Recent Advances and Design Challenges in Wireless Health Monitoring

D. M. M.-P. Schreurs[1], M. Mercuri[1], P. J. Soh[1], Guy A. E. Vandenbosch[1]

Abstract – **Triggered by the increase of the elderly population, various aspects of health monitoring are being investigated by researchers worldwide. The earlier approaches in health monitoring were based on devices attached to the patient's body involving the use of wires that impose restrictions on patients and introducing discomfort, besides affecting device operation. Although some of these devices are still used nowadays, wireless technologies are being intensively investigated with the aim of providing an effective and efficient health monitoring solution, towards improving the well-being of the elderly. They represent the new emerging solution to promote the health both in home and clinical environments, and are also predicted to proliferate in the next years. This paper discusses the recent advances, design challenges, practical limitations, and solutions of recent advances in health monitoring using wireless technologies, presenting also an example of contactless health monitoring system.**

Keywords – **Fall detection, health monitoring, radar remote sensing, tag-less localization, wireless technologies.**

I. Introduction

The number of senior citizens, older than 60 years, is expected to grow to more than one billion by 2015, 2.5 billion by 2050, and 3.6 billion by 2100 [1]. This trend is causing a growing need for long-term health monitoring within the home environment. Shortage of nursery homes, increasing personal care costs, and privacy preference motivate the senior's wish to stay longer at home. However, this does not come without obvious risks, especially when living alone.

In fact, a lot of people suffer from chronic health conditions, including heart disease, lung disorders, diabetes, sleep disorders, and somnambulism. In particular, sleep disorders are recognized as detrimental to both physical and mental health, and affect the population on a large and growing scale [2]. Technologies available for diagnostics and treatment of chronic diseases typically involve recording of biomedical signals such as brain activity (EEG), heart pattern (ECG), muscle activity (EMG), eye movements (EOG), blood oxygenation saturation (SpO_2), and respiration effort.

[1]D. M. M.-P. Schreurs, M. Mercuri, P. J. Soh, and G. A. E. Vandenbosch are with the Div. ESAT-TELEMIC, KU Leuven, Kasteelpark Arenberg 10 Box 2444, 3001 Leuven, Belgium. E-mail: [dominique.schreurs, marco.mercuri, pingjack.soh, guy.vandenbosch]@esat.kuleuven.be

[2]P. J. Soh is also with the School of Computer and Communication Engineering (SCCE), Universiti Malaysia Perlis (UniMAP), Pauh Putra Campus, 02600 Arau, Perlis, Malaysia E-mail: pjsoh@unimap.edu.my.

These measurements are made through sensors attached to the patient's skin. However, these devices involve the use of wires that impose significant discomfort and restrictions on the patient, which can affect strongly the results.

In addition to chronic health problems, fall incidents are considered a major risk for elderly living independently, as falls often result in serious physical and psychological consequences, or even death [3]-[4]. Rapid detection of fall incidents can reduce the mortality rate and raise the chances to survive the event and return to independent living. For these reasons, fall detection has become of primary interest. Conventional health monitoring systems for this purpose are based on a wristwatch or necklace with a button that is activated by the person in case of emergency. However, persons in such situation may already be unconscious or no longer be reflexive to press the button.

Wireless technologies have been recently investigated to overcome some of these problems. They represent an emerging key aiming at health monitoring at home, of which the general public will benefit in terms of diagnosis, treatment, and detection of emergency situations, while providing reliable data communication through minimized interference with the patient's normal activities [5].

A distinction can be made between contactless and wearable approaches. In the first category, attention has been focused mainly on contactless vital signs monitoring [6] and contactless fall detection and tagless localization [7]. These academic works are based on radar techniques. More precisely, the Doppler shifts caused by the mechanical movements of heart and chest (lungs) can be detected and in this way vital signs can be measured. Examples of these academic developments are based on either CW (Continuous-Wave) Doppler radar [8] or UWB-IR (Ultra Wideband Impulse Radio) radar [9]. However, the capabilities of such devices are currently limited to heartbeat and respiration rate monitoring under "ideal conditions", e.g., constrained to a single motionless person (i.e., seated or lying down). Similar architectures are also investigated in rescue missions and behind/through-wall target detection which aim to search for living humans after an earthquake or building collapse [10]. These approaches try to detect the vital signs of a person. The problem that still limits their success is the difficulty in detecting the weak life signal from the strong echo waves of the background. Regarding contactless fall detection and tagless localization, a complete health monitoring system has been proposed and analyzed in [11]. The system, combining radar, wireless communications, and data processing techniques, is based on a hybrid approach by which a single tone is alternated with a stepped frequency waveform [12], and it has been tested on human volunteers in a realistic room setting with no constraints in their movements. Radar approaches are also used in applications of reconnaissance of objects buried under ground and surveillance [9].

Although a contactless system represents the ideal solution for long-term monitoring, avoiding therefore the disadvantage to wear the device, there are some biological parameters that are not movement related or that are too weak to be detected with the current contactless approaches. For these situations, a wireless wearable approach represents currently the most suitable solution. This second category, also referred to as body area networks (BAN), involves systems consisting of several miniaturized sensors, attached to the person's body, integrating one or several worn wireless transponders relaying continuously data to medical practitioners or caregivers. A typical BAN integrates mainly sensors for vital sign monitoring. However, physiological sensors, such as ECG and SpO_2 sensors, have also been developed. Other sensors such as a blood pressure sensor, EEG sensor, and smartphone interface are under development. For example, a wearable healthcare monitoring system that integrates an ECG device and an accelerometer with a mobile device in a Bluetooth-based network has been presented in [13]. The feasibility of using a wearable UWB radar pulse sensor integrating a Bluetooth module has been demonstrated in [14] for arterial stiffness measurement. A device combining wireless feature and a pulse oximeter to capture patient's heart beat and blood oxygen saturation by measuring the amount of light transmitted through a non-invasive sensor attached to the patient's finger is shown in [15]. The use of a tri-axial accelerometer placed in a mobile phone to detect falls has been proposed in [16].

