SPWID 2016

The Second International Conference on Smart Portable, Wearable, Implantable and Disability-oriented Devices and Systems

ISBN: 978-1-61208-480-0

May 22 - 26, 2016

Valencia, Spain

SPWID 2016 Editors

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SPWID 2016

Foreword

The Second International Conference on Smart Portable, Wearable, Implantable and Disability-oriented Devices and Systems (SPWID 2016), held between May 22-26, 2016, in Valencia, Spain, is an inaugural event bridging the concepts and the communities dealing with specialized implantable, wearable, near-body or mobile devices, including artificial organs, body-driven technologies, and assistive services

Mobile communications played by the proliferation of smartphones and practical aspects of designing such systems and developing specific applications raise particular challenges for a successful acceptance and deployment.

We take here the opportunity to warmly thank all the members of the SPWID 2016 Technical Program Committee, as well as the numerous reviewers. The creation of such a broad and high quality conference program would not have been possible without their involvement. We also kindly thank all the authors who dedicated much of their time and efforts to contribute to SPWID 2016. We truly believe that, thanks to all these efforts, the final conference program consisted of top quality contributions.

Also, this event could not have been a reality without the support of many individuals, organizations, and sponsors. We are grateful to the members of the SPWID 2016 organizing committee for their help in handling the logistics and for their work to make this professional meeting a success.

We hope that SPWID 2016 was a successful international forum for the exchange of ideas and results between academia and industry and for the promotion of progress in the areas of smart portable devices and systems.

We are convinced that the participants found the event useful and communications very open. We hope that Valencia provided a pleasant environment during the conference and everyone saved some time to enjoy the charm of the city.

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Development of an Active Sensor for non-Invasive Arterial Blood Pressure Monitoring

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***Abstract*—The paper presents the latest results of developing a new method of noninvasive continuous blood pressure monitoring. This method is based on the principle of pulse wave compensation. It is shown that sensors for such a measurement should be not only smart, but also active. In this connection, the concept of smart sensors is expanded to the concept of active sensors. The technical design of the active sensor for noninvasive pressure measurement is described. The results of active sensor calibration and testing are under discussion. The main section of the report is devoted to the development of software for active sensor control - its intellectual stuffing. We describe and justify a new principle of active measurement of quasi-periodic processes - pulse wave compensation based on prediction patterns. The progress achieved in research and the ways for further investigation is outlined in the conclusion.**

Keywords- smart and active sensors; compensation method of measuring; noninvasive arterial blood pressure monitoring.

I. INTRODUCTION

The progress in information technology involves accelerating development and deployment of fundamentally new medical technologies of patients' evaluation, as well as a substantial revision of the classical ones. A central problem here is the development of informative, efficient and reliable methods for measuring and processing medical and biological data of the patient.

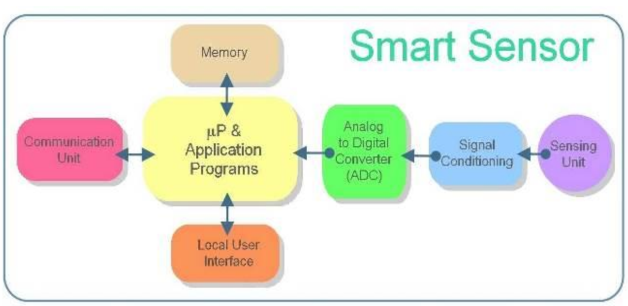

Figure 1. The principal components architecture of smart sensors [1].

One of the main trends in solving this problem, as well as in the broader field in the development of industrial control and monitoring systems is a full implementation of smart sensors, whose main task is maximizing automation of measurement and minimizing the role of man in this process [1]. An important feature of a hardware implementation of smart sensors is the use of microprocessors (μP) and as a rule the use of wireless communication units (CU), see Figure 1.

Our initial goal was to provide the features of a smart sensor for a popular medical device – arterial blood pressure (ABP) monitor. But it turned out, that the use of powerful computational tools, embedded in such a device (i.e., modern microcontrollers), can produce much better results than in the case of a standard digital tonometer. The reason for it is not in the nature of the measurement itself **-** continuous or single-sampled – it lies in the nature of the value measured. It is known that all the non-invasive measuring devices measure not the pressure itself, but some value associated with it (the volume of the blood vessel, its wall displacement, the force with which the wall acts on the sensor unit, etc.) [2]. Using *compensation method*, we succeeded in measuring the ABP in absolute units i.t., in mmHg. However, to measure the pressure by the method put forward we had to go even one-step further to introduce a feedback loop, depending on observed data sensor control. As a result, the idea of an active sensor was created, whose engineering expands the design of smart sensors as shown in Figure 2.

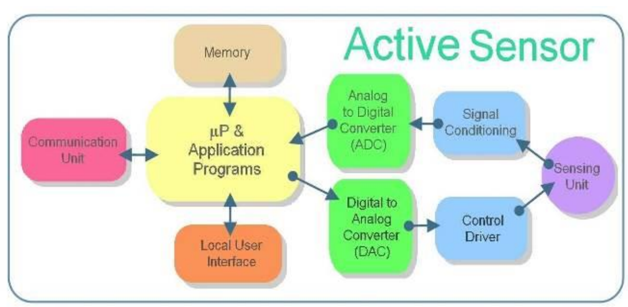

Figure 2. The principal components architecture of active sensors.

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In this paper we present the latest results of research and development concerning the active sensor for non-invasive arterial blood pressure monitoring. After a short overview in Section 2 of existing non-invasive methods and after a short discussion in Section 3 of compensation based ABP monitoring principal, in Section 4 we describe in detail the structure and design of the sensor developed. In Section 5 we report the results obtained in real experiments with active blood pressure monitoring and in Section 6 we schedule ways to overcome the shortcomings associated with the incomplete compensation of time–varying blood pressure caused by a simplified structure of the PID (proportional–integral–derivative) feedback controller. Finally, the Conclusion briefly summarizes the problems discussed in the article.

II. CURRENT ABP MONITORING APPROACHES

Jan Peňáz was among the first to put forward a method of continuous non-invasive blood pressure monitoring [3]. His method described in 1973 was aimed at reducing the risks of arterial cauterization. The Peňáz method employed the idea of "vascular unloading", based on the assumption that in the "unloaded state" the pressure inside the blood vessel is equal to the outside pressure.

The basic element of the device proposed by Peňáz is a small finger cuff (Figure 3 (A)) that has an infra-red (IR) light source on one side and a light receiver on the opposite side. The blood volume of the finger is estimated via the absorbance of IR light. The signal obtained by such a plethysmograph is further used in a feedback loop to control the pressure in the cuff. The pressure is controlled in such a way that blood volume in the finger is kept constant in time and equal to the volume which corresponds to the unloaded vessels state defined during the calibration process. In this case the oscillations of the controlling pressure are approximately equal to pressure in the arteries. Later some formulae were proposed for recalculating pressure from finger vessels to brachial arteries, which made it possible to verify the method with respect to classical procedures.

Figure 3. Modern approaches to ABP monitoring systems .

Another technique which provides continuous non-invasive blood pressure monitoring is arterial tonometry [4]. Like the Peňáz method arterial tonometry is based on pulse oscillation estimates, but here the principal of arterial unloading is different. In this case the cuff is placed on the wrist, so the sensor is over the radial artery (Figure 3 (B)). The sensor presses the artery to the radial bone until it is flattened enough but not occluded. At this intermediate position arterial wall tension becomes parallel to the tonometer sensing surface and arterial pressure is then the remaining stress (perpendicular to the surface) measured by the sensor. The pressure needed to flatten but not occlude the artery is known as the "proper hold-down pressure" and is calculated by a complicated algorithm which includes the preliminary estimate of systolic, diastolic and pulse pressures over a range of "hold-down pressures".

Currently the devices employing these methods include in particular the CNAP^{TM} (Peňáz's approach, Figure 3, left) and T-line from Tensys Medical (arterial tonometry approach, Figure 3, right).

The methods of continuous non-invasive blood pressure monitoring have both advantages and disadvantages [5]. We believe that the main drawback of these non-invasive methods is the following: irrespective of the method of vascular unloading the control of the unloading is exercised on the basis of integral parameters (blood vessels filling, overall sensor force, sensor displacement, etc.). It enables monitoring an average pulse wave of ABP but does not guarantee the details of the pulse form. We proposed a new method of arterial blood pressure monitoring that is aimed at local unloading of arterial walls by compensating local pressure.

III. COMPENSATION BASED APPROACH TO ABP MONITORING

When analyzing well–known methods of non–invasive continuous measurement of blood pressure, we can conclude that the best results of monitoring non–stationary dynamics of blood pressure are achieved by the so–called compensation methods or methods similar to them.

Compensation methods are applied for measurement of various physical quantities and are based on the compensation of an unknown measured value by controlled counter value and nullification of their difference. The simplest example of the compensation method is the use of balance scales on which unknown mass W_u is measured using a set of weights W_v, Figure 4 (A). The predetermined position of the balance beam or the associated arrow serves as a null indicator of the balance scales.

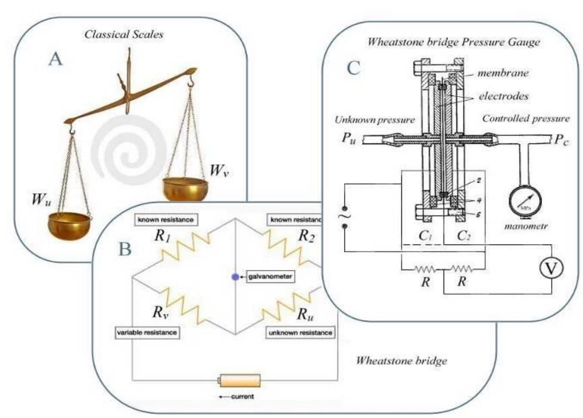

Figure 4. Measurement of physical quantities by compensation methods.

Compensation methods as methods of high precision are used for mechanical and electrical measurements, in embodiments having a bridge and half-bridge circuits, Figure 4 (B, C). Note that compensation methods are usually used to measure the static variables – constant unknown resistance R_u in bridge circuit on Figure 4 (B) and the steady pressure P_u in aggressive environment on Figure 4 (C) .

