Smart Sensors for Ubiquitous Health Monitoring of Wheelchair Users

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Abstract—This documents describes the development of a healthcare monitoring solution prototype for wheelchair users. Unlike most of healthcare monitoring solutions available, the presented system intends to introduce the use of standardization in smart biomedical sensors by employing the IEEE 1451.4 standard. These smart sensors are connected in a wired network fashion to an embedded acquisition and processing system on the wheelchair, i.e. a microcontroller. PPG (photoplethysmography), BCG (ballistocardiogram), and skin conductance sensors are responsible to acquire biological signals from the user in an unobtrusive way. PPG and BCG provide extraction of the heart rate by using signal processing techniques such as digital filtering and peak detection. Together with skin conductance it is possible to qualitatively determine the user’s stress. RFID technology is used for identification of the wheelchair user through and RFID tag. All the data acquired by the system is then transmitted through wireless communication (IEEE 802.15.4) to a coordinator computer. The coordinator computer runs a developed graphical user interface for visualization and storage of the transmitted data, both in the local computer on in a remote webserver. Hence, a electronic health record of the patient can be build. By empowering the wheelchair object with smart sensors and wireless communication technology, it is possible to create a network of smart wheelchairs that could be used in nursing home, clinical facilities, or even in a hospital.

Index Terms—smart sensors, biomedical sensors, IEEE 1451.4, TEDS, e-Health, digital filters, smart wheelchair, skin conductance, photoplethysmography, ballistocardiogram.

I. INTRODUCTION

With the continued cost reduction of electronic hardware and the introduction of new electronic devices each year, it has been possible to develop a variety of solutions for healthcare. The interest in these area has grown over the last decade, with the development of new monitoring systems and biomedical sensors. Nevertheless, pervasive computing currently plays a minor role in institutions like nursing or supported living facilities for the elderly, but this situation could rapidly be changing [1].

Ubiquitous and pervasive technology allow patients to receive services such as prevention, diagnosis, therapy and prognosis management, both locally and remotely, with the help of communication and information technology [2]. Elderly people and patients with physical disabilities who need to use a wheelchair in their daily life can benefit from continued use of this technology [3]. By using unobtrusive biomedical sensors which been developed in order to reduce the induced stress of the utilization on patients monitoring healthcare solutions allow patients to have an autonomous life while physicians keep track of the subject health.

Currently, most of the developed healthcare solutions lack the use of standardization. Standards need to be used not only in the high-layer of applications, like health information systems and electronic health records [4], but also in the “sensing devices-layer”, which acquire biological signals. Plug and play features in sensors are very important since no configuration is needed by the user. Standards that provide particular focus on providing both interoperability and plug and play capabilities already have been developed [5], [6].

In the sensor’s technology area, there has been a group of standards named IEEE 1451 which deals with various aspects of sensors, such as the definition of sensors and actuators, the format of data sheets and how to connect/disconnect from a system, transforming a sensor in a smart sensor. One of those standards, IEEE 1451.4, provides an interface for analog sensors enabling them with (digital) data describing the sensor[7]. The developed solution makes use of smart sensors which are compliant with the 1451.4 standard. These smart sensors are responsible to acquire biological signals PPG (photoplethysmography), BCG (ballistocardiogram) and skin conductance. Also, temperature, relative humidity sensors and an accelerometer are used for environment and motion analysis. Based on the information contained in the smart sensors, digital processing techniques are applied to sensors signals.

To do this, an appropriated measurement unit embedded in the wheelchair was developed. The measurement and processing unit is based on a PIC microcontroller. The microcontroller is responsible to acquire and realize basic signal processing on biological signals, communicate with the sensor networks and interpret its data, and send sensors data to a coordinator computer through wireless (802.15.4) or wired communication (USB). The processing of RFID information, and read and write appropriate data to a SD card are performed by the microcontroller. By using 802.15.4 wireless protocol, it is intended to establish a wireless network of smart wheelchairs that transmit data to a coordinator computer, which stores data both locally and on a remote database. A GUI (graphical user interface) application was also developed to manage the coordinator computer and remote database, and to display real-time data acquired by the smart wheelchairs.