This paper presents design challenges, practical limitations, and their solutions for wireless health monitoring systems based both on contactless approach and on WBAN. An example of contactless health monitoring system is also reported.

In Section II, system design challenges and solutions for a contactless health monitoring system and WBAN nodes are presented. Antenna design challenges and proposed solutions are discussed in Section III. Experimental results on human volunteers are then presented in Section IV.

II. SYSTEM DESIGN CHALLENGES

In order to discuss system-level challenges and possible solutions, two systems are considered: a radar sensor based health monitoring system aiming at contactless fall detection, tagless localization and vital signs monitoring for in-door applications on one hand, and a BAN node for telemetry applications on the other hand.

A. Radar Sensor Node

Remote health monitoring can be made contactless and therefore non-invasive by adopting radar techniques. The radar is used to transmit an RF waveform to a target and to receive the reflected echo, on the basis of which the target's speed, vital signs, and absolute distance can be extracted.

Due to its narrowband nature, a pure CW radar presents a simple hardware architecture. However, although it can easily measure the target's velocity and vital signals' parameters without ambiguity by exploiting the Doppler effect, it is not able to determine absolute distances and to separate reflections temporally. On the other hand, an UWB-IR radar is able to detect both distance and speed. However, its main problem lies in the hardware requirements which are tougher as the bandwidth needs to be much larger. Moreover, it cannot detect speed by exploiting the Doppler effect. It means that, for a good accuracy in speed, it requires a high accuracy in target position since speed information is obtained dividing space displacement by its time interval. The better the distance resolution, the wider the bandwidth will be. A wider bandwidth increases both complexity and power consumption of the radar transceiver, involving high-speed analog-to-digital-converters (ADCs) and high level processors.

On the contrary, the narrow instantaneous bandwidth combined with the large effective bandwidth (sequentially over many pulses) of a stepped frequency continuous wave (SFCW) radar implies that the hardware requirements become less stringent. Lower-speed ADCs and lower level processors can be used. Moreover, the receiver bandwidth is smaller, resulting in lower noise bandwidth and higher signal-to-noise ratio, increasing the radar sensitivity. In addition, it can exploit the Doppler effect like in a conventional CW radar. For that reason, an SFCW radar represents the most suitable architecture for wireless health monitoring, aiming in particular at contactless fall detection, tagless localization, and vital signs monitoring.

In this section, a radar sensor based on an SFCW architecture is presented. It alternates a CW signal with an SFCW waveform to monitor persons and, at the same time, to satisfy the European and Federal Communications Commission (FCC) UWB spectrum masks. Moreover, it presents wireless communication features.

Fig. 1 shows a simplified block diagram of a radar-based health monitoring system [11]. It consists of a sensor, combining both radar and wireless communication features, and a base station for data processing. A radar waveform *T(t)* is continuously generated and sent to the target. The reflected echo *R(t)*, containing speed and absolute distance information, is collected by the receiver. The resulting *I(t)* and *Q(t)*

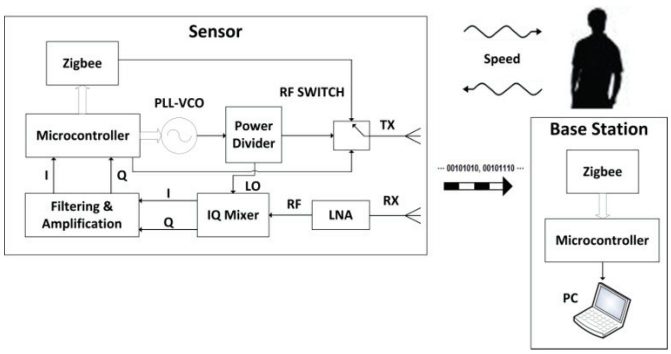


Fig. 1. Simplified block diagram of the contactless health monitoring system

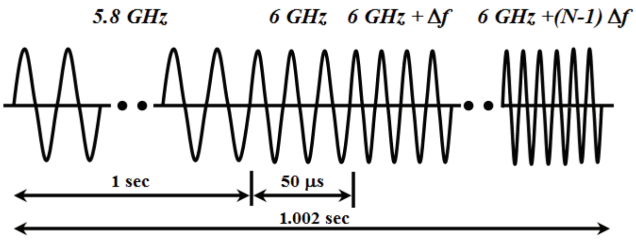


Fig. 2. Designed radar waveform

baseband signals are digitized and transmitted wirelessly to the base station for data processing, in order to distinguish a fall event from normal movements (e.g., walking, sitting and lying down, no movements, standing up), and to determine the target's absolute distance and vital signs' rates.

The radar sensor node consists of three main parts, namely the radar module, the Zigbee module, and the microcontroller. It integrates a wideband voltage controlled oscillator (VCO) with a phase locked loop (PLL), a power divider, an RF switch, a low noise amplifier (LNA), a gain block, an In-phase and Quadrature (IQ) mixer, and baseband filters and amplifiers. It transmits a single tone at f_{ISM} = 5.8 GHz in the ISM band, alternated with an SFCW waveform working in the UWB band between 6 and 7 GHz, as shown in Fig. 2. Each tone lasts 1 s and is used to continuously detect the speed or the vital signs of a person, while the SFCW waveform is used to determine the target's absolute distance.