We consider the compensation method as the fundamental basis to measure the varying blood pressure. The application of this method for measuring the non–static dynamic quantity has become possible, for two reasons. Firstly, the fact that blood pressure changes are not so fast, its rhythm is of the order of one beat per second, and its spectrum fits into the range of a few tens of Hz. Secondly, there are relatively cheap, high performance microcontrollers available for which a change in pressure is quasi-static.

IV. The Structure and Design of the Sensor

As noted above, our method of measuring ABP is aimed at local, without cuffs, unloading of arterial walls by means of compensating intra–arterial pressure by controlled pressure. Our method of measurement is similar to the method of bridge measurement of unknown pressure P_u in Figure 4 (C), but it has dynamic, “adaptive” features as in the Peňáz method [3]. The appearance of the device developed for continuous ABP monitoring and its typical use for measuring intra–arterial pressure is presented in Figure 5.

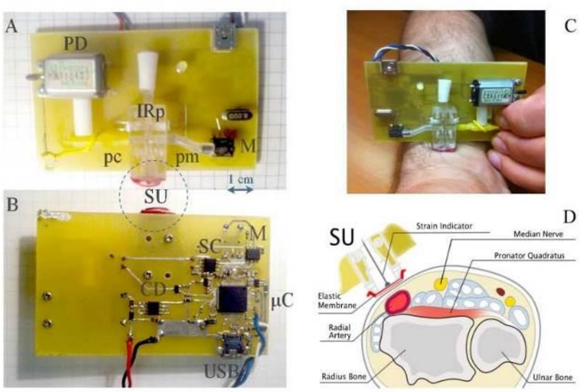

Figure 5. Active sensor for continuous blood pressure monitoring and its typical arangement on the patient’s wrist.

Note that the device developed (Figure 5(A,B)) conforms exactly to the concept of an active sensor (see Introduction, Figure 2). The main role here is played by the programmable microcontroller μC (STM32L152RBT), that has the communication I/O (USB) on one side, and on the other side it has *pm* (pressure monitor – signal line) and *pc* (pressure control – control line) interfaces to the measuring element SU (sensor unit). The signal line *pm* contains membrane displacement indicator SC (signal conditioning) and pressure gauge with instrument amplifier M, the control line *pc* includes the pump PD and the chip for pump control CD (control driver). Both lines – *pm* and *pc* – are tubes coming from the fluid-filled cavity of the measuring element SU. SU is a camera with an aperture covered by an elastic membrane (of red color in Figure 4(D)) and containing a thin rod with one end attached to the membrane, the other end partially overlaps IR-radiation flux inside optoelectronic infrared pair IRp. Thus, this rod gives us a way to measure the membrane deformation as it serves as an indicator of displacement.

When the pressure in the SU chamber and directly behind the membrane are different, the membrane will be deformed in one or the other direction by moving the rod in the direction of deformation. This movement will bring about a change in the area of IRp flux, which is registered by displacement indicator SC. If the controlling pressure in the SU returns the membrane to flat, not deformed state, we can conclude that pressures on both sides of the membrane are equal, thus external pressure will be measured. If SU is arranged above the radial artery, as shown in Fig. 5(C,D), then, assuming that at certain SU position the external pressure is equal to ABP, we will measure its value. This is precisely the key concept, underlying our compensation measurement technique.

V. Experiments with the Sensor Developed

The above qualitative description of the sensor can be illustrated by quantitative results obtained in the device calibration and testing. Figure 5(A) shows the deformation of the membrane displacement (in ADC units of SC indicator) in response to a uniform increase over time of the pressure P inside the SU at constant external atmospheric pressure. ΔP in Figure 5(A) is the pressure readings of M measured from atmospheric pressure, so ΔP is the pressure difference across the membrane. It can be seen from the graphs that for small (± 5 mmHg) difference in pressures on both sides of the membrane the signal of displacement indicator SC is linearly proportional to the pressure difference. Figure 5(B) shows the dynamics of the membrane displacement and the corresponding changes of pressure P within the SU when external pressure uniformly increases and control line *pc* is enabled for its compensating.

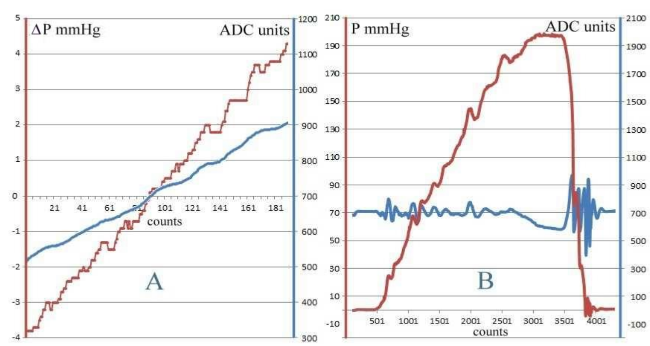

Figure 6. The dynamics of the SU membrane displacement (ADC units) in response to changes of the pressure difference inside and outside the SU.

The compensating pressure P in SU is produced by the pump when the PWM (pulse-width modulated) voltage, formed by CD chip, is applied to the PD.

Not specifying the details of the mechanism of formation, the resulting P may be thought of as proportional to the control signal, which the microcontroller sends to CD in response to the measured by SC membrane displacement (as well as its previously stored values). The algorithm of simple PID [6] controller was originally selected as feedback control algorithm.

On the basis of the calibration data numerous experiments on monitoring blood pressure were carried out. The results of one of the measurements are given in Figure 7. The lower part of the figure graph shows that the use of the PID control makes it possible to hold the membrane close to undistorted (flat) state for all the time of blood pressure measurement. The quality of such regulation can be estimated by the value of uncompensated difference in pressures equivalent (proportionate) to deviations of the membrane from the flat position (a kind of the membrane jitter). To illustrate the compensation, Figure 7 (top) shows a graph of pressure P inside the chamber SU (the data from pressure gauge M), as well as the same data corrected by the membrane jitter values.

It can be seen that in comparison with P uncompensated difference in pressures ΔP is not large. Figure 7 also shows that P coincides generally with the corrected blood pressure, but at times with the rapid change in the shape of the pulse wave, the controller fails to keep track of these changes.

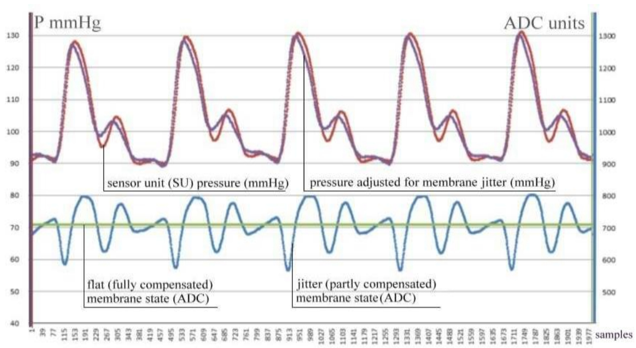

Figure 7. The dynamics of the SU membrane displacement (ADC units) while compensating the time-varying outside blood pressure.

The overall impression of the first results of monitoring blood pressure with the help of the above–described active sensor is ambivalent. On the one hand, the sensor does its task – it can be used not only to measure the main parameters of blood pressure – systolic and diastolic blood pressure (range of ABP), but also to observe the varying pulse wave, and not through indirect measurements but by measuring the pressure itself (in mmHg). On the other hand, as we could initially assume a simple PID controller does not provide full compensation, which leads to the distortion of the pulse wave components. Practically, the compensation algorithm based on PID control proved to be very difficult in changing its settings.

VI. Pulse Pressure Compensation Control on the Basis of Pulse Wave Patterns

These shortcomings associated with the incomplete compensation of time–varying blood pressure have a simple explanation.

Firstly, the PID controller used is the special case of linear regulators class and it is well known that linear methods for treating biological signals, especially, linear adaptation methods are used in limited ranges of their changes and only under strictly controlled (for example, laboratory) conditions. This is due to highly non-linear non–equilibrium nature of living systems [7]. Even a small change in physiological state can lead to considerable changes in the result.

Secondly, PID control takes into account only local characteristics of the signals, as it is customary in the theory of dynamic systems. In the case of dynamic systems, such a regulation is natural, as such systems are deterministic. However, living systems and their subsystems are known to be poorly described by models of dynamic systems, even if there is a freedom of choice of corresponding differential equations coefficients [7]. Living systems are much more consistent with models of stochastic, non–deterministic systems.

Thirdly, in solving technical problems low order of PID control is preferred due to its ease of implementation. When we deal with a complex biomedical signal, particularly, an ABP signal, low order of control is a drawback.

These facts indicate that the pulse wave of blood pressure is much more similar to a wideband pulse signal than the sum of harmonic components. In view of the foregoing it can be assumed that a random point process is a closer mathematical model of blood pressure pulsating. A point process is an increasing sequence of time moments (points) of certain homogeneous events, such as the arrival pulse wave, with a random length of time intervals between them. Excluding the changes in the pulse wave pattern, and treating them as a stream of homogenous sequence of events, it would be legitimate to use theoretical methods dealing with the problems associated with the heart rate.

One such well-developed theory today is the theory of radar signal processing. A major problem of this theory is determining the unknown arrival time of an electromagnetic pulse emitted by radar and reflected from a target. The method of matched filtering should be noted as one of the most effective methods for solving this problem. In this method the filter response is matched with the radar pulse, so that the maximum signal of the filter output is observed at time, when the reflected pulse arrives. Essentially, the matched filter generates a covariance of sent and received pulses, and it is well known that if both pulses have the same, up to a constant factor, shape, the maximum of the covariance will be achieved when they coincide. The corresponding displacement between pulses will estimate the time of arrival.

The principle of the matched filtering could be interpreted in a different way, more suitable for our tasks. Let’s assume at some moment, the time T between the ABP pulse wave is known along with the shape of the current pulse, further considering the shape of next pulse to be same as the current one, we can easily predict the future signal.

