the data synchronization process a 100 Hz square wave was simultaneously captured with the eight sensors (Fig. 5) one of which is captured by the BS (clock reference). It can be seen that the delay between the BS and the remaining SNs is shorter than 1 ms, which is less than the time stamping and the sampling period.

V. Lab experiments

Two systems have been developed: one is meant to capture data of the lower limb activity (ProLimb), and the other to

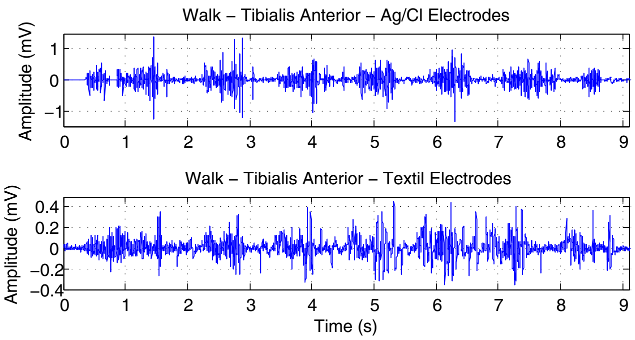


Figure 6. EMG signals obtained with the ProLimb network when (top) the leggings; (bottom) gel electrodes, are used.

capture the ECG and respiration rate (SIVIC).

A. ProLimb system

Two types of locomotion exercises, gait and isometric, were performed to validate the ProLimb WeaSN.

Using a first prototype of the e-leggings, where textile electrodes with an area of 5 x 10 mm^2 separated by 10 mm had been sewn, being the SN interconnected with textile wires, the EMG signal shown in Fig. 6 was obtained for the *Tibialis Anterior* muscle — the wearer performed 5 steps. This figure shows the same signal obtained when gel electrodes are used. It can be seen that the signal obtained with the textile electrodes show a worst signal to noise ratio (SNR) but still allows us to identify the occurrence of the bursts that characterize the muscle activity. Notice that the signals were obtained in different experiments, with different electrodes placed in not exactly the same points and thus the two signals cannot be point-to-point compared. The use of textile conductors to interconnect the SN did not corrupt data transmission. Tests performed to evaluate bit error rates (BER) revealed that using a 1 m copper wire, a BER of 10^{-20} is obtained with a total jitter of 0.05 UI, while a BER of 10^{-7} is obtained with a textile conductor (the parallel of four 1 m yarns with R=1.5 Ω/cm, L=8 nH/cm) at the same jitter value.

The sensor network was then used without the integration in the leggings and with conventional electrodes, so that it could be used simultaneously with a commercial system comprising several Trigno (from DELSYS) wireless sensors.

The SNs were placed in the following muscles of the thigh and leg: *Vastus Intermedius, Rectus Femoris, Biceps Femoris, Long head, Short head, Tibialis Anterior* and *Gastrocnemius Medialis*. The experiments followed a preparation protocol including skin preparation and placement of the electrodes by a physiotherapist. For the sEMG electrodes, cut Ag/AgCl electrodes were used to obtain a separation between the respective centers of 2.5 cm.

Each Trigno module shows a volume of 37x26x15 mm^3, comprises an EMG channel and a triaxial accelerometer.

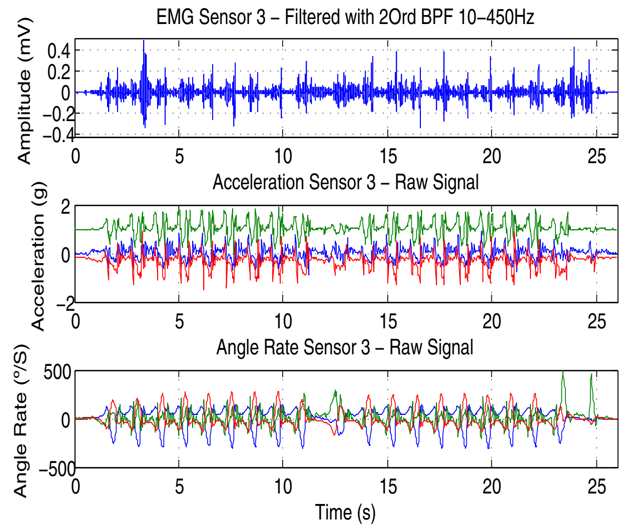


Figure 7. sEMG, acceleration and angular rate signals captured with the proposed WeaSN during gait exercise.