One of the main challenges is represented by the microcontroller. It in fact programs the PLL to generate the radar waveform, it acquires and digitizes the I/Q baseband signals, and manages the Zigbee communication. Moreover, it controls the RF switch to connect alternately the radar transmitter and the Zigbee module to the transmitter antenna. The latter is used by the Zigbee module also to receive frames. The microcontroller should represent a small area to have compact solution and it should operate with a sufficient clock rate to achieve its tasks but at the same tame to limit power consumption.

A SFCW waveform consists of N coherent CW pulses (called *burst*) whose frequencies are increased from pulse to pulse by a fixed increment Δf (Fig. 2). One of its main advantages is that for each pulse the relative I/Q baseband signals consist of direct current (DC) levels, requiring thus the acquisition of one single I/Q pair per pulse width. This makes the sample rate lower compared to other ultra wideband radars. However, the main challenge is represented by the SFCW waveform design. In fact, in order to detect the target's absolute distance, the burst interval ($N \cdot T$) should be sufficiently short such that the target may be assumed static during a burst. If that condition is not satisfied, the target will be positioned to a wrong distance and its corresponding peak, indicating its absolute distance, will result in spread in the range profile. Moreover, the burst should be also sufficiently short to increase the transmit power level, in order to cover the size of a typical room environment, satisfying at the same time the spectral masks. In addition, since each burst is generated in between single tones, its duration should be shorter than the sample rate used to acquire the speed signals, avoiding thus loss of information. Taking into account all these considerations, an adequate burst interval should be around 2 ms. In order to cover a 5 x 5 m^2 room with a resolution of 15 cm, the SFCW waveform should have N = 40 CW pulses and a fixed increment Δf = 25 MHz [11]. This means that each CW pulse is T = 50 µs long. Therefore, the challenge is to perform several operations in this short period. In fact, during this interval, the sensor should generate a new CW pulse, taking into account both the time to program the synthesizer of the PLL and the VCO's settling time, acquire a pair of I/Q baseband samples, taking into account the time required to sample and hold the signal, and to send the latter to the Zigbee module, considering the time required to handle this operation. It should be also considered that the shorter is the VCO's settling time, the wider will be the PLL loop filter increasing the phase noise of the system. The latter is an important parameter to be controlled especially for vital signs detection.

The baseband I/Q signals are filtered, amplified, and positioned to the centre of the ADC's dynamic range. The samples are mapped in frames of bytes and are then sent by Zigbee to the base station for data processing. The challenge here is to manage properly the synchronization of the system and the wireless transmission. In fact, it is of vital importance that the Zigbee transmission must be executed in between speed sampling instants to avoid loss of information.

System-on-chip integration of wideband Doppler-based radars allows having an energy-efficient, inexpensive, and compact solution. However, the main challenge is represented by the wideband VCO with PLL. In fact, it should present low power consumption as well as be very linear and have a wide frequency tuning range. Moreover, the output power should be about 0 dBm and it should be fast enough to generate the desired waveform while presenting also a low phase noise.

B. Wearable Sensor Node

With regard to wearable sensor nodes, we focus in this section on the wireless communication aspect on the physical (PHY) layer, and not on the sensing part as this is strongly dependent on the bio-parameter(s) to be monitored. Examples on the latter have been introduced in the Introduction section.

The transmission of medical information in an on-body health monitoring system basically needs to be performed for two different purposes: 1) for communicating the collected physiological signals from the biosensors to an on-body central node (BCU); and 2) for sending measurements from the wearable system to a remote medical station, thus communicating in the extra-BAN (EBAN) mode [17]. The user's mobility and comfortableness can be hindered by the use of wires, besides the increased risk of system failure [18]. The use of conformal antennas to transmit such measurements collected from clothing integrated sensors is certainly a more favourable approach [19].

In general, multiple autonomous wearable sensor nodes can form a body area network (BAN). Challenges with the use of this technology include interoperability, system complexity, security and privacy, power efficiency and data consistency due to interference. For communication in the EBAN domain, i.e., scenario (2) described above, BCUs must be capable of seamless data transfer across various wireless standards, e.g., Bluetooth, Zigbee, WLAN and preferably be compliant with mobile phone communication standards as well [20]. This will then ensure efficient migration across networks and offer uninterrupted connectivity to a mobile monitoring station. In 802.15.6 alone, three PHY layers have been defined: narrowband (NB), ultra wideband (UWB) and human body channel communications (HBC). Complicated hardware is foreseen when a generic transceiver front-end is needed to combine this wireless functionality with the sensing part into a worn integrated hardware module with a small footprint. Several hardware architectures have been proposed for the narrowband BAN PHY in [21].

The most commonly employed wireless communication standards in Wireless Personal Area Networks (WPANs) are the IEEE 802.15.1 (Bluetooth) and IEEE 802.15.4 (Zigbee) prior to the release of the IEEE 802.15.6 (WBAN) standard. Both originated from the IEEE 802.15 Working Group. Zigbee is especially interesting, as it aims for low-cost, low data-rate solutions with long battery life and very low circuit complexity. It operates in the 2.4 GHz industrial, scientific, and medical (ISM) band (16 channels), in the 915 MHz band (10 channels) and in the 868 MHz band (one channel) [22]. In contrast to 802.15.4, the new 802.15.6 WBAN standard caters for the shorter communication range on body, with extremely low power consumption in standby mode and enough power in its active mode. Moreover the larger data rate in WBAN enables the support of various non-medical applications such as entertainment and disability assistance [23, 24]. A comparison of both systems is summarized in Table I.