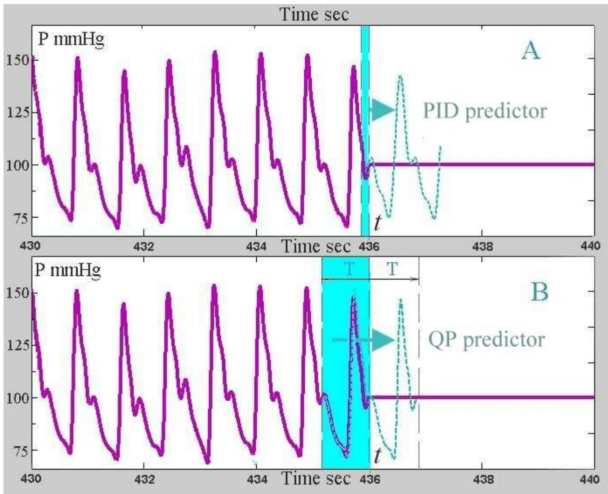

Figure 8. Two methods of predicting the pulse signal future: A - by PID regulation and B - based on the local quasi-period (QP) estimate.

For this purpose, we should take an existing fragment of the signal of duration *T*, immediately preceding the current time and move it to the time *T* into the future (see Fig.7). From the theory of the matched filter, it follows that the expected ABP pulse will coincide with the prediction. This idea is illustrated in Fig.7, which also shows a comparison of the proposed signal prediction with the prediction carried out by PID. As stated above, the PID controller makes a prediction on the basis of current, local estimates of the fundamental frequency, the amplitude of the main component and amplitudes of neighboring harmonics.

VII. Conclusion

Summing up the results of the investigation we can conclude that when we use compensatory ABP measurement it is the current pattern of the pulse wave which efficiently predicts the expected signal. This pattern must be dirigible enough to change significantly with changes in the state of the object measured as well as changes in the conditions of its active measurement.

For this reason, realized by the classic regulators including the PID controllers, patterns in the small parametric models of ABP waves are of little use for active blood pressure measurement. An idea of forming the adequate patterns in the task is given in the above qualitative reasoning. It is based on the property of quasi-periodicity of ABP signal.

This property lies in the fact that high variability of the period of heart contractions occurs only at long time intervals, whereas at short time intervals of several seconds or a few heart beats its changes are generally small and fall within a couple of percent.

Therefore, estimating the current period, more precisely the quasi-period *T*, we can get a pulse wave pattern as a signal fragment of *T* duration immediately preceding the current time moment. Thus, the task of building a pattern for current pulse of the signal is reduced to the task of effective evaluation of its current quasi-period.

Acknowledgment

This work was supported by RFBR (Russian Foundation for Basic Research) grant 14-07-00496-a.

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 ISBN: 978-1-61208-480-0

Mobile and Personal Speech Assistant for the Recognition of Disordered Speech

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***Abstract*—Recently, Assistive Technologies tend to exploit speech-based interfaces as a means of communication between humans and machines. While they perform very well for normal speech, their efficacy is very limited for people suffering from a variety of speech disorders. Moreover, in the systems targeted for disordered speech, the recognition performance is highly diminished by the environmental factors related to the disease. This limits the practical applicability of these solutions. To overcome this problem, we propose a Mobile and Personal Speech Assistant (mPASS) – a platform providing the users with a set of tools, which enable to intuitively create their own speech recognition system corresponding to their needs and capabilities. Our long term vision is that handicapped users, without computer science and artificial intelligence knowledge, will use this platform to design at home their own speech recognition system, tailored to the domain, vocabulary, and language they find most useful. As a result, a personalized speech recognizer will be created, which can be used with diversified speech-based applications.**

***Keywords*–dysarthric speech recognition; personal speech assistant; speech recognition for assistive technologies; mPASS platform.**

I. INTRODUCTION

The ability to speak, communicate and exchange thoughts is one of the fundamental needs of human beings. Unfortunately, it cannot be sufficiently satisfied in case of people suffering from a variety of speech disorders. As a result, communication situations, which are natural part of everyday activities, can become a formidable obstacle requiring help of an accompanying person. In addition, current technological achievements in the fields of ambient and assisted living, control of smart devices, smart homes, etc. tend to exploit speech-based interfaces as a core means of communication between humans and machines. Moreover, motor functions impairments, which call for the use of Assistive Technologies, are very often associated with speech production problems. Standard automatic speech recognition (ASR) systems, targeted for regular speakers, perform very poorly for people with speech disorders [1]–[3]. Hence, a significant group of people is not able to use many voice-controlled state-of-the-art technology advances, which could support independence in handling their daily activities.

It is estimated that 1.3% of the population encounters significant difficulties in speech-based communication [4]. The ability to use speech-based interfaces would significantly improve the lives of people suffering from speech impediments, in particular those with accompanying motor skills disorders. However, there are many diversified speech disorders and it is very challenging to design a single ASR system, which could recognize the impaired speech in each particular case [3]. Traditional methods of constructing ASR systems, used with success for normal speakers, fail in such a task – they require large-scale databases, which are not feasible to be created for disordered speech. Adaptation of standard ASR systems to the disordered speech led to the very limited system performance [3][5][6]. There have been several attempts to design a speaker-dependent dysarthric speech recognition systems [1]–[9], but they were trained mainly in the laboratory environments. Only a few of them were created and tested in real usage scenarios [2][4] with the limited achieved performance, which was not sufficient for the practical implementation [4].

The design of a disordered speech recognition system with a good recognition performance for diversified speech impediments is very challenging. In order to increase the practical application of disordered speech recognizers in Assistive Technologies, we present a concept of a mobile and Personal Speech ASSistant (mPASS) – a platform providing the users with a set of tools for building an ASR system, which is tailored to their speech disorders, needs, and capabilities. The mPASS toolchain is designed for non-technical user – the expert knowledge, in particular the knowledge about speech recognition, is not required. One of our key goals is a user-centric interface design allowing to use the platform by people with motor functions impairments and other disabilities. The user can choose the scope, in which he/she wishes to use the system, record training samples, and create personalized speech recognizer, which can be later used as a core engine for different speech-based endpoint applications. In case of people with severe motor disorders and/or accompanied intellectual disabilities the help of a user's carer or other person can be mandatory to operate the system, however the technical background of such a person is not required.

The mPASS platform is to be exploited at users' home. Therefore, the users are not obligated to attend long recording sessions at a remote location, which is a significant obstacle for the handicapped users. By maximizing their comfort, more speech samples can be collected and, at the same time, users' motivation to work with the system is improved. Moreover, the samples are recorded in the environment in which the ASR system will be later used – this should increase the recognition performance. Such an approach was never practised for a disordered speech thus far. By realizing this idea, we envision that we will be able to engage in our study many users, who will create different types of ASR systems, addressing diversified needs and being successfully used in many practical deployments.

This paper is organized as follows: Section II provides a brief overview of related work, while Section III depicts design challenges that are driven by the analysis of previous approaches. Sections IV and V present the mPASS solution and its architecture. The preliminary results are discussed in Section VI and Section VII concludes the paper.

II. RELATED WORK

In the recent years, an increased attention has been put towards the design of disordered, in particular dysarthric, speech recognition systems (dysarthria is the key group of speech disorders) [1]–[9]. The investigated related works were mainly targeting the limited-vocabulary, discrete speech recognition systems focused on the command and control target applications. The final dysarthric speech recognition system was task specific and could have been used only with one, selected, speech-based application. This assumption was driving the methodology selection and ASR system set-up. A common practice was also to use the speech recognizers designed for natural speakers and adapt them to dysarthric speech (e.g., Dragon Dictate, Swedish solution Infovox or traditional models based on the Hidden Markov Model (HMM) solutions) [1][5][6]. The performance of these recognizers was limited, especially in case of severe speech impairments. Although, in general, the top performing systems presented 80-90% of accuracy, they were obtained in the laboratory conditions. The trials conducted in more realistic environment revealed that the external factors (such as background noises) significantly degraded the investigated systems to unacceptable levels [4][5]. Substantially, the diminished performance did not allow for practical exploitation, as concluded from the the year-long project VIVOCA [4].

III. Design challenges

The analysis of related works led to the conclusion that the system performance in normal, practical usage situations is influenced by the degree of speech disorder and motor functions impairments, environmental factors (e.g., noises), system access technology design, etc. User motivation was also thoroughly depicted by other researchers as a crucial element of a successful system usage. From the performance perspective, it was assessed as even more important than a degree of speech impairment – better motivated users with severe disorder can train the system better than less motivated ones with milder disorder [5]. These factors have significant impact on the design of an ASR system as a whole.

Taking into consideration the outcomes of the related works, it turns out that the challenges in the design of such a system for the disordered speech focus on two factors:

1) the core speech recognition technology, which calls for the development of new techniques targeting disordered speech, especially with regard to acoustic modelling
2) disability-oriented, user-centric system design, taking into account the user needs, which allows for a comfortable usage in the presence of accompanying difficulties

Usually, the second factor is perceived as much less important, especially at the research stage of product development, and it does not influence performance. However, when designing the system for the demanding and diversified group of, often handicapped, people, its importance becomes equally relevant as the technical excellence of the core speech recognition technology. Hence, our goal is to address both these challenges and come up with a solution which would conveniently combine novel research outcomes with the user-centric design. Substantially, we also perceive a positive practical verification of a solution as a key challenge as well as an important success measure.

IV. mPASS approach – Mobile and personal speech assistant

To address the above challenges, we propose a platform which allows *non-technical users* to build their own speech recognition systems, tailored to their particular needs and speech disorders. Our vision is that disabled users, without computer science and artificial intelligence knowledge, will use the mPASS platform to define the domain, vocabulary, and language that is most useful for them in order to communicate effectively with the outside world. They will then train their own ASR system and adapt it to their individual way of speaking. The mPASS system allows to create different types of speech recognizers, at different levels of complexity, ranging from small-vocabulary, command-based systems, to dictation-based systems with different vocabulary sizes for the recognition of sentences and phrases. The more complex systems are envisioned for people with mild and moderate speech disorders, since the users with severe speech disorders usually do not use speech in such broad contexts.