Each channel has 16-bit resolution and sampling frequency of 4 kHz for EMG and 300 Hz for the accelerometer. The modules include their own sEMG electrodes with a separation of 1 cm, but need adhesive to fix the modules to the person's body. In the experiments eight Trigno modules were placed as near as possible to the conventional electrodes in order the two systems could monitor the same muscles activity.

In the gait exercise thirty-six steps round-trip were performed in a straight line. Fig. 7 shows the kinematic data and EMG signals captured by the SN placed on the *Tibialis Anterior* muscle of the right leg. It can be seen from Fig. 7 and 8 that similar information is provided by the two systems. As these signals were captured in different muscle points and with different systems, a one-to-one comparison of the two signals cannot be performed. Muscular activity is usually evaluated after the analysis of the captured sEMG signal bursts, i.e., relevant information on the specific time and amplitudes which define bursts onset and offset. It can be seen that the ProLimb WeaSN allows capturing a sequence of bursts which are clearly distinguishable among them and which occur timely synchronized with the kinematic data, allowing for clearly characterizing the muscle activity.

The isometric exercise measurements were made with the left leg *Tibialis Anterior* muscle within a System 4 Biodex dynamometer. The performed exercise corresponds to a unilateral maximum voluntary contraction with the foot in a neutral position with zero degrees flexion. Fig. 9 shows the sEMG signals captured by both systems.

Again, although one cannot directly compare the two sEMG signals, since the systems are independent, have different configurations and are not placed exactly in the same place, it can still be seen that muscle activity patterns

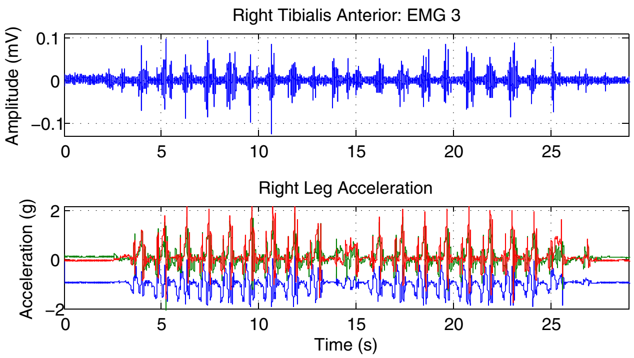


Figure 8. sEMG and acceleration signals captured with the Trigno modules during gait exercise.

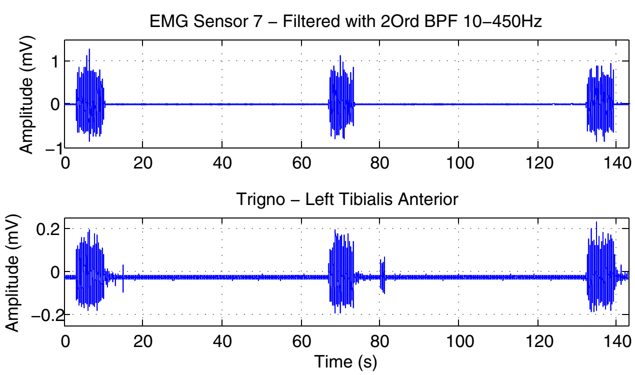


Figure 9. sEMG data captured during a isometric exercise with the ProLimb system (top) and with the Trigno system (bottom).

can be correctly detected and that the two signals present similar information in terms of onset, offset and SNR.

B. *SIVIC system*

The ECG and respiration rate measurements were performed using the same hardware but instead of using the *EMGV3* module, a new signal acquisition module (*EMGV4*- Fig. 10a) based on the ADS1298R from Texas Instruments was used. The difference between the two modules is that the *EMGV4* module provides eight input channels allowing the measurement of a 12-lead ECG.

Now the experiments consisted of using a chest elastic band (Fig. 10c) with three snap buttons for the electrodes placement, necessary for a 1-lead ECG measurement. With this configuration the hardware performance and the ECG signal quality were evaluated with three different sets of electrodes (Fig. 10b): conventional Ag/AgCl electrodes, textile electrodes with a diameter of 1.7 cm and textile electrodes with a diameter of 1.2 cm.