The main challenge of health-monitoring is the associated resource efficiency, both in terms of power and cost, besides hardware complexity. The number of functionalities in a node and the resulting power consumption may need to be balanced. A summary of typical medical applications in WBANs and their communication and power consumption requirements are summarized in Table II [25].

TABLE I
COMPARISON BETWEEN WBAN (IEEE 802.15.6) AND ZIGBEE (IEEE 802.15.4) [23]

	IEEE 802.15.6	IEEE 802.15.4
Supported applications	Medical, entertainment, sports, etc.	Home, industrial automation, etc.
Range	2-5 m	10-100 m
Data rate	kbps to 10 Mbps	20, 40, 250 kbps
Power consumption	0.01mW (standby) 40mW (active)	25-35 mW
Network size	Max 256 devices per BAN	Up to 65K devices
Safety	Meet regulation requirements for SAR and HPPA	None

TABLE II
TECHNICAL REQUIREMENTS FOR SELECTED BAN APPLICATIONS [25]

Application	Data rate	Bit error rate (BER)	Desired battery life
Deep brain stimulation	1 Mbps	$< 10^{-3}$	> 3 yrs
Hearing aid	200 kbps	$< 10^{-10}$	> 40 hrs
Capsule endoscopy	1 Mbps	$< 10^{-10}$	> 24 hrs
Drug dosage	< 1 kbps	$< 10^{-10}$	> 24 hrs
ECG/EEG/EMG	< 1.6 Mbps	$< 10^{-10}$	> 1 week
O2/CO2/BP/ Temperature/ Glucose monitor/ accelerometer	< 10 kbps	$< 10^{-10}$	> 1 week
Audio	1 Mbps	$< 10^{-5}$	> 24 hrs
Video/Medical Imaging	< 10 Mbps	$< 10^{-3}$	> 12 hrs

Besides these requirements, the node's hardware complexity also depends on the WBAN architecture selection - the operation of the biosensor and the BCU. Several WBAN architectures proposed include Managed Body Sensor Network (MBSN), Autonomous BSN (ABSN), and Intelligent BSN (IBSN). The first architecture, MBSN, is the most intuitive type of health monitoring mechanism, involving continuous real-time information transmission between a patient and a base station. However, in non-urgent monitoring applications, a periodic transmission can be adopted instead of continuous, enabling reduced power consumption. Another option would be to periodically define a transmission interval and to add a trigger transmission for emergencies to enhance power-efficiency. The latter requires a portion of the received medical information to be temporarily stored locally in the BCU, which consequently adds to hardware complexity. More advanced biosensors in BCU are sometimes equipped with an emergency response mechanism. For example, an intelligent glucose biosensor is able to inject insulin into a patient's body when necessary, upon instruction by the medical practitioner. However, this requires a secure and reliable EBAN network, i.e., WLAN, GSM, etc, and all decisions on follow-up actions to be taken by the biosensor when an emergency occurs need to be triggered by the medical practitioner. A less complex system can be certainly enabled if the EBAN wireless communication hardware can be reduced. The same glucose biosensor can be made to measure, interpret its own diagnosis and decide to inject insulin to the patient, forming an ABSN with several other such biosensors. Such autonomous network will allow the patient to be more independent, besides allowing cost reduction in terms of medical personnel's intervention. In short, hardware can be simpler and power consumption can be lowered due to the reduction of long range communication needs. An IBSN, on the other hand, is a combination of both MBSN and ABSN mechanisms - biosensors are programmed to take their own corrective decisions for straightforward cases, whereas a medical practitioner's diagnosis and instruction is required for more complicated situations. Large scale BAN deployment, e.g., multiple nodes across a common worn vest for a patient should also incorporate intelligent interference reduction. Besides, this feature is also useful for operation in densely occupied environments. Examples of such mechanisms include the use of beamforming, direction of arrival (DoA) estimation, multiple-input-multiple-output (MIMO) system, etc. [26], [27].

To summarize, there should be no “one-size-fits-all” concept in determining the ideal WBAN hardware. These nodes need to be optimized according to their purposes to be appealing and practical. For example, a patient with high risk of cardiac abnormalities should be constantly monitored using less complicated hardware to enable the BAN node to be energy-efficient, whereas patients with high falling risks should be monitored only when trying to move from any stationary position. However, a common hardware "core" is required, i.e., a BCU (wireless module and biosensor) and the conformal antenna, which must be “designer-friendly” as to allow maximum flexibility and scalability to suit specific applications. Ultimately, the hardware must be wearable, lightweight, guaranteeing comfort and not susceptible to human-errors (e.g., forgetfulness, improper BCU attachment) to enable continuous monitoring without the user realizing it.

III. Antenna Design Challenges

The integrated home-monitoring system consist of two important sub-systems which require wireless communications to channel back these sensed informations to a base station or amongst WBAN nodes, each sub-system needs a specific set of antennas. The requirements and challenges in designing such distinct set of antennas are listed in the following sections.

A. Radar Antennas

An indoor radar system requires a set of compact, high-gain and unidirectional radiators to effectively determine target location, vital signs, and daily activities (sitting, walking, standing, etc.). Higher gain enables optimal transmit power combined with guaranteed reception of target-reflected signals containing location, vital signs, and movement information [28]. The simultaneous unidirectional and wideband requirements, which have risen due to the use of the SFCW radar, described in section II.A, are challenging for antenna designers, as a full ground plane limits antenna bandwidth. Secondly, to avoid a more complex and more lossy RF design (e.g., circulator, divider) resulting from a single transmit-receive radar antenna, the combination of two elements for the same purpose translates into a larger radiator footprint. To maintain compactness, both elements are required to be closely-spaced, which gives rise to high mutual-coupling. One such proposed antenna is presented in Fig. 3 [29].