II. PROPOSED PLATFORM ARCHITECTURE

The overall monitoring system is represented by Figure 1. The full platform consists in 3 basic tiers: sensing, data acquisition and processing, data storage and visualization.

The smart sensing part is accomplished by a network of smart sensors (IEEE 1451.4) designed to be embedded in a
wheelchair, thus creating the concept of a smart wheelchair. The sensors are connected to a MCU which has the task of acquiring sensor’s data and realize basic processing on it. Finally, the MCU data can be transmitted wirelessly to a PC (Coordinator) which runs a graphical user interface application responsible of managing the wireless network of smart wheelchairs and store the data locally or remotely. The software application also allows the user to visualize real-time data or offline saved data. A remote database and web server assures saved data visualization through a website. Development of a website was not an objective of this work, but an appropriate database was build in order to provide data access to the webserver.

A. Sensors

Biological signals acquisition is performed by three smart sensors with biomedical sensors. Namely, PPG, BCG and skin conductance sensors. On the other hand, wheelchair motion signal is provided by an 1-D accelerometer, and environment information is retrieved with temperature and relative humidity sensors.

1) PPG sensor: Photoplethysmography (PPG) is a simple and low-cost optical technique used to detect blood volume changes in the microvascular bed of tissue. An increase in blood volume within a tissue will result in an increase in the optical path length, and thus a decrease in the intensity of transmitted light. The PPG waveform comprises an AC physiological waveform attributed to cardiac synchronous changes in the blood volume with each heart beat which typically has a frequency between 0.5Hz and 4Hz, corresponding to 30 beats/minute and 240 beats/minute [8]. The pulsations in arterial blood modulate the transmitted light owing to the absorption of the red light by hemoglobin. When the blood in the artery is at its peak the absorption of light passing through it is also high keeping the transmitted light minimum. Between the pulsations when blood recedes in the artery, the absorption of the light is also minimum and a peak is produced in the transmitted light [9].

The used sensor makes use of a red LED with a wavelength of about 660 nm. Photodetection is done by a PNP phototransistor, which produces a current proportional to the detected light. To simply convert this current into voltage, a resistor connected in series with the phototransistor emitter was used. Typically, this signal has a DC component can change with various due to numerous factors explained in [8]. With the used sensor it was verified that it was is normally between 100 and 200mV. To amplify the AC component, one must first attenuate the DC component, otherwise the amplifier output will easily saturate. A simple passive high pass filter was used to perform this with a cut frequency of 0.7 Hz, which revealed reasonable results. To perform AC signal amplification while centering its DC value in a constant value, an instrumentation amplifier was used. From the output of this instrumentation amplifier, comes the signal which can be connected to the analog input channel of the MCU. The amplified AC signal is centered at about 1.65V, to make use of the full dynamic input range of the ADC.

2) BCG sensor: The ballistocardiogram reflects the mechanical activity of the heart and is one of the oldest non-invasive methods for cardiac evaluation. [10]. Is is based on sensing structures associated with different natural frequencies and damping that bring information about cardiac output [11]. To acquire the BCG signal, an electromechanical film (EMFi) can be used. This sensor generates a voltage when it is mechanically deformed, but because of its capacitive nature only the change of an external force can be measured. The output voltage change \( \Delta V \) of an EMFi sensor is defined by equation 1, where \( C \) is the sensor capacitance, \( S_q \) the sensitivity coefficient of the sensor, and \( F \) represents the impact force [12].

\[
\Delta V = \left( \frac{1}{C} \right) \cdot S_q \cdot \Delta F
\]  

(1)

In order to convert its current into voltage, the EMFi sensor needs to be connected to an charge amplifier (Figure2) . The voltage produced by the charge amplifier has positive and negative voltage values, so an equivalent scheme as for the PPG sensor was used. An instrumentation amplifier changes the reference voltage of the signal to 1.65V.
measurements a lower frequency can be used as suggested in [13]. Until up to 5 Hz allows heart rate to be evaluated.