A male subject without previous reports of cardiac problems performed the experiments. The measurements were carried out in a seated position without skin preparation and with a stabilization time of five minutes after electrodes placement.

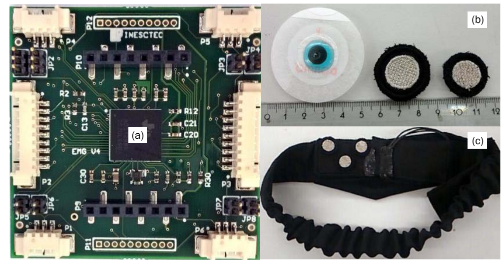


Figure 10. (a) ECG acquisition board (4 x 4 cm^2); (b) Electrodes used in the experiments; (c) Chest elastic band.

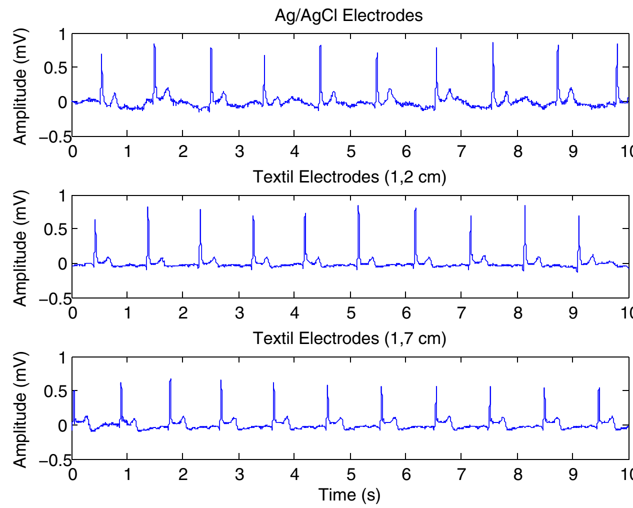


Figure 11. ECG signal acquisition with conventional and textile electrodes.

Fig. 11 shows the ECG signals captured with the three sets of electrodes. A sampling frequency of 1 kHz was used and the signals were filtered with a high-pass filter with a cut-off frequency of 0.5 Hz, for baseline wandering removal. In the graphs it is possible to verify that similar SNR are obtained with the gel and the 1.2 cm diameter textile electrodes. The signals obtained with the 1.7 cm diameter textile electrodes show a lower amplitude due to their larger area and rigidity that do not allow for a proper fit to the body anatomy.

C. *Respiration rate*

To capture the respiration rate two methods were used: impedance pneumography (IP) and capacitive electrodes.

The IP based respiration rate measurements were made with the ADS1298R integrated respiration circuit, using two electrodes to measure changes of the thorax electrical impedance occurred during breathing. A high-frequency AC current is injected into the tissue causing a potential difference to be developed across any two points between the electrodes. This potential difference is related to the resistivity of the tissue between the electrodes.

The capacitive electrodes (Fig. 12a) based respiratory rate detection consists of measuring the variation of the amount

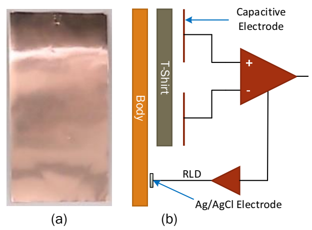


Figure 12. Capacitive respiration rate measurement.

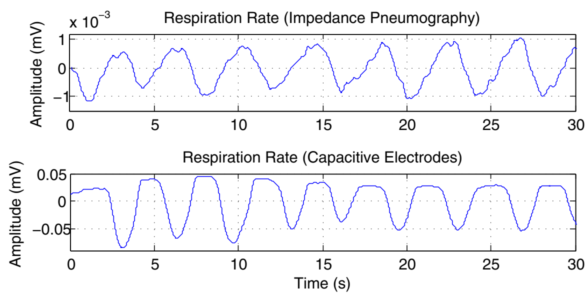


Figure 13. Respiration rate detection based on impedance pneumography (top) and capacitive electrodes (bottom).

of energy accumulated by a conductive surface, caused by the expansion of the person's thorax. In this experiment, two 5 x 5 cm^2 copper strips placed on the person's chest, over a cotton T-shirt, were used together with the placement of a gel electrode in the hip to establish the body reference, as shown in Fig. 12b).