The personalized speech recognizer can be used later on with many diversified speech-based applications. The proposed mPASS platform is available on a desktop computer as a web-based application providing tools for creating user- and task-dependent speech recognition systems. The models created and trained with this application can be than ported to a mobile device and used in the final speech-based application of interest (where the models for the disordered speech need to substitute or complement the ASR models for the natural speech). Hence, the speech recognizer built by using the mPASS toolchain *can be used with many different speech-based applications*, which were not available to the disordered speakers thus far. Those applications are widely exploited in the environmental control systems, command-and-control systems (e.g., to steer some home appliances with voice commands), control of mobile device functions, exploited in converters transforming (possibly disordered) speech to text or to a synthesized speech, and many more. Some examples of such end-point applications, currently being developed by us to showcase the capabilities of the mPASS technology, are:

1) dictation-based, task-specific application allowing to “translate” impaired speech during a conversation in a restaurant, bank, at the doctor’s office, etc.
2) educational game, targeted for autistic users, aiming at helping them in speech therapy classes
3) mobile communication application for users with very severe speech disorders and motor skills impairment (the user exploits a few sounds he/she can produce to control an image-based “communication book”)

Having in mind the identified challenges, we present below the key objectives the mPASS aims to accomplish. They also constitute the differences between our approach and the related works.

In contrary to other approaches, the process of building a disordered speech recognizer with mPASS should be *automated* and should limit the need for external help to minimum. Since the influence of practical usage constraints is tremendous, they should drive the system set-up.

The ASR system should be created *at user’s home* and a training process can span across longer period of time, if necessary. Thus, the time spent on training the recognizer

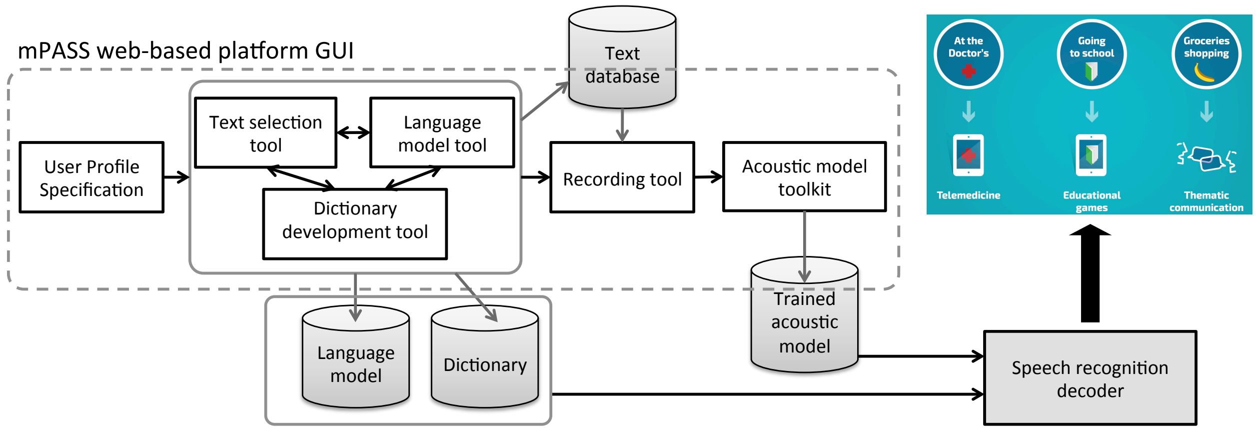

Figure 1. Mobile Personal Speech Assistant architecture – an overview.

can be adjusted to the user's health condition, motivation and other factors. Such an approach also minimizes the problem of reduced performance in case of systems which were trained in the silence conditions, but are used in the environment with existing background noise.

Finally, the mPASS toolchain is intended to allow for the *exploitation of existing resources*, which are proved to be good for creating speech recognition systems. *Novel approaches* are to be provided only where necessary, e.g., while building acoustic models for dysarthric speech, where we are developing a new method of the dysarthric speech recognition based on the modified speech classification methods.

At this stage the targeted language is polish, however the platform by design is language-agnostic and could be used for building speech recognizers for other languages as well.

V. SYSTEM ARCHITECTURE

The mPASS platform guides the user through the steps required to build the speech recognition system (Figure 1). During the process the user follows the on-screen instructions. The core part of the platform is a web-based application – a client side is implemented by using *AngularJS* framework and the server side is based on the *Node.js* framework. The voice is captured by the HTML5 function *getUserMedia*. The client and the server exchange data in the JSON format. The speech recognition system trained with this application is then incorporated with a target speech-based application, on a mobile or embedded device. The below steps present how the process is organized and which consecutive actions are expected to be executed by the user:

1) The user has to create a *profile* which is strictly related to the level and scope of the envisioned system usage (e.g., command-based, recognition of sentences, continuous speech). There can be different profiles created for the same user, each targeting different kind of speech recognizer for different tasks (e.g., containing vocabulary/training sets for controlling TV, going to doctor's office, restaurant, etc.). Based on the selected system level, the baseline speech unit is automatically defined as word, syllable or phoneme.
2) **Creating texts to be recorded, dictionary and language model**: These elements are usually combined and they influence each other. For instance, in command-based ASR systems it could be most convenient to start with a vocabulary, while for the other ones it could be better to start with a set of texts for recording. The mPASS toolchain further guides through the next steps, including support for intuitive creation of language model and dictionary. The final relation between text selection, dictionary and language model is proposed automatically.
   a) **Text selection tool**: it is equipped with several phonetically balanced and phonetically rich texts for polish language. They have been created based on a well-known poems and short stories for children in order to make them easy to pronounce by the disordered-speech users. It is also possible to create the text automatically based on the existing dictionary and language model.
   b) **Dictionary tool**: Dictionary contains the list of words that the system will be able to recognize. It can be created either manually or by extracting words from the texts selected previously for recording or from the language models defined by the user. It is also envisioned that the dictionary tool will automatically suggest additional entries that could maximize ASR performance. For that purpose the dictionary will be analyzed by the mPASS platform in terms of length of the words, phonetic differences between them, and others. There is also an option to substitute frequently unrecognized words with their synonyms based on the user input or automatic suggestion from the mPASS system.
   c) **Language model tool**: The purpose of this tool is to create grammar or statistical n-gram language models. In the first case the user is supported to manually create grammar rules via dedicated interactive graphical interface (technical knowledge is not necessary at this step, initial examples are provided automatically). Alternatively, the mPASS system can automatically modify the pre-loaded generic statistical n-gram model for a given language, in order to align it to the scope of the desired ASR system.
3) **Recordings**: The user records selected texts and/or word lists. There is a minimal suggested number of recordings specified. In addition, the system also gives a possibility to add new recordings at a later time, pause and resume the recording sessions. The tool also allows to play additional audio information on the attached headphones. The supplementary audio-visual information is supposed to help

people with intellectual disabilities, visual impairments, children, etc. We also aim to supply the tool with mechanisms allowing for monitoring and potential correction of wrong recordings – the user will be given a real-time feedback information.

4) **Training the acoustic model**: This step is an automated background process. Only experienced (developer-type) users are allowed to change some of the parameters, e.g., choose different methodologies/techniques, such as HMMs or Support Vector Machines (SVMs). We are also developing our novel acoustic modelling methods, which will be included in the mPASS system.
5) The obtained acoustic model, dictionary and language model are then *exported* to be used in the desired target speech-based application. Optionally, the initially created acoustic model can be later on extended based on additional recordings collected while creating other user profiles for different contexts.

All recordings, recorded texts, dictionaries and language models are stored in a database. The user may wish to share them with others (if agreed) in order to help develop better ASR systems for the other users in the future.

From the user perspective, the recording tool functionality is the most important part of the mPASS platform. It is, however, also most vulnerable to possible errors – wrong recordings, additional background noise and other factors affecting the recorded material will directly influence the acoustic model and its performance. Hence, in order to tune our interface design and system features to real user needs, we have performed initial recording sessions with several users having diversified speech disorders: one adult with explosive speech and associated motor impairments, 4 teenagers presenting variable levels of dysarthria and 4 healthy children 3-6 years old with impaired speech typical to their age. Those trials helped to improve the system design and obtain initial database used by us for the evaluation of acoustic modelling techniques. Currently, the key components of the mPASS platform are implemented and it can be used for further evaluation.

VI. Preliminary results

Initial performance trials were executed by the adult with explosive speech and cerebral palsy. With the mPASS platform, he created an ASR system for the exemplary voice-controlled mobile application, which allows to send an SMS or e-mail with one of predefined messages [10] to a recipient from a phone contact list. User-defined voice commands are used to control the application. The user recorded 8 messages of his own choice (e.g., “I will be back in 1 hour”) and several action commands (“up”, “down”, “OK”, etc.) - all together 21 phrases, 30 times each. The ASR was using the HMM-based acoustic model. The recognition performance obtained in the laboratory environment was approx. 99%, whereas in a real environment (home/office) on average 84%. Additionally, we investigated performance measures related to the person’s judgement of system’s applicability and usability. We compared the time required to complete particular actions, including the time lost for necessary repetitions when recognition errors occurred, with the time needed for the same action to be completed by using the regular touch input (the person controls mobile phone installed on a wheelchair with his chin). The results were averaged over 20 trials – they presented that the voice-controlled version outperformed the manual entry for up to 49% – considering the time gain which was observed with the voice input in comparison to manual input (Table I). Substantially, the user assessed a voice-controlled mobile speech assistant as the preferred option, which is the most important success measure.

TABLE I. Comparison of the time required to complete an action with a voice-controlled and manual entry

Action	Voice input	Manual input	Gain
Send SMS to caregiver	31s	56s	45%
Send e-mail to caregiver	33s	65s	49%

The initial trial presented above constitutes a first proof-of-concept evaluation. At this stage, the obtained performance results cannot be directly compared to the ones presented in the related works, since they were gathered for different usage scenario and with different ASR system, especially with regard to the selected vocabulary. However, in general, the system performance reached very high levels for the laboratory environments, often higher than those reported in related works. They were accompanied with a very promising outcome for the real usage environments, which was rarely achieved before. More detailed performance evaluation of the ASR systems created with the mPASS platform, based on the database of recordings collected from another 7-10 users, is a part of the future work.