3) Skin conductance sensor: During a sympathetic response eccrine glands in the skin produce ionic sweat lowering the resistance of the skin and increasing conductivity. The change in conductivity can be used to infer differing arousal states in individuals[14]. A skin conductance sensor can capture this activity, which, in turn reflect activities of the autonomous nervous system. However, since autonomous nervous system also controls many physiological parameters such as heart rate, respiration, blood pressure and others, skin conductance has been used along side traditional physiological parameters like heart rate variability and respiration[15]. To acquire a subject’s skin conductance, two electrodes are placed on the skin connected to the positive input of an operational amplifier. A voltage connected to the negative input of the operational amplifier acts as a reference constant voltage which is applied to the skin. The operation amplifier presents a non-inverting configuration, in which one the resistors is provided by the human skin. This provides a variable voltage gain proportional to the skin conductance.

As said, skin conductance sensing is performed by two electrodes placed on the user’s skin, in which one of them is connected to an non-inverting amplifier configuration, with a gain relation \( A_v = 1 + \frac{R_1}{R_2} \). This operational amplifier is powered with +5V.

\[
G_{\text{skin}} = \frac{V_{\text{out}} - V_+}{R_1 V_+} \quad (2)
\]

A set of feedback resistors \( R_1 \) exists with the following values: 30kΩ, 100kΩ, 300kΩ and 1MΩ. These resistors allow a maximum conductance range of 333\( \mu \)S, 100\( \mu \)S, 33\( \mu \)S and 10\( \mu \)S, respectively. These resistors can be selected by a multiplexer which in this case are controlled by the digital PIO of the DS28E04-100 EEPROM.

B. Embedded Data Acquisition and Processing System

The wheelchair’s acquisition and processing system can be interpreted as an IEEE 1451 NCAP. It is composed by a PIC microcontroller, PIC24FJ128GA010, from Microchip.

Some important features of this MCU include 128 kB of Program Memory, 8kB of SRAM, five 16-bit timers, two UART modules, two SPI modules, two I2C modules, a 10-bit ADC with 16 analog channels acquiring up to 500 kS/s, a maximum of 16 MIPS operation at 32 MHz, up to five external interrupt sources, a operating voltage range from 2.0 to 3.6V, selectable power management modes (like Sleep, Idle, Alternate Clock modes) and a Hardware Real-Time Clock/Calendar (RTCC) for calendar functions. Furthermore, external modules such as an RFID reader, a XBees 802.15.4 module and a FTDI chip are interfaced with the MCU. Figure 4 represents the internal blocks used and the interfacing with the external modules.

Two external modules are connected to the MCU: the XBee 802.15.4 module and a FTDI chip. The XBee module transmits and receives data to the available 802.15.4 wireless network of smart wheelchairs, while the FTDI allows communication through USB for a single wheelchair. UART2 interfaces with an RFID reader for user identification. The SPI2 peripheral is used for the SD Card writing and reading.

To acquire data from the connected sensors, the ADC samples eight channel at 1 kS/s rate which is enough for the sampled analog signals. When the ADC is enabled, a 16-bit timer is used to signal an interruption with a period of 1 ms.

A 11.1 V DC lithium-polymer battery with 1500 mAh capacity powers the embedded system. A 9V power supply can also be connected. A input power source connector is connected to two LDOs which permit an DC input voltage

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![Figure 3: Non-inverting amplifier configuration circuit](image)

![Figure 4: Microcontroller Platform](image)
from 9 to 15V and can provide 3.3 and 5V to the system. The MCU operates at a voltage of 3.3V since it is the required voltage of most of the interfaced modules. 5V are used for powering the RFID reader, 1-Wire signaling protocol requirement and to power analog sensors which require 5V.