Fig. 13 shows the signals obtained (after equal filtering) with the two methods. The measurements were made at different times, but as it is observable, the respiratory rate can be perfectly detectable in both cases, whereas the signal provided by the copper strips shows a higher amplitude variation and is less intrusive.

VI. Conclusion

This paper describes the operation and characteristics of a wearable sensor network being explored to develop more comfortable and easier to manipulate wearable data capture systems to capture human bio-signals and physical activity, namely for human locomotion and cardiorespiratory data capture. Conductive yarns sewn in the fabrics are used to wire-up sensor nodes, as well as to realize surface EMG and ECG detection electrodes.

Experimental results obtained in a biomechanics laboratory, for a real-time human locomotion data acquisition, show that the proposed WeaSN performance is equivalent to that of a commercial system, with the advantage of consuming lower wireless transmission power, using either textile and gel electrodes. It provides also the measurement of an additional quantity - angular velocity.

Other experimental results are presented for the use of the developed sensor nodes in the acquisition of an ECG and respiratory rate. The ECG has been captured with conventional and textile electrodes showing that these can be used to obtain signals with the similar signal to noise ratio of that obtained with conventional gel electrodes. In the capture of respiratory rate impedance pneumography and capacitive electrodes methods have been used, the latter providing a less intrusive for the wearer approach.

The final wearable system is meant to be a garment with embedded textile electrodes and sensor nodes interconnected with conductive yarns and supplied from a single battery placed in a central processing module.

Acknowledgment

This work has been financed by the ERDF - European Regional Development Fund through the COMPETE Program (operational program for competitiveness) and by National Funds through FCT - Fundacão para a Ciência e a Tecnologia (Portuguese Foundation for Science and Technology) within projects PROLIMB PTDC/EEA-ELC/103683/2008 and SIVIC PTDC/EEI-ELC/1838/2012. The authors acknowledge the collaboration of U. Minho - Departamento de Engenharia Têxtil (ProLimb) and Departamento de Ciência e Tecnologia Têxteis - U. Beira Interior (SIVIC) in the design of the textile fabrics and electrodes.

References

[1] OECD. Health at a glance 2013: OECD indicators. [Online]. Available: http://dx.doi.org/10.1787/health_glance-2013-en

[2] L. Piccini, O. Ciani, and G. Andreoni, "New emerging biomedical technologies for home-care and telemedicine applications: the sensorwear project," in *New Developments in Biomedical Engineering*, D. Campolo, Ed. InTech, Jan 2010.

[3] W. Tao, T. Liu, R. Zheng, and H. Feng, "Gait analysis using wearable sensors," *Sensors*, vol. 12, no. 2, pp. 2255–2283, Feb 2012.

[4] J. Xing and Y. Zhu, "A survey on body area network," in *IEEE 5th International Conference on Wireless Communications, Networking and Mobile Computing*, 2009, pp. 1–4.

[5] H. S. Lee, "Wearable personal network based on fabric serial bus using electrically conductive yarn," *ETRI*, vol. 32, no. 5, pp. 713–721, Oct 2010.

[6] A. Catarino, A. M. Rocha, M. J. Abreu, F. Derogarian, J. da Silva, J. Ferreira, V. Tavares, M. Correia, and R. Dias, "E-legging for monitoring the human locomotion patterns," *Journal of Textile Engineering, The Textile Machinery Society of Japan*, vol. 59, no. 6, pp. 153–158, Set 2013.

[7] J. Carvalho and J. Machado da Silva, "A transceiver for e-textile body-area-networks," in *MeMeA 2014, 9th IEEE International Symposium on Medical Measurement*, Jun 2014.

[8] F. Derogarian, J. C. Ferreira, and V. M. Tavares, "Design and implementation of a circuit for mesh networks with application in body area networks," in *Digital System Design (DSD), 2012 15th Euromicro Conference on*, 2012, pp. 896–901.

[9] F. Derogarian, R. Dias, J. C. Ferreira, V. M. G. Tavares, and J. M. Silva, "Using a wired body area network for locomation data acquisition," in *27th conference on Design of Circuits and Integrated Systems (DCIS 2012)*, Avignon, France, Nov 2012.