VII. Conclusion and Future Work

The mPASS system proposes a unique combination of an intuitive, user-centric system design with the top performing ASR tools. It provides an automated toolchain, which enables to easily follow the process of creating a speech recognition decoder. We believe that by using this technology the wide variety of users, with different speech impairments, will be able to build disordered speech recognition systems – tailored to their needs and achieving high recognition performance. Substantially, the users will be allowed to create and train the system at home environment. The initial results are very promising, especially taking into account a positive users’ feedback. Our findings revealed that the voice-controlled input was perceived as up to 49% better than traditional manual input by a person with severe speech impediments and motor skills disorder. In the future, we plan to evaluate the mPASS platform with more users in several scenarios related to different mobile applications, which will be based on the ASR systems trained with mPASS. By using the proposed toolchain, we hope to achieve disordered speech recognition systems ready to be used in practical conditions with a variety of endpoint speech-based applications. Hence, our solution could be effectively exploited by people with speech impairments and assist them in their daily activities.

Acknowledgements

The presented work is financed by the National Centre for Research and Development in Poland under the grant no. LIDER/032/637/l-4/12/NCBR/2013. The authors would like to thank Michał Koziuk for his help with the system implementation.

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Touchscreen Thimbles: Enabling Intuitive Interaction

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***Abstract*—We are interested in the challenge that the ever-increasing use of capacitive touchscreens represent. This is because they rely on the electrical properties of the human skin, which changes over time, often becoming less conductive with age. So, we consider the potentially increasing problem this would represent, with a layered model of skin interactions, as a frame from which to conceptualise different classes of approaches and better understand the design requirements. Conscious of the need to retain a sense of touch, rather than a stylus with a conductive tip or gloves with conductive finger tips, we then propose suitable thimbles for touchscreens. We conceptualised a *touchscreen thimble* to be worn on the body that would maintain a sense of touch. So it would be a pros-palliare, rather than a prosthesis. Developing a *proof of concept* prototype consisting of a rigid base combined with a conductive fabric for the finger tip, to address the limitations we observed with existing approaches. We conclude by considering the wider potential of our idea in designing a preferable future for touchscreen use. Including, the potential for multiple *touchscreen thimbles* to be worn simultaneously to support multi-touch gestures, and for them to be worn as an enhancement for able-bodied users.**

Keywords–Electrodermal; Conductivity; Skin; Age.

I. INTRODUCTION

The ever-growing availability of capacitive touchscreen displays is largely attributed to the intuitive interaction they afford [1][2]. They rely on the electrical properties of the human body to detect when and where on a display the user touches [3]. So, they can be controlled with light touches of fingers [4], which is contributing to their dominance for device interaction [5]. The touchscreen enables the user to interact directly with what is displayed, rather than via a mouse, touchpad, or other intermediate device [3].

In Digital Cultures, the practices and socio-cultural meanings of emerging digital technologies, the popularity of smartphones and tablets is driving the acceptance of capacitive touchscreens for many types of information appliances [6]. This is because the usability of capacitive touchscreens is considered preferable to other forms of device interaction [7]. Often where computer keyboards and mice do not allow a suitably intuitive, rapid, or accurate interaction [3]. So, there is a growing trend towards capacitive touchscreens for user interfaces, shown by their increasing integration into the design of different products. So much so, that they are starting to become unavoidable given their ever-increasing availability in devices [5]. However, for those with limited *electrodermal conductivity* of the skin, they can be difficult to use [8]. This can be acute as *electrodermal conductivity* of the skin is known to decrease with age [9].

In Section II we will introduce the Seven Skins conceptual framework to consider a preferable solution to skin with limited electrodermal conductivity. In Section III we define, explore and prototype our *touchscreen thimbles*. In Section IV we conclude by consider the benefits and limitations of our approach, as well as potential future work.

II. SEVEN SKINS

In considering a preferable solution to skin with limited *electrodermal conductivity*, we explore assistive sensor devices (ASDs) with reference to Professor Irene McAra-McWilliam's 'Seven Skins' concept.

1) Worn separate device:
-fitness tracker, phone or watch as sensor
2) Sensors integrated in clothes or wearable layers:
-smart textiles
3) Passive and attached to the body:
-biostamp temporary sensor tattoo
4) Interactive sensors on the body:
-myoelectric prosthetic hand
5) Interactive sensors integrated with the body:
-targeted muscle reinnervation prosthetic
6) Passive and embedded in the body:
-sensor pill monitoring vital body signs
7) Active and embedded in the body:
-pace-maker

The first, outermost, layer considers ASDs worn on the body. These can generally be grouped into devices that have come to include sensors, such as smart phones and watches; and those with sensors dedicated to monitoring physiology, such as *fitness trackers*. The next, the second, layer considers ASDs integrated into clothes or wearable layers, such as smart sports shirts [10]. The third, emerging, layer considers ASDs fixed on the body, such as the recent temporary bio-stamp sensor tattoo [11]. We consider these to be passive because they only monitor and are not interactive.

The middle, fourth, layer considers ASDs on the body, such as myoelectric sensors for gesture control of a prosthetic hand. While the sensors of the prosthetic hand can be considered to be within the second layer, they are part of a system which is perceived to have a deeper attachment to the body at the fourth layer. This is because the prosthetic hand is perceived to be an extension of the body [12].

The fifth layer considers Interactive ASDs integrated with the body, such as those that enable targeted muscle reinnervation for real-time control of multifunction artificial arms [13]. The next, sixth, layer considers passive ASDs that are embedded in the body, such as tiny pill monitors that can measure vital signs from deep inside the body [14]. The innermost, seventh, layer considers ASDs embedded in the body, such as pace-makers.

III. TOUCHSCREEN THIMBLES

Considering a potential ASD for limited *electrodermal conductivity* of the skin, it would preferably be at the first layer in the seven layer *skins* explored in the previous section. So, a worn device that would compensate for limited *electrodermal conductivity*. Furthermore, it would preferably maintain a sense of touch to ensure intuitiveness.

A stylus with a conductive tip would work, such as the Jot Mini or Apple Pen. However, it would be far less intuitive, because it would lack a sense of touch. Gloves with conductive finger tips would also work, such as the iGlove or Totes SmartTouch. However, they would not be well suited for indoor use. Other approaches provide finger nail like extensions, intended to help provide accurate interaction for individuals with larger fingers, such as the Tech Tips. However, such approaches also lack a sense of touch.

We conceptualised a *touchscreen thimble* to be worn on the body that would maintain a sense of touch, while being suitable for indoor as well as outdoor use. So it would be a pros-palliare, instead of a prosthesis, as it is an *addition* that *covers* rather than *replaces*. We explored our concept as shown in the sketch of Figure 1(a), being in the first layer of the *seven skins*. A thimble consisting of two essential parts, a rigid cylindrical base, and a thin conductive fabric for the finger tip. With the conductive fabric ensuring an intuitive sense of touch would be maintained, while compensating for insufficient electrothermal conductivity by being highly conductive. We then explored suitable materials, developing a number of prototypes, as shown in Figure 1(b). Including a *proof of concept*, which consists of a rigid base made from antler, combined with a grey conductive fabric for the finger tip. While any rigid material (e.g. ABS plastic) would have been suitable for our *proof of concept*, antler was chosen because of experiences of exploring its aesthetic qualities in addition to being suitably rigid.

IV. CONCLUSION

For those with skin that has limited *electrodermal conductivity* [8][9], the ever-increasing availability [5] of capacitive touchscreen displays could become challenging. However, our *touchscreen thimbles* would ensure that the intuitive usability of capacitive touchscreen displays would be maintained for them. Also, multiple *touchscreen thimbles* could be worn to support multi-touch gestures. Future development should test a prototype with potential users, especially non-technical users, to provide experimental evidence of the effectiveness of our proposed solution. Specifically, experimental confirmation of the increase of efficiency in interaction compared to bare fingers, and the increase of intuitiveness compared to alternative solutions (e.g. conductive stylus). Given that the use of *touchscreen thimbles* could indicate that users have problems in using touchscreens, some users may be self-conscious about their use. So, consideration should be given to the aesthetic, as well as functional, design. For example, considering the potential for personalised *touchscreen thimbles*, not just for accurate sizing, but also for custom styling. An *innovation through tradition* design approach, which enables connecting new technology with familiar traditions (e.g. textiles) could be helpful in determining desirable aesthetics for the adoption of *touchscreen thimbles*.

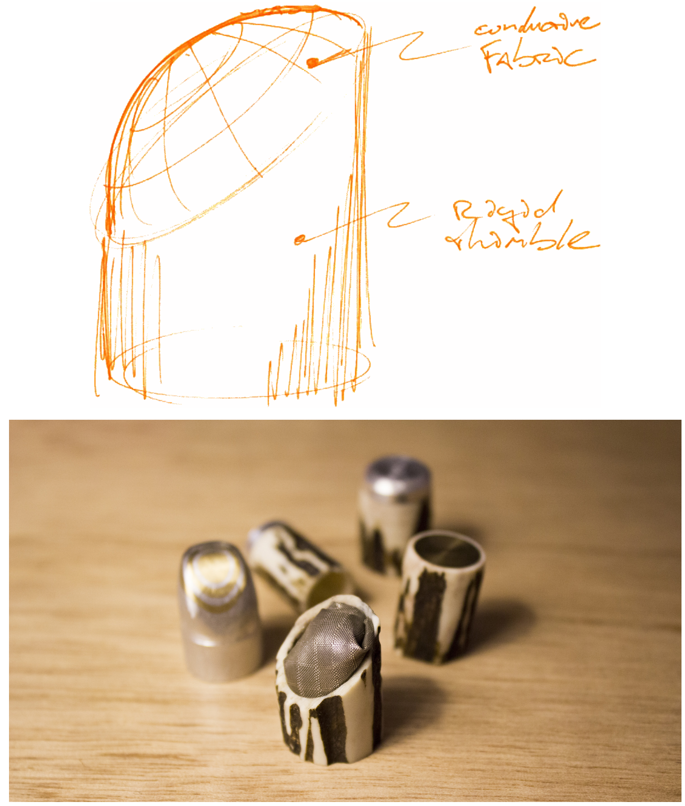

Figure 1. Sketch (a)bove showing the design of the thimble consisting of two essential parts, a rigid cylindrical base, and a thin conductive fabric for the finger tip. Photo (b)elow showing a number of thimbles resulting from exploring suitable materials, with the front thimble showing a rigid base made from antler combined with a grey conductive fabric tip.