A smart sensors network is connected to this system, and it will be explained in detail in chapter IV.

III. IEEE 1451.4 STANDARD OVERVIEW

The IEEE 1451.4 standard [7] defines a mechanism for adding self-describing behavior to traditional transducers with analog interface. It establishes the concept of a transducer that has both digital and analog interface, i.e., a Mixed Mode Interface (MMI). The analog interface is responsible for transmitting the signal which reflects the quantity sensed by the sensor. Furthermore, the digital interface provides communication with an embedded memory within the transducer in order to read the IEEE 1451.4 TEDS. The TEDS is a data structure embedded on the transducer that describes its identity, type, operation and attributes. The TEDS can essentially be divided into two types: Basic TEDS and Template TEDS (Figure 5). The Basic TEDS provides a simple identification of the sensor while the Template TEDS contains information about the operation and attributes of the sensor.

![Figure 5: TEDS structure](image)

The IEEE 1451.4 MMI ensures the robust transfer of a transducer signal and digital TEDS to an NCAP or data acquisition system. Protocols, timing, and electrical specifications in this standard define two classes of MMI, allowing analog and digital signals either to sequentially share a single connection or to be available simultaneously on separate connections.

Class 1 MMI allows a transducer signal and power, or digital TEDS data and power, to make alternate use of the same connection. Digital logic levels are defined as 0 V = logic state 0, and –5 V = logic state 1. In general, Class 1 MMI uses inverted logic and has a shared data connection.

On the other hand, a Class 2 MMI (Figure 6) uses separate connections for transducer signals and digital TEDS data. The return path can be shared. It allows simultaneous access to a transducer signal and digital TEDS data, and connections may share a common return. Digital logic levels are defined as 0 V = logic 0, and +5 V = logic 1. Class 2 uses positive logic and has a separate data connection. Both Class 1 and Class 2 have to use open-drain line drivers for the digital line.

For this project, Class 2 MMI operation has been chosen because it does not involve to design switching circuits for each sensor, hence simplifying the smart sensor circuit. Also, one can make use of existing sensors because a separated/external digital line can be easily add which is much more practical.

![Figure 6: IEEE 1451.4 Class 2 MMI architecture example, [16]](image)

IV. SMART SENSORS NETWORK

The embedded smart sensor network (Figure IV) is composed by IEEE 1451.4 smart sensors connected through a 1-Wire multidrop bus to the MCU. The analog outputs of each sensor are individually connected to the analog ports of the microcontroller. Such architecture allows identification of each smart sensor, and thus its function on the network. It can also allow to set controllable features of the smart sensors.

Each smart sensor is connected through a Smart-sensor Interface Board (SIM). To allow a more simple use of existing analog transducers and to establish a set of connections for the SIM to the sensor network, a interface board was designed (Figure 8). More, a SIM Extension board allows up to eight...
SIM to be connected to the sensor network. Besides providing connections to the SIM, the Extension board has built-in circuit (9) referred as NCC (Network Control Circuit), to control the 1-Wire network.