In the case where ultra-wideband (UWB) radar operation is required, it is important to note that the allowable mean Equivalent Isotropically Radiated Power (EIRP) is limited to -41.3 dBm/MHz (between 6 and 8.5 GHz). This is defined in the European Telecommunications Standards Institute (ETSI) and Federal Communications Commission (FCC) UWB spectral masks, which makes the unidirectional radiation requirement more challenging. However, the forward transmit power must still be kept as high as possible to ensure sufficient received power after reflection and absorption by the subject. Besides reducing power efficiency, the existence of back-radiation will also limit the radar's detectable range due to reduced directivity. To avoid this, a full ground plane or a cavity backing could be utilized to effectively suppress back radiation, which then results in a thicker radiator. Furthermore, the miniaturized footprint objective results in a small spacing between transmit and receive antennas, giving rise to mutual coupling. This can be solved either using a vertical wall, utilizing miniature electromagnetic bandgap (EBG) elements or simple microstrip sections [29]-[32].

B. On-body Antennas

Implementation of wearable on-body antennas is not simple as their performance is prone to degradation due to the power absorption by and dielectric coupling to the human body [33], [34]. Due to the general planar antenna requirement that the antenna radiators need to be conductive, various researchers have proposed various material types for wearable antennas' development. They include an array of flexible or rigid planar materials which are thin enough to be wearable, typically applied as their substrates. The type of substrate used will then determine its flexibility and suitability to be worn by a user. Techniques to realize the antenna topologies include conductive painting/printing, utilizing conductive strands to form an array of wires/meshes, coating non-conductive textiles, etc. For the more popular planar configuration, these conductors are generally implemented on a layer of substrate (flexible or thin and rigid) such as textiles, polymers, or dielectrics [35].

To implement properly operating antennas, designers must first understand the properties of utilized materials and their manufacturing processes. Conductive textiles can be easily cut and formed using manual tools such as scissors, whereas the printing technique requires dedicated printers and a more customized process due to the use of specialized conductive ink/liquid. Other processes such as sewing and embroidering also require specialized sewing machines [36]. The conductivity of the resulting conductive layer used to form the radiator will certainly depend on the adapted materials and processes. Moreover, the ability to manufacture prototypes with more complex geometries and tight accuracy requirements also depend on this process factor.

The design procedure for such antennas slightly differs

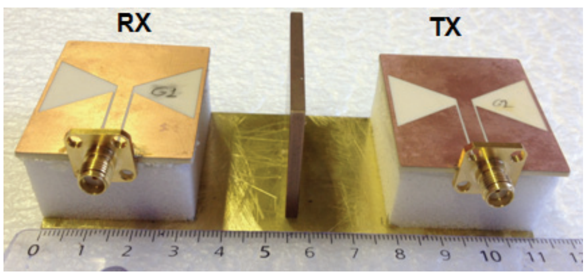
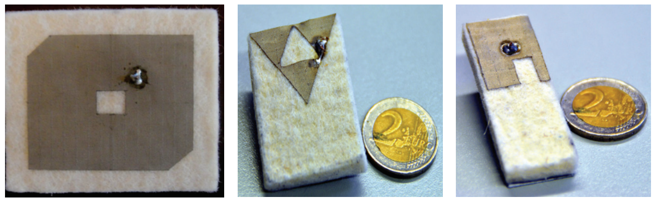


Fig. 3. Prototype of bow-tie TX/RX antenna array for broadband indoor radar applications

compared to their conventional counterparts. Their calculation and performance analyses can be performed using the various types of electromagnetic solvers, but the choice of their input material parameters must be made with care by the designers. For commercially available conductive textiles, its conductivity is typically specified in Ohms per square unit [37], and this can be optionally converted for use in solvers as Siemens per meter. The conductivity of the conductive textiles might not affect the reflection performance drastically, provided i larger than 1000 S/m [38]-[39], The substrate properties, on the other hand, are crucial in determining the correct reflection and radiation performance of a flexible radiator operating in planar condition. An ideal completion to such software-assisted analysis would be to co-simulate the final antenna in proximity of a human body model. These models range from approximate models such as one proposed in [40] to the very detailed Hugo model. Designers can also opt for specialized software able to simulate human movements (e.g., sitting, walking, running, etc.) [41].

Design parameters such as reflection coefficient, input impedance, radiation pattern, efficiency, gain, and fidelity behaviour are gathered and analysed similarly as any conventional antenna. Nonetheless, it would be more preferable that access to a three-dimensional EM solver is enabled, considering that further study on the effect of the antenna's physical changes (e.g., bending, crumpling, etc.) needs to be performed at this stage as final validation. Several examples of antenna validations have recently been reported in [42]-[44]. However, depending on the implemented materials, the prototyping of such antennas are more challenging compared to conventional antennas. Firstly, the lower accuracy expected from the use of manual tools, in the case of dimensioning commercial conductive textile, is expected to impose some extent of limitations on the choice of the antenna geometry. The same goes for printing, i.e., a costlier printer is required to enable a more accurate antenna dimensioning during prototyping. It is worth to mention that differences related to fabrication accuracy and costs will be more or less diminished once mass production technologies are applied for both textiles and flexible printed structures. Next, the overall topology should be also manufacturable using a continuous piece of conductive cloth/imprint [45] to avoid the need for galvanic connections further down the design chain. This will then avoid the need for re-connections of separated conducting elements using similarly conductive components (epoxy, glue, paint or ink, etc.).