Where capacitive touchscreens are used for control in critical situations, *touchscreen thimbles* could be worn as an enhancement for able-bodied users, ensuring intended interactions when under duress. Similar to how the able-bodied can benefit from directional hearing aids, because they improve the signal-to-noise-ratio of speech occurring in noisy backgrounds, irrespective of hearing disability.

The seven layer *skins* of ASDs provided a framework from which to conceptualise different classes of approaches in the development of our *touchscreen thimbles*. So, future work should consider further development of the framework, as well as its applicability to *spaces* of a similar nature in designing preferable futures. While the *touchscreen thimble* prototype worked anecdotally as a *proof of concept*, future work would be required to confirm their effectiveness more broadly. Furthermore, future work regarding the conceptualisation of *touchscreen thimbles* should consider that while they would be worn on the body as ASDs, they interface beyond the body to a networked device. So, potentially interfacing to a meta-level beyond the touchscreen through the Internet.

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ACKNOWLEDGMENT

We would like to thank Professor Irene McAra-McWilliam for describing her 'Seven Skins' concept to us.

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Intarsia-Sensorized Band and Textrodes for the Acquisition of Myoelectric Signals

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***Abstract*— Surface Electromyography (sEMG) has applications in prosthetics, diagnostics and neuromuscular rehabilitation, and has been an increasing area of study. This study attempts to use a fully integrated smart textile band with electrical connecting tracks knitted with intarsia techniques to evaluate the quality of sEMG acquired by knitted textile electrodes. Myoelectric pattern recognition for motor volition and signal-to-noise ratio (SNR) were used to compare its sensing performance versus the conventional Ag-AgCl electrodes. Overall no significant differences were found between the textile and the Ag-AgCl electrodes in SNR and prediction accuracy obtained from pattern recognition classifiers. On average the textile electrodes produced a high prediction accuracy, >97% across all movements, which is equivalent to the accuracy obtained with conventional gel electrodes (Ag-AgCl). Furthermore the SNR for the Maximum Voluntary Contraction did not differ considerably between the textile and the Ag-AgCl electrodes.**

Keywords—Textrodes; Smart Textiles, Pattern Recognition; Intarsia.

I. INTRODUCTION

Surface Electromyography (sEMG) has applications in prosthetics, clinical diagnostics, and neuromuscular rehabilitation devices. sEMG has been used for rehabilitation robotics, treatment for stroke and spinal cord injury patients [1], phantom limb pain treatment [2], and as a tool for non-invasive EMG monitoring [3].

Traditionally Ag-AgCl electrodes are used to acquire sEMG signals because the electrodes limit motion artifacts due to its conductive layer and, most often, ensure high quality signal acquisition. However, these electrodes when used for extended periods of time cause skin irritation [4]. Additionally when used in upper limb applications for prosthetics, and therapies for amputees, they are difficult to apply and remove with limited dexterity.

The use of smart textile technology to address these problems in sEMG signal acquisition has been an increasing area of study as well as the study of textile applications in biosignal monitoring such as Electrocardiography (ECG) and Electroencephalogram (EEG) [5][6]. Textile electrodes, also known as textrodes, have been studied extensively in multiple forms including screen printed, knitted, woven and embroidered sensors [6-8]. In the case of the study by Zhang screen printed electrodes for sEMG monitoring showed promise in movement identification for transradial amputees using both offline classification techniques as well as live recordings [7].

This study proposes a fully integrated textile solution for sEMG monitoring in the form of an arm-band fabricated using intarsia knitting, for electrical connection of the recording device with the textrodes. The textrodes were done with a knitted silver fabrics similarly to those used in [9].

Intarsia is a well-known and spread knitting technique in textile manufacturing that enables textile electronic integration at the level of fabric production by using conductive yarns which form knitted courses through the fabric. This technique has been previously introduced for e-textiles for ECG recordings [10], electro-stimulation [11] and even for thoracic bioimpedance recordings [8][9].

Using this sensorized armband and textrodes, this study aims to assess:

1st The performance of the textile electrode for sEMG recording on the upper limb

 ISBN: 978-1-61208-480-0

2nd The feasibility of using a fully textile sensorized arm-strap for the application of classification of hand-movements from sEMG recordings

For that, the functionality of both textrodes and Ag-AgCl electrodes was studied using the intarsia-sensorized band, performing a study based on their offline pattern recognition accuracy, i.e. the performance of classifiers using signals for the two types of electrodes was analyzed. The SNR was then used to compare the signal quality and floor noise for the two electrodes during Maximum Voluntary Contraction (MVC).

Section II describes the methods for textile fabrication of both the band and the textrodes as well as the protocol for the electrode signal comparisons and analysis. Section III describes the results of the signal comparison in the raw signal, percent accuracy and signal to noise ratio (SNR), Section IV discusses the significance of the singal equivalencies and Section V describes the future work to be done.

II. Methods

A. *Textile Fabrication*

The textile band was used as an interface with the EMG amplifier and the electrodes. The band was fabricated using an intarsia flat knitting machine SHIMA SEIKI SRY 12 gauge with multiple feeders of cotton and silver yarn 110 f 34 dtex HC+B. The electrical pathways were knitted together with cotton and sewn to an elastic fabric made with elastan. Snap buttons were used for interconnecting the textile conductive pads and the electrodes through the fabrics. A zipper was added at each end of the strap as the closing mechanism.

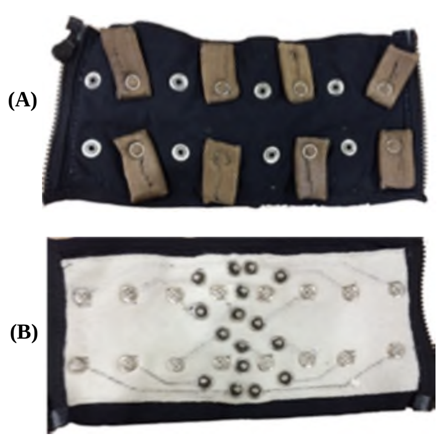

Figure 1. A. Textile Band (internal) with Textrodes attached. B. External Intarsia knitted cotton piece with snap connectors for wire connections to amplifier

The design of the band allows for a total of eight sEMG channels to be simultaneously recorded, however for the purpose of this study only four channels were used. For both gel and textile electrodes, 1 cm of space was allotted between the electrode pair used to form a differential input and 3 cm between each channel of the band. The band with textile electrodes is shown in Figure 1:

The textrode was fabricated with a conductive knitted Shieldex Technik-tex fabric sewn around a foam pad. A snap connector was attached to connect to the textrode to the band. The dimensions are 1x3 cm as shown in Figure 2.

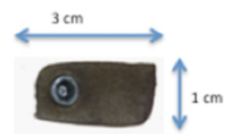

Figure 2.Foam padded textrode

B. *Protocol for Ag-AgCl and Textrode Comparison*

The upper arm movements of six healthy volunteers between 23-30 years old were recorded (3_male and 3 females). The band was placed on the dominant arm with the channels assigned as seen in the Figure 3 with channel 2 placed on the Extensor carpi ulnaris muscle. An AgCl electrode was used for reference in all measurments.

Two experimental measurement setups were used for the test, one with only AgCl Electrodes in the band then and another with only textrodes. The EMG activity was recorded with the subjects

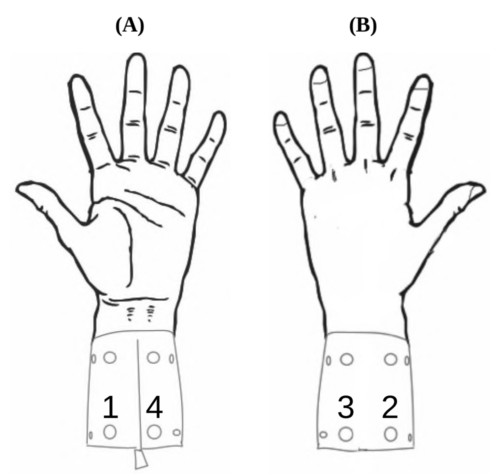

Figure 3. Band Placement with Channel Numbers. Channel 2 on Extensor Carpi Ulnaris

performing a ramp exercise for each type of contraction. The ramp exercise involves one MVC and then increasing contractions up to MVC guided by a computer program, namely BioPatRec [12]. The executed contractions included the following in order; open, close, flex, extend, pronate, and supinate. Three repetitions of each contraction were performed and the EMG superficial biopotential was acquired with a sampling frequency of 2000 Hz for a measurement time of 10 seconds and a duty cycle of 50%. *i.e., 5 second contraction time with a 5 second rest between contractions*.

The textrodes were wet with 2 ml of undistilled water to improve skin electrode interface [6]. Between tests the subject had a bandage placed on top of the band ensure the placement of the electrodes remain consistent while the AgCl electrodes were replaced with the textrodes.

C. *Signal Analysis:*

Signals were acquired and analyzed for offline accuracy in Pattern Recognition using BioPatRec software [12]. Four signal features for the pattern recognition were used including mean absolute value, wave length, zero crossing and slope change with a linear discriminant pattern recognition algorithm in a one-vs-one topology [12][13]. The sEMG recordings from the six hand movements were then processed. In order to calculate the percent classification accuracy a total of 206 time windows of 0.2 seconds were extracted per subject: 84 for training, 41 for validation and 83 for testing.

$$SNR_{db} = 10 * log_{10} \frac{S_{RMS}{}^2}{N_{RMS}{}^2} = 20 * log_{10} \frac{\sqrt{\frac{1}{n}\Sigma_1^n S_i^2}}{\sqrt{\frac{1}{n}\Sigma_1^n N_i^2}} \quad (1)$$

The SNR for the movements was then calculated using Equation 1 above where SRMS represents the RMS of the signal amplitude during movement and NRMS represents the RMS of the floor noise during rest.The strongest SNR signal of each of the four channels was used for each movement. Therefore to perform the SNR calculation equal levels of ground noise was found to be similar for all channels within the same test. The distribution and mean of the SNR for MVC was then found across all subjects.