The NCC consists of two quad-analog switches that sets up the network as a 1-Wire multidrop bus or as a bus of EIGHT single point-to-point connections to an eight channel analog multiplexer. The system needs to switch between two buses topologies so the microcontroller platform can identify which of the EEPROMS is associated with which analog channel. This way it is possible to link the information of the Basic TEDS to an analog channel without having to previously program the TEDS with the measurement location of the sensor or to use an address programmable EEPROM, like an I2C memory type used in [17]. This requires 4 additional digital pins from the microcontroller: one pin to control the analog switches (SW) and three pins (ABC) to control the multiplexer selection input/output (IO) bits from 000 to 111 (0 to 7). Each multiplexer IOs is virtually associated with an analog channel. To find the 64-bit ROM from each memory, SW must be zero (point-to-point connections), and ABC should go from 000 to 111 to go through all IOs of the multiplexer. Once the ROM codes are retrieved from the memories, SW should go high (multidrop bus) and the last multiplexer ABC combination that has an EEPROM available should be selected by the microcontroller. This multiplexer channel will be responsible to transmit and receive all the multidrop bus information. All the other channels from the multiplexer go into a high impedance state. With this circuit the system becomes more plug-and-play than the system implemented in [17], despite using more digital pins. The connection between the microcontroller platform and the SIM extension board is made through a 20-pin cable. There are 8 pins to all 8 analog channels, with 3.3V voltage, 5V voltage, 1 ground, and 4 remaining pins for desired future connections. 3.3V and 5V will be used to power the biomedical sensors, conditioning circuit and the 1-wire EEPROM DS28E04-100, if used. The 1-wire EEPROM DS24B33 is powered by the data line. The connection board includes a set of 8 connectors with 5 pins associated with the maximum 8 of the system. Each SIM board connects to the SIM connection board through a 5 pin connector, which has pins for analog input, digital data, 3V, 5V and ground.

To connect transducers having +5V as maximum output voltage the analog input channels of the microcontroller, an operational amplifier attenuation scheme was implemented in each one of the sensors, to limit the output a maximum 3.3V [18]. In Figure 10 is represented the circuit responsible for this. This circuit was used in for all the transducers except for the accelerometer.

**Figure 8: Designed IEEE 1451.4 Interface Board**

**Figure 9: SIM extension board’s Network Control Circuit**

**Figure 10: Operation Amplifier attenuator circuit**

**V. EMBEDDED PLATFORM PROCESSING FUNCTIONS**

**A. Digital filters**

Digital filters are responsible for removing existing artifacts from the signals acquired by this embedded platform. The designed smart sensors do not perform any type of analog filtering, except the PPG transducer. It was decided to implement IIR filters over FIR filters. IIR filters can have abrupt cutoff frequencies with less coefficients. This is highly useful for this work since real-time filtering is desired. Fewer coefficients means fewer multiplications, and fewer multiplications leads to less calculation time [19]. However, they can have nonlinear phase, which is not desirable. The transfer function for the IIR filter is given by equation 3.

$$H(z) = \frac{b_0 + b_1 z^{-1} + \ldots + b_N z^{-N}}{1 + a_1 z^{-1} + \ldots + a_M z^{-M}}$$  \hspace{1cm} (3)
The filter coefficients were calculated using the bilinear z-transform method [20]. To ease the design process the filter’s coefficients were calculated with the software MATLAB\textsuperscript{1}, using FDAtool\textsuperscript{2} from the Signal Processing Toolbox. A group of low pass filters with 1, 3, 5, 10 Hz was designed. The magnitude and phase frequency response of this filters are respectively illustrated in Figures 11 and 12.

![Figure 11: Designed digital filters magnitude frequency responses](image1)

To realize multiplications and sums of the coefficient fractional numbers and its input samples in the microcontroller, two’s complement fixed-point arithmetic was used instead of floating point arithmetic. The ADC was configured to return the samples values using  \( Q_{15} \) fixed-point notation. For second and fourth order filters  \( Q_{14} \) notation was used for the filters coefficients.

All the designed filters were implemented using direct form II (or canonical section) structures which present minimum number of delay lines for the  \( H(z) \) function. For high order filters, direct form realization of the filters is very sensitive to many adverse effects of finite wordlength such as quantization errors, so it is crucial to break the discrete-filter function into second order (or biquadratic) sections (equation 4 )\cite{19}.

\[
H_2(z) = \frac{b_0 + b_1 z^{-1} + b_2 z^{-2}}{1 + a_1 z^{-1} + a_2 z^{-2}} \tag{4}
\]

The second order filters were implemented with a single biquadratic section. This second order sections present a difference equation expressed by equations 6 \cite{20}, \cite{19}.