In [26], [44]-[45], several fully textile antennas have been proposed and evaluated. They can be categorized into three types - first, a narrowband microstrip antenna (NBA), second, a broadband planar inverted-F antenna (BPIFA), and third, a dual band PIFA (DBPIFA). Their topologies are shown in Fig. 4. Their on-body performance has been evaluated in [33], [38], [39]. The antennas feature a full ground plane which enables effective shielding against absorption by the human body. The antenna prototyping materials consist of two textile types, conducting and non-conducting. The former is used to form the radiator, ground plane and shorting wall, whereas a non-conductive felt is used as the substrate. For the conductive materials, a commercial nickel and copper coated



(a) (b) (c)

Fig. 4. Fabricated prototypes of the all-textile antennas: (a) narrowband antenna [33], (b) broadband PIFA [38], and (c) dual band PIFA [39]

polyester fabric, known as ShieldIt Super, is used. It features an adhesive reverse side to easily secure it onto the felt substrate, which can be activated by ironing. The 50 Ω SMA connector is used to feed power into the radiators, and was connected to the textile by soldering, as it is able to withstand a 200°C maximum temperature. More details on the fabrication procedure can be found in [45].

IV. Experimental Results

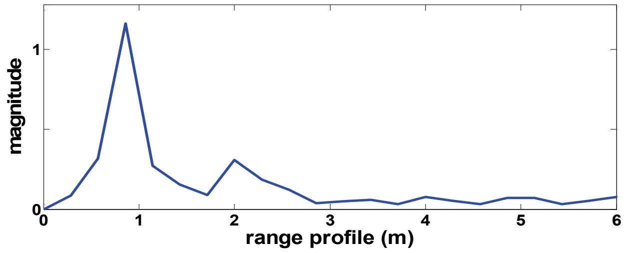
In this section, in the first two parts, experimental results regarding the health monitoring system are presented. Experimental tests have been conducted with real human volunteers who were allowed to move freely in a real room environment. Furniture and metallic shelves were deliberately positioned to enable the existence of clutter and reflections, mimicking a typical room setting. The sensor has been fixed to the wall at 1.5 m of height. Finally, the third part presents the on-body evaluation of the textile antennas and their Specific Absorption Rate (SAR).

A. Tagless Localization

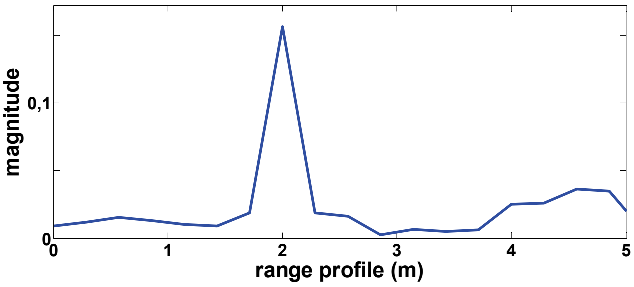
The target's range profile is determined applying the Inverse Fast Fourier Transformer (IFFT) to the complex I/Q SFCW samples. The main problem of this operation is to separate the weak target's reflection from the much stronger reflections of backscattering, cross-coupling between the two antennas, and cluttering. These unwanted effects involve strong reflections that overwhelm the much weaker target's reflected signal. However, their contributes can be eliminated by a compensation that consists in determining the range profile without any person in the room that characterizes the total contribution from backscattering, cross-coupling, and cluttering. The resulting magnitude is then subtracted from the range profiles determined with the target in the room. After that, the range profile should be shifted by a fixed factor to compensate the effect of the phase shift at the target surface together with the additional phase difference between the mixer and the antenna. This value is determined through calibration using a flat metal plate, placing it at a known distance from the antennas to evaluate its range profile.

Fig. 5a represents the range profile of a person in the room at 2 m in front of the antennas, before eliminating the undesired reflections, originating from the clutter, antenna's cross-coupling, and backscattering, and prior the calibration process. The target's peak, which is supposed to indicate its related absolute distance, is entirely overwhelmed by the unwanted reflections. After the compensation and the calibration steps, the target's peak can be perfectly distinguished as shown in Fig. 5b.

However, it has to be noted that this detection can be successful only if backscattering and crosstalk are strongly reduced by appropriate antenna design.



(a)



(b)

Fig. 5. Range profile of a target at 2 m away the antennas before (a) and after (b) compensation and the calibration

B. Fall Detection

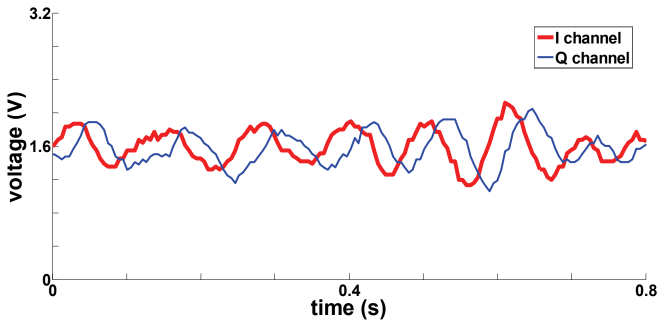
The speed samples are processed using the Least Square Support Vector Machine (LS-SVM) with Global Alignment (GA) kernel machine learning approach to discriminate fall event from normal movement [46]. It aims at distinguishing different changes in speed experienced during a fall and a normal movement. The technique consists of two phases, namely training and testing. Both phases use the digitalized speed signals as input.