Signal comparison was done from the sEMG recording obtained from the extension of the extensor carpi ulnaris on channel 2 for wrist extension of each subject, the strongest contraction signal, for both recordings performed with Ag-AgCl electrodes and textrode. For visual comparison the difference in SNR, the RMS and the power frequency spectrum were calculated and evaluated. Finally a Two One Sided T-test (TOST) [14][15], was performed to show that both electrodes had equivalent performance in accurately identifying the signal features for pattern recognition. The TOST was used to show that the mean difference between the two groups is less than the accepted error of 4.1%, which is the variation observed in classification accuracy with long term use of sEMG [16]. For $H_{01} = \mu_{text} - \mu_{AgCl} \geq 4.1$ and $H_{02} = \mu_{text} - \mu_{AgCl} \leq -4.1$.

The t value was calculated using Equation 2.

$$t = (\bar{x}_{text} - \bar{x}_{agcl} - \delta)/\sqrt{\frac{s_{Ag/AgCl}{}^2}{n} + \frac{s_{textile}{}^2}{n}} \quad (2)$$

Where $\bar{x}$ is the sample population mean, δ is the accepted error, *s* is the sample standard deviation and n is the sample size The H01 is rejected if -t ≤ -Critical T value, and H02 is rejected if t ≥ Critical T Value. The following section shows the results of the signal comparison.

III. Results

A. *sEMG Recorded*

A typical recording obtained with the sensorized strap is shown in Figures 4 and its power spectrum in Figure 5 filtered with a 50hz Notch filter.

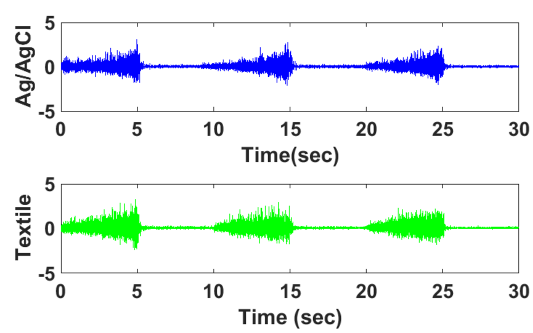

Figure 4 .Example of Ramp Recording obtained from Ch. 2 Extension in Extensor Capri Ulnaris

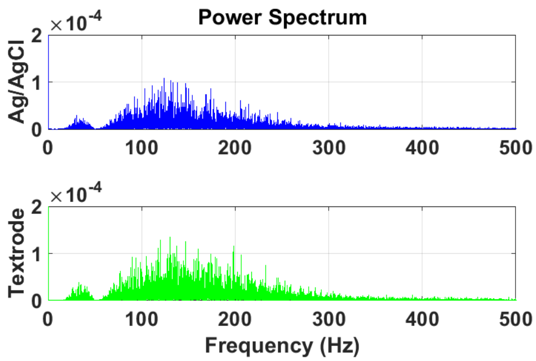

Figure 5. Power Density Spectrum Ag-AgCl and Textrodes.

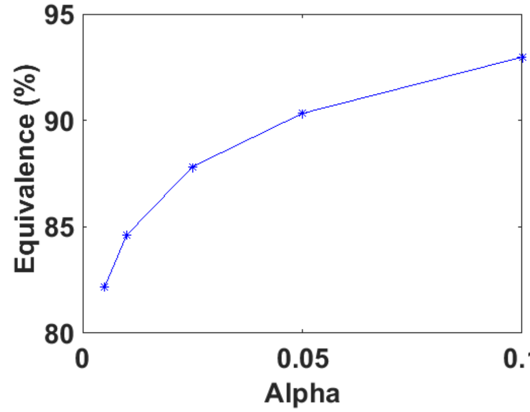

Figure 8.Percent Equivalence for each Confidence level

B. *Pattern Recognition Accuracy*

In Figure 6, the accuracy obtained with both type of electrodes for each of the six movements shows, in either case, values superior to 95%. The average across movements is 98.89% and 98.96% for the Ag-AgCl and the Textrodes respectively.

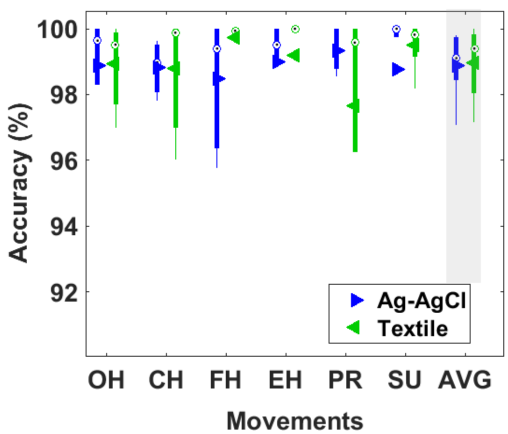

Figure 6. Movement Specific Pattern Recognition Accuracy for Six Core Movements across 6 subjects (Open Hand, Close Hand, Flex Hand, Extend Hand, Pronate, Supinate and Average).

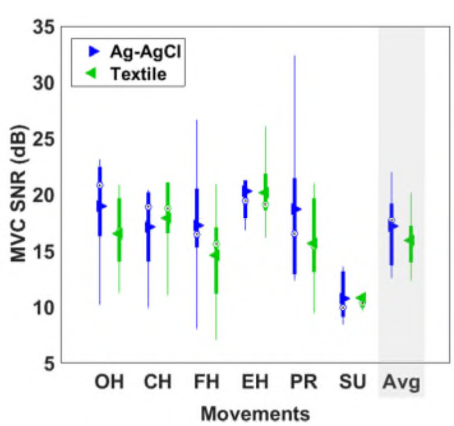

Figure 7. SNR during MVC for core six movements.

Table I indicates the values obtained for the t-statistic tests performed in the TOST procedure, indicating statistical equivalence.

TABLE I. T-VALUES FOR TOST

Margin Error	Movements					
	OH	**CH**	**FH**	**EH**	**SU**	**PR**
$\delta = 4.1$	-4.667	-5.201	-3.517	-3.907	-15.27	-2.783
$\delta = -4.1$	4.806	5.100	6.514	4.309	12.232	3.974

Note 1: $\alpha = 0.05$, δ =[-4.1,4.1], T-critical=1.8125 for n=6
Note 2: after outlier removal for SU, T-critical =1.833 and n=5

C. *Maximum Voluntary Contraction Signal to Noise Ratio*

Figure 7 shows the SNR obtained during the Maximum Voluntary contraction. The average difference in SNR across all the movements and all the subjects is 0.084%. The maximum difference in SNR between means across all six movements is 1.67%.

Figure 8 shows the average percent equivalence across six movements for which, the SNR obtained can be considered equivalent for increasing confidence levels. The percent equivalence ranges from 82.0% to 93.0% can be found for confidence levels ranging from 99.5% to 90% respectively. In the following section the significance of these equivalencies is discussed.

IV. Discussion

As seen in Figure 4, the two signals are visually similar in both in the amount of floor noise and the signal intensity during the contractions. This similarity is further supported by the power spectral density plots showing that the two electrode readings have similar frequency content. Additionally, the pattern recognition accuracy obtained with the classifiers in the offline simulation testing is above 95 in all cases as shown in Figure 6. Moreover, the average accuracy across all movements is very high and almost identical 98.89% and 98.96% for the textrodes and Ag-AgCl electrodes respectively. Regarding the observed accuracy, it can be seen that the recordings produce an statistical significant equivalent classification performance below 4.1% error, which is the average drop expected during the day in this kind of upper limb prosthesis [16]. Although it can be observed in Figure 7 that the movements show a range of performance in which the electrodes performance differs over two percent in accuracy such as in FH and SU the overall performance on average remains in the same range.

From the signal quality perspective, it is observed that the recordings produce a high equivalence at high confidence level with 93.0% equivalence at 90% confidence level.

The observed good performance and similarity between the signals of both the Ag-AgCl electrodes and the textrodes on the sensorized armband fabricated using intarsia-knitting techniques suggest that the use of intarsia as a method of forming electrical connections is valid. There is no evidence of cross talk between the signals on the band or noise added by the knitted pathways. The intarsia electrical pathways therefore seem to provide a reliable interface for sEMG recordings.

Due to the high level of accuracy of the textile electrode found in the offline pattern recognition system, their use in this sensing application would be able to provide a fully integrated textile solution to reduce the application time and skin irritation due to the long-term use of conventional gel Ag-AgCl electrodes. The lack of chemical agents from the adhesive and hydrogel layer and avoidance of irritation of the skin upon removal of the electrode are a certain advantage to increase patient comfort by avoiding skin irritation [3].The textrodes would be useful for muscular therapies for amputees such as treatment for phantom limb pain and strengthening of the residual muscle as well as myoelectric prosthetics, which require wearing electrodes for extended periods of time.

V. Conclusions

The preliminary results from this study suggest both that the intarsia technique for knitting in electrical pathways and the knitted textile electrodes could provide a quality interface for sEMG monitoring of upper arm movements.

Further research must be done in finding the effect of washing and wearing of the textrodes and the sensorized intarsia knitted band on the classification performance and signal quality of the recordings. Furthermore live recordings of transradial amputees for movement prediction must be performed in the next steps.

The viability of using textile electrodes and the intarsia knitted electrical pathways on healthy subjects with offline pattern recognition using a fully integrated intarsia-based textile system has been successfully evaluated. Given the current penetration of intarsia knitting in the textile industry, we might have found the way towards true volume manufacturing of seamless integrated textile-electronic sensorized garments.