\[
w(n) = x(n) - a_1 w(n-1) - a_2 w(n-2) \tag{5}
\]

\[
y(n) = b_0 w(n) + b_1 w(n-1) + b_2 w(n-2) \tag{6}
\]

Finally, the fourth order filter uses a structure with two cascaded biquadratic sections . This can be seen as a multiplication of two biquadratic sections (equation 7 )\cite{20}, \cite{19}.

\[
H_4(z) = \prod_{k=1}^{N_s} \frac{b_{0k} + b_{1k} z^{-1} + b_{2k} z^{-2}}{1 + a_{1k} z^{-1} + a_{2k} z^{-2}}, N_s = 2 \tag{7}
\]

The difference equation resulting from the described cascaded structure is expressed in equations 8 and 9 \cite{20}, \cite{19}.

\[
w_1(n) = x(n) - a_{11} w_1(n-1) - a_{21} w_2(n-2) \tag{8}
\]

\[
y_1(n) = b_{01} w_1(n) + b_{11} w_1(n-1) + b_{21} w_1(n-2) \tag{9}
\]

\[
w_2(n) = y_1(n) - a_{12} w_2(n-1) - a_{22} w_2(n-2) \tag{8}
\]

\[
y_2(n) = b_{02} w_2(n) + b_{12} w_2(n-1) + b_{22} w_2(n-2) \tag{9}
\]

\[
H_4(z) = \prod_{k=1}^{N_s} \frac{b_{0k} + b_{1k} z^{-1} + b_{2k} z^{-2}}{1 + a_{1k} z^{-1} + a_{2k} z^{-2}}, N_s = 2 \tag{7}
\]

The developed algorithm to calculate the heart rate in real time from the PPG signal is based on the time difference between two consecutive signal peaks, To approach this in a simple way, a threshold is set in which all the values above that can be a potential maximum. The PPG the signal does not maintain a constant peak-to-peak amplitude during time, which means that the threshold value should be adapted during time.

The purposed solution pretends to estimate the average of the signal during a time interval, i.e., a time window (Figure 13). This windowed signal average will then be a part of a weighted sum with the previous threshold (equation ). By

1http://www.mathworks.com/products/matlab/
2http://www.mathworks.com/help/signal/ref/fdatool.html
doing this, a sense of memory can be provided to prevent the threshold to change dramatically between time windows, in case the signal present artifacts. A window of 1024 samples was used.

\[
threshold_i = 0.35 \times \text{mean}_i + 0.65 \times threshold_{i-1}
\] (10)

Having calculated the threshold value, the absolute maximum can be found. The peak is detected by comparing each acquired sample with the higher value stored until that moment. If the acquired value is bigger, then it is stored as potential maximum, and so one. It reaches a point where the signal drops below the threshold and the potential maximum until that time becomes the absolute maximum. Knowing the number of samples between peaks, it is possible to calculate the heart rate through equation (11).

\[
\text{heart rate (bpm)} = \frac{\text{num. of samples between 2 peaks}}{\text{sample rate}} \times 60
\] (11)

C. Template Description Language Parser

To correctly identify the content of the template TEDS, a TDL parser has to be available in order to decode the existing propriety and control command values. To do so, a parser was implemented. The parser workflow was developed focusing on the resources available in the microcontroller, because it was meant to be implemented in it.

In essence, the parser uses the template as a “map” to get the available commands data. It also has to correctly process the chained control commands such as SelectCase and Case, which can be compared as switch case statements of C language, to only retrieve the selected data from the TEDS. In general, the used templates make use of this control commands.

The beginning of the workflow of the TDL parser function is illustrated by the flowchart of Figure 14.

Briefly, it can be said that the template is read until the EndTemplate is processed. Each identified propriety belonging to a selected Case is stored into a microcontroller structure described by Figure.

D. Main System Workflow

To demonstrate where embedded system routines are applied, the main events will be displayed. Figure 15 shows the workflow of the beginning of the embedded application.