In the first phase, a training set containing movement activities (i.e., walking and falling) is used to estimate an activities classification model that is validated in the testing phase. Before learning a model, the raw radar data has to be preprocessed. Each activity is grouped in a segment of 2 seconds, considered sufficient to cover the details of the activities and mainly the fall event. Given such segments, the data is standardized such that each dimension has zero mean and unit standard deviation. Then is transformed using the Short Time Fast Fourier Transform (STFT) from which only the magnitude spectrum is retained. Prior to the learning phase, the data is standardized again with the previous procedure. Once the learning process is finalized, the model is created and available to be validated in the testing phase. In doing that, an independent test set, with data not used to learn the classification model, is needed.

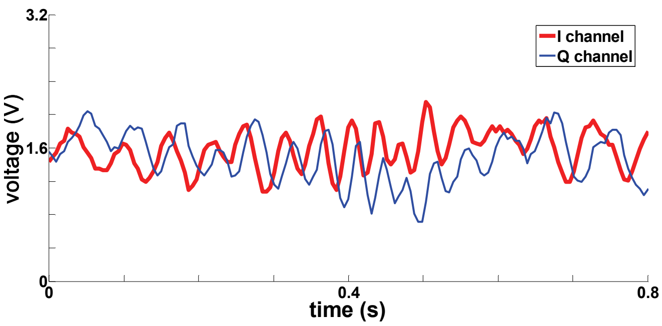
The LS-SVM with GA kernel technique has been tested building a data set containing 80 activities measured from two persons. It includes 20 walking signals acquired for each person, who was allowed free movement in the whole room, and 40 frontal fall signals, acquired with each subject located at known distances from the antennas.

Figs. 6a and 6b show the speed signals corresponding to walking and a fall, respectively. The results clearly show the difference between the two types of movement. During a fall, the speed continuously increases until the sudden moment when the fall is finished. During walking, the speed is quite constant over time.

The LS-SVM model is trained using the data of a single person (target 1) and then validated using the data from the other person (target 2). This process is repeated two times since data from two persons was available. The classification results, shown in Table III, have indicated a success rate in distinguishing fall events from normal movements of 95%, without reporting false alarms.



(a)



(b)

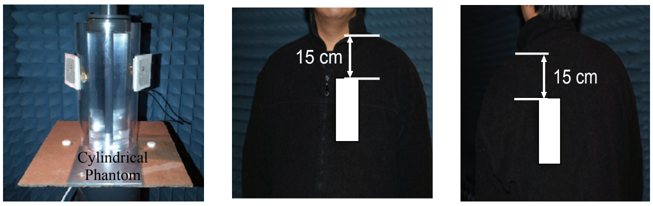
Fig. 6. Speed signal during (a) a walking movement and during (b) a fall event. The frequency of the signal is proportional to the radial velocity of the person during the movement. An inflating mattress has been used when invoking falls. For that reason, the fall speed signals do not stop suddenly but there is also the effect of the rebounds on the mattress

TABLE III
CLASSIFICATION RESULTS USING THE LS-SVM WITH GA KERNEL

Target 1		Target 2	
% False Positives	% Success Rate	% False Positives	% Success Rate
0	95	0	95

C. On-body Evaluation of Textile Antennas

As mentioned in section III.B, all antennas under consideration (Fig. 4) feature a ground plane to act as a shielding mechanism against on-body performance deterioration. They operate at the 2.45 GHz ISM band (for NBA and BPIFA), and at both 2.45 GHz and 5.2 GHz (for DBPIFA). How small the effect might be, it is still very useful to quantify and understand their behaviour using a realistic phantom and a real human volunteer. The phantom consists of two cylinders, with its inner cylinder filled with a commercial body-emulating liquid. This is shown in Fig. 7(a), and the NBA was evaluated using this method. Meanwhile, the next evaluation was a real human volunteer, of 1.78 m height and 90 kg. Evaluations are carried out on two potential locations on the human upper torso, i.e., on the chest and back, where vital bio-signals are most likely to be collected from.



(a) (b) (c)

Fig. 7. On-body evaluation setups: (a) on-body phantom, and (b) on a real human volunteer (chest), and (c) back

Antennas under test (AUTs) are evaluated both vertically and horizontally on each location to determine changes due to their orientation (denoted as CH and CV for chest, and BH and BV for back). A 10 mm felt spacer is attached to the ground plane of both AUTs to emulate a realistic spacing where the health monitoring system is to be placed, as shown in Fig. 7(b)-(c). Both BPIFA and DBPIFA were evaluated using this setup.

The antennas' reflection coefficients (S_{11}) are measured using a Vector Network Analyzer, while their radiation performance is assessed in an anechoic chamber. The frequency range between the S_{11} borders equal to -10 dB is taken as their bandwidths (*BW*), while the centre frequency (f_c) is located at their respective *BW* centre. All measurements were performed in free space (FS) prior to on-body (OB) evaluations. The S_{11} changes for all antennas are summarized in Table IV, and Figs. 8 to 10. As observed, the *BW* remained unchanged for the narrowband antenna when placed on the phantom. However, the effect of on-body placement of this antenna is more affecting the radiation efficiency and gain rather than its reflection performance. On the other hand, the *BW* is generally degraded for both PIFAs when placed on the chest, regardless of antenna orientation. For back placement, however, a different behavior is observed. The *BW* on BV is slightly increased, especially in the upper bands for both BPIFA and DBPIFA. This indicates a strong influence on the antenna's Q, causing a BPIFA *BW* expansion of up to 220 MHz relative to free space. The centre frequency f_c is generally moving down the frequency scale when placed anywhere on the body. Nonetheless, f_c for BH and BV orientations are on average less affected by on-body placement.