After the device is powered on, it immediately enters sleep mode where it waits for an external pin interruption, that will only be provided if an valid RFID tag is passed through the RFID reader. When the correct tag is read, the device enters idle mode where it waits for a peripheral interruption, whether from the UART1, UART2 or external pin, after it has evaluated what smart sensors are available in the sensors

figure 13: threshold calculation concept using mean window

figure 14: TDL parser flowchart first section

figure 15: main function flowchart
network. An interruption from UART1 means that there is received data available to be processed. UART2 interruption is flagged when there is RFID data available, which means that a tag has passed again through the RFID reader. An external button can also flag an interruption allowing the device to start acquiring smart sensors data and transmit it to a coordinator computer, if available. If the microcontroller cannot connect to a coordinator computer, it will save the data onto the SD card.

VI. RESULTS

A. Smart Wheelchair Sensor Measurements

Several tests related to the system functionality and reliability were carried out. Thus, after establishment of the 802.15.4 wireless network, acquisition tasks were performed during a time interval of 100s. The obtained results are presented in Figures 16-21. All the presented signals were sent to the coordinator computer through wireless at a rate of 100S/s, but were acquired by the 10-bit ADC at a sample rate of 1 kS/s. Likewise, appropriate digital filtering has been automatically applied to all the transmitted signals. Using the information available in the smart sensors, i.e., defined TEDS, both the embedded system and graphical user interface application can calculate the physical values of the demonstrated. This proves that the IEEE 1451.4 smart sensors can have an enormous advantage to physicians which simply have to connect the sensor without previous configurations to be able to see a physical representation of the signals.

B. Cardiac Activity assessment

Tests were performed on five male subjects to acquire PPG and BCG signals and evaluate the subjects heart rates. These tests had the duration of 5 minutes (300 seconds). Different methods are mentioned in literature referring heart rate estimation through BCG [21], [22], [23], [10]. However in the present work the objective was to acquire BCG waves
during daily activity taking into account the benefits of an unobtrusively measurement. In future work, advanced data processing of the acquired BCG is planed to be implemented on the computer coordinator side.

Figure 22 presents an example of acquired data. It shows 10 seconds from 300 seconds of acquired data at a rate of 1kS/s, transmitted via wireless at a 100S/s, and stored into the coordinator computer. Real-time heart rate calculation by the microcontroller was performed throughout the test and send to the coordinator computer. Heart rate estimation and important information about the subjects, such as body mass index, is summarized in Table I.

![Figure 22: PPG and BCG waves from Subject V](image)

Table I: Heart rate estimation of five test subjects

<table>
<thead>
<tr>
<th>Subject</th>
<th>HR mean (bpm)</th>
<th>Age</th>
<th>Height (m)</th>
<th>Weight (kg)</th>
<th>BMI (kg m(^{-2}))</th>
</tr>
</thead>
<tbody>
<tr>
<td>I</td>
<td>80</td>
<td>33</td>
<td>1.77</td>
<td>92</td>
<td>29.4</td>
</tr>
<tr>
<td>II</td>
<td>53</td>
<td>24</td>
<td>1.79</td>
<td>95</td>
<td>29.6</td>
</tr>
<tr>
<td>III</td>
<td>87</td>
<td>45</td>
<td>1.80</td>
<td>85</td>
<td>26.2</td>
</tr>
<tr>
<td>IV</td>
<td>88</td>
<td>25</td>
<td>1.89</td>
<td>65</td>
<td>18</td>
</tr>
<tr>
<td>V</td>
<td>77</td>
<td>24</td>
<td>1.92</td>
<td>70</td>
<td>19</td>
</tr>
</tbody>
</table>

Subjects with a BMI (Body Mass Index) lower than 18.5 are considered underweight; normal weight is considered between 18.5 and 24.9; overweight subjects go from 25 to 29.9; and obesity is considered over a BMI of 30. Knowing this we can state that subjects I, II and III have overweight, subject IV has underweight and subject V has normal weight. It is also known that subject II is a sportsman and has bradycardia, as it can be seen from its lower heart rate. The remaining subjects present a below average heart rate, near 80 bpm. However, this values should yet be compared in the future with commercial medical calibrated systems to validate its veracity.

C. Smart Wheelchair Wireless Communication Range and Current Consumption

Indoor range tests were realized to evaluate the performance of the embedded system wireless data transmission to the coordinator computer in closed space environments. Due to the number of prototypes limitations tests were realized with just one smart wheelchair. The tests were realized with the higher data transmission rate, i.e., by sending real-time data acquired from eight analog channels of the ADC. This test was conducted in indoor facilities (9th floor of Instituto Superior Técnico’s North Tower) also to evaluate the indoor wireless transmission when there exists also other type of communications using the 2.4GHz bandwidth of 802.15.4, such as WiFi and Bluetooth. The received signal strength in the locations represented by Figure 23 is presented in Table II. This proves that the used XBee Pro 802.15.4 modules enables the smart wheelchair to communicate in a indoor environment within acceptable distances.

![Figure 23: 9th building floor plant of Instituto Superior Técnico’s North Tower with transmission location points and receiver location.](image)

Table II: Signal strength of the received data and its approximate distance to the transmission location

<table>
<thead>
<tr>
<th>Location point</th>
<th>Received Signal Strength (dBm)</th>
<th>Coordinates (x,y)</th>
<th>Distance d from R to T&lt;\textgreater&gt; (m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>T1</td>
<td>-43</td>
<td>(18.75,7.5)</td>
<td>5.25</td>
</tr>
<tr>
<td>T2</td>
<td>-49</td>
<td>(21.75,14.25)</td>
<td>12.37</td>
</tr>
<tr>
<td>T3</td>
<td>-60</td>
<td>(21.75,20.25)</td>
<td>18.25</td>
</tr>
<tr>
<td>T4</td>
<td>-68</td>
<td>(14.25,21.75)</td>
<td>20.01</td>
</tr>
<tr>
<td>T5</td>
<td>-76</td>
<td>(8.25,21.75)</td>
<td>22.15</td>
</tr>
</tbody>
</table>

Current consumption measurements were made by connecting the power supply in series to a Keithley 2000 Digital Multimeter, connected to the computer through a GPIB-USB adapter cable. Using LabVIEW it was possible to acquire data from the digital multimeter with a 75ms interval.

Since the maximum total estimated current consumption value is 169.7 mA. In theory the used battery (11.1V, 1500mAh) should be enough to power the system to roughly 8 and a half hours in Transmission Mode. This value could be increased if a battery with large current capability is used.
A functional prototype solution for a smart wheelchair to monitor its user’s health was presented in this document. The solution is based on a platform essentially divided into three tiers. The sensing tier is accomplished a network of IEEE 1451.4 smart sensors, which represents a different approach from many health monitoring systems. The second tier of the platform consists on a data acquisition and processing system which makes use of this information to appropriately convert the smart sensor analog signal into a physical unit and to apply appropriate signal processing, such as digital filtering and a basic heart rate calculation algorithm. These first two tiers together create the concept of smart wheelchair. The third tier of the platform consists on the coordinator computer and a remote database. Smart sensors data was gathered in order to demonstrate the advantages of the solution. As it was concluded, the IEEE 1451.4 smart sensors proved to be a gain for this solution. However some limitations exist by using just IEEE 1451.4 Standard templates. It can be said that the templates should reflect better the proprieties of the biomedical sensors in future works.

Two performance tests were realized Indoor wireless range measurements were made able to verify that there was stable wireless communication to a distance between the smart wheelchair and the receptor of 22.15 meters. It was also demonstrated that the smart wheelchair consumes a maximum of 169.8 mA. The full system consumption could be optimized in future developments.

### REFERENCES