TABLE IV
EVALUATION SUMMARY FOR THREE TEXTILE ANTENNAS [26], [33]

Ant. Type/OB location	FS	CH	CV	BH	BV
NBA BW (GHz)	0.28	0.29			
NBA fc (GHz)	2.46	2.37			
NBA η_{rad} (%)	42.2	38.1			
BPIFA Gain (dB)	4.80	3.40			
BPIFA BW (GHz)	1.28	1.16	0.99	1.23	1.50
BPIFA f_c (GHz)	2.62	2.53	2.47	2.60	2.62
BPIFA η_{rad} (%)	80.9	32.9	38.9	59.6	56.2
BPIFA Gain (dB)	1.53	NA	NA	NA	NA
DBPIFA BW(L) (GHz)	0.33	0.28	0.35	0.29	0.30
DBPIFA BW(U) (GHz)	0.56	0.54	0.54	0.52	0.57
DBPIFA fc(L) (GHz)	2.14	2.09	2.14	2.09	2.09
DBPIFA fc(U) (GHz)	5.28	5.21	5.12	5.20	5.15
DBPIFA η_{rad}(L) (%)	80.3	37.3	NA	56.6	NA
DBPIFA η_{rad}(U) (%)	79.3	43.8	NA	52.9	NA
DBPIFA Gain (dB)	1.80	NA	NA	NA	NA

The final evaluation of these antennas is to determine whether these antennas are safe for use on a human body. This safety level is defined by the SAR, which is the level of absorbed electromagnetic energy by the human body.

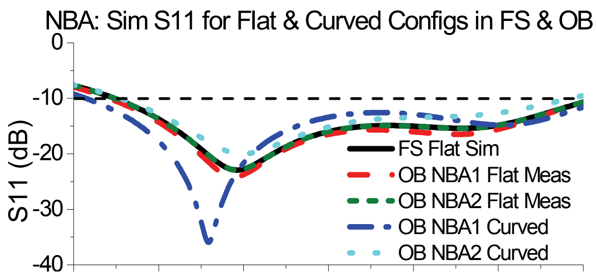


Fig. 8. Measured and simulated S_{11} and axial ratios for NBA in free space and on phantom [26]

As specified in the IEC 62209 standard [47], its limit is 2 W/kg averaged over 10 g of tissue. CST simulations using a realistic Hugo body model were performed to determine the SAR for each antenna. The detailed simulation procedure is described in [48]. Their results summarized in Table V are well below the 2 W/kg safety limit, indicating that these antennas with full ground plane are safe for further circuit integration and deployment on human users.

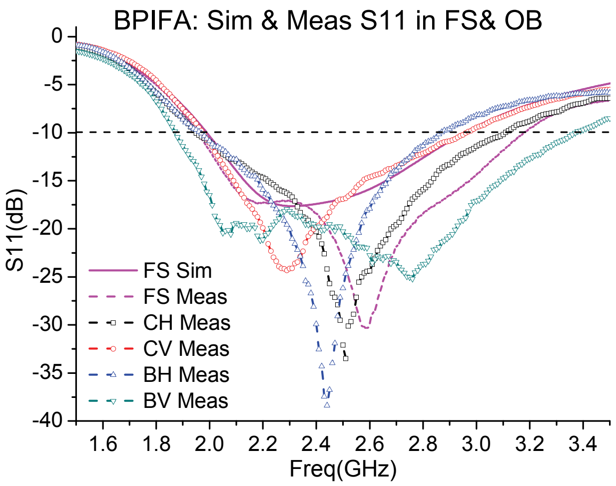


Fig. 9. Measured and simulated S_{11} for BPIFA in free space and on body locations [45]

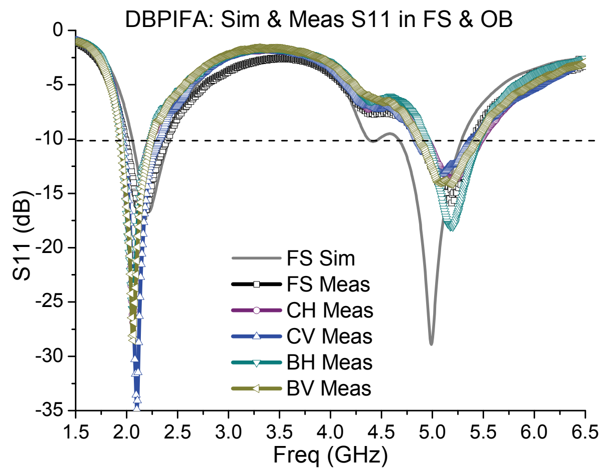


Fig. 10. Measured and simulated S_{11} for DBPIFA in free space and on body locations [38]

TABLE V
EVALUATED SAR FOR TEXTILE ANTENNAS [26], [48]

SAR/freq.	NBA	BPIFA	DBPIFA
2.45 GHz (W/kg)	0.014	0.492	0.276
5.20 GHz (W/kg)	NA	NA	0.583

V. CONCLUSION

Contactless and wearable approaches have been investigated for long-term patient health monitoring in the home environment. Although the feasibility of the basic system principles have been demonstrated, it is also important to solve several challenges that realistic environments and situations impose. This paper presented the recent advances, design challenges, practical limitations and potential solutions for wireless health monitoring systems, which combine radar based remote monitoring and wearable BAN. Experimental evaluations with real human subjects have also been reported.

ACKNOWLEDGEMENT

This is an extended version of the paper "Wireless Health Monitoring: Design Challenges" presented at the 11th International Conference on Telecommunications in Modern Satellite, Cable and Broadcasting Services - TELSIKS 2013, held in October 2013 in Niš, Serbia. This work was supported by FWO-Flanders, KU Leuven GOA, Hercules, and IOF projects, and the Malaysian Ministry of Education (MoE) - Higher Education Sector.

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