Acknowledgements

The authors would like to express their gratitude to SHIMA SEIKI EUROPE LTD and Syverket AB for their support to this project.

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Performance Evaluation of DSDV, DSR, AODV and TORA MANET Routing Protocols for Body Monitoring in Free Space Environments

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***Abstract*— A Mobile Ad hoc Network (MANET) is a constantly self-configuring and no-centralized stationary infrastructure network, in MANET, the mobile nodes connected via wireless channels. Routing protocols considered as important elements to forward data in such a dynamic network topology. They are needed when it is required to gather information from people like in health monitoring. This paper firstly introduces a concise overview about the MANET. Then, it shows the study of four routing protocols in terms of their capability scaling with the network growth and intensity of nodes, in addition to examine how they behave in two mobility models (regular cases of patients monitoring). It presents dissections and assessment of each protocol performance.**

***Keywords*— MANET; DSDV; DSR; AODV; TORA; Random mobility model; Grid mobility model.**

I. INTRODUCTION

Nowadays, there has been a continuous evolution in the technology of mobile devices, such as smart mobile phones, Personal Digital Assistants (PDAs), mobile sensors, and vehicles. This technology comes with a huge advantage, such as flexibility of using the mobile device anytime and anywhere despite of time and location constraints. On the other hand, the mobility feature of these devises lead to limitations in their resources in terms of memories, hard drives and the volume of batteries [1]. Although of these limitations, there are new applications came up rapidly to take benefit of mobile devices. This technology is mainly being used in personal and business, for example games, maps navigation, entertainment, banking/finance, and shopping. The considerable revolution of these mobile devices is the capability of connecting to networks, nevertheless, because of their limitations they require and efficient routing protocols. These mobile devices or nodes connect with each other's by a mobile ad-hoc network (MANET), which is a network of wireless stations without centralized or fixed infrastructure. These networks encounter periodically changes in its topologies. Each node in this network performs as peer router with other, all nodes represents routing stations and they collaborate to transfers data. Because of the absence of the base station and dynamic changing of topology, the process of forwarding data among these nodes depends on the efficient performance of routing protocols [2][3].

MANET routing protocols can classified as reactive, proactive or hybrid protocols. The reactive protocol does not preserve any routing information or state of the network until it get request of transmission, at that moment, lookup for a route and establish connection, for the reason that it called on-demand protocol. On the other hand, the proactive calculate and maintain all routes previously, and make periodic update of its routing tables. Finally, the hybrid protocols came to take advantages of the proactive and reactive algorithms, initially it set up routing proactively, and when there are newly joining nodes it deal with them through reactive flooding [4].

MANET routing protocols used with several applications, such as house monitoring (smart home), habitat monitoring, which it help ecologists to gather environmental data that affecting people, animals, and plants, in addition to healthcare applications, that provide healthcare anywhere and in any time. This work concentrates to study MANET routing protocols, for facilitating people monitoring, which is an important factor in healthcare applications. Ad hoc network routing protocols are required to send data to patients and receive information from them. People mobiles in different patterns and the density of them changed from location to another, for example, in healthcare monitoring, the number of monitored people increased in and near to the hospital and health centers, as a result, it is important to know the behavior of each MANET routing protocol to built the proper healthcare systems.

This paper performs a performance evaluation of the following four mobile ad hoc network protocols, destination-Sequenced Distance-Vector Routing Protocol (DSDV), Dynamic Source Routing (DSR), Ad-hoc On-demand Distance Vector routing (AODV) and Temporally Ordered Routing Algorithm (TORA), for body monitoring in free space. The approach of assessment includes two scenarios; the first one is used to examine the behavior of each protocol according to the density and scaling in numbers of nodes, the second one is used to study how these routing protocols will react in two different mobility models.

Rest of this paper organized as follows. Section 2 gives a brief description about MANET routing protocols under study. Related work is addressed in Section 3. Section 4 shows the scenarios and simulations environment. While Section 5 presents the results and discussions of them. Finally, Section 6 concludes this paper and gives a glance about future work.

II. MANET ROUTING PROTOCOLS UNDER STUDY

A. *Destination-Sequenced Distance-Vector Routing Protocol (DSDV)*

One of Ad-hoc proactive protocols, it is a table-driven routing scheme; all node of the network maintains a routing table contains of all destinations and the number of hops that a packet would need to reach to the destination. The DSDV solves the routing loop problem (loop paths), by providing a sequence number for each entry in routing table. This algorithm send a periodic updates of entire routing table, between these periods it send smaller routing updates between nodes frequently [3][5].

B. Dynamic Source Routing (DSR)

DSR one of reactive routing protocols. In this protocol, the source node overflows the entire network with a route discovery request, it identify each route request with the packet source and destination of this route. The target node (destination) respond to the route request, by scans its own cache for a route before sending route reply to the initiator of route request. If there no route found, the destination execute its own route discovery mechanism in order to reach to the source [6].

C. Ad hoc On Demand Distance Vector (AODV)

One of Ad-hoc On-demand a routing protocol, it builds routes between nodes only when it required by source nodes, and does not allow the nodes that not founded in the active path to preserve information about this route. AODV develop trees to connect multicast group members. Moreover, it uses the sequence numbers to guarantee the free route loop problem. AODV builds routes using a route request and route reply query cycle, when a source node desires a route to a destination; it distribute a route request (RREQ) packet across the network. Nodes receiving this packet update their information for the source node and set up backwards pointers to the source node in the route tables [3] [5].

D. Temporally Ordered Routing Algorithm (TORA)

TORA is hybrid routing protocol; the paths establish with proactive routes initially, and then servers request additional routes through reactive flooding. This protocol has high energy consumption in contrast with other protocols of its type [3]. The main objective of TORA is to limit control message propagation in the highly dynamic mobile computing environment. Each node explicitly initiates a query when it needs to send data to a particular destination [7].

III. Related Works

R. Lacuesta et al. presented a comprehensive investigation in ad hoc networks routing protocols [3]. Their chapter deliberates the protocols in term of secure and non-secure routing protocols, addition to analyze protocols performance with different metrics. They use OPNET (Optimized Network Engineering Tool) simulator to emulate the DSR, OLSR (Optimized Link State Routing Protocol), and AODV protocols, the simulated topologies contain 50, 100, and 250 nodes, moreover, there are two scenarios, one with movable nodes with failures and the other consists of fixed node. Their results present the OLSR and AODV as superior in performance than DSR; however, OLSR is worst protocol in the performance of data link layer.

A comparative analysis of four MANET routing protocol in different mobility models of ad hoc network was introduced by N. S. Samshette and others [8]. They simulated three mobility models, Random Walk, Random waypoint, Random direction. The routing protocols under test are DSR, DSDV, AODV, and AOMDV (Ad-hoc On-Demand Multipath Distance Vector routing). Their results had shown that the AODV as best throughput in Random Walk mobility model, and DSR smallest throughput for all mobility models. However, they use only TCP traffic and 25 nodes.

Rahman et al intend to evaluate the performance of three routing protocols by using The Network Simulator (NS2) [5]. Their study includes AODV, DSDV, and Improvement of DSDV (I-DSDV) protocols. The number of simulated nodes set in 5, 10, 15, 20, 25, 30, and 35 nodes. Their results demonstrate that the AODV has better performance than other, while I-DSDV reduces the number of dropped data packets compared DSDV, however it produces more computation overhead.

In 2008, J. Lloret et al. try to take advantage by designing a group-based ad-hoc network in order to gain flexibly and efficiently. In addition, splitting the network in groups, it also improves the fault tolerance and gains more other benefits. They studied the DSR, OLSR, and AODV MANET routing protocols behavior when these protocols work in groups, compared to an individually-based network [9]. In this case, OPNET simulator was used to create the suggested topologies. The results show that grouping nodes in ad-hoc network lead to achieving better performance, in addition to decrease the extra and overhead traffic of network. OLSR appeared as the best protocol in group-based topologies [9]. In the same year, Garcia and others studied the three MANET protocols mentioned above. In this case, they examined the performance of protocols in Wireless Sensor and Actor Networks (WSAN). The experiment proves the possibility of implementing these protocols in WSAN. OLSR appeared as most suitable protocol for WSAN than AODV and DSR, however OLSR was the worst in terms of consumed throughput rate.

N. Meghanathan investigated the performance of stability-oriented MANET routing protocols [11]. His study included the Associativity-Based Routing (ABR) protocol, Flow-Oriented Routing (FORP) Protocol and Route Assessment Based Routing (RABR) protocol. This findings show that FORP has the best routes stability than others, however, this demand higher end-to-end delay and higher energy consumption per packet.

Moreover, M. Tarique et al suggested studying the MANETs network according to other performance metrics rather than network throughput [12]. They examined packet delay, overhead control packets, energy consumption, connectivity, and shadowing effects. Their study was performed under short-hop and long-hop routing by using DSR routing protocol, because, it is important for routing protocol is to decide whether a mobile node should use many short-hops or a few long-hops. The results show that the long-hop routing is better in term of reducing the delay per packet and packet loss, while short-hop routing is better in keeping network life as long as possible.

In field of people monitoring, M. Garcia and others propose soccer team players' remote monitoring system through using wireless sensor network [13]. Their procedure intended to use the Wireless Body Area Networks (WBANs) to know the physical state of players during matches. They studied the network topology and mobility models of the soccer players. ZigBee Routing Protocol (ZRP) was used for the communication between nodes. The routing protocol was AODV. The results clarify that the additional routing hops is required in case of high mobility to achieve lower network load, however, the management traffic is low in their proposal. The same authors proposed a wireless body sensors network for soccer team in [14], which it is a continuous work of the previous study. The routing of information can be done via the nodes of the same team or through the nodes of the other team; therefore, they added a security system to let the information be decrypted only by players and coach of the same team. In addition, they narrow radio coverage area to be no longer than four meter to avoid eavesdropping from any place of the stands. Furthermore, they improved the mobility model and increased the number of simulations. Additionally, they proposed an energy harvesting system to provide enough energy for that very low transmission range. Simulations results show that the management traffic is low, which in all cases of high and low mobility is below than 250 Kbps.

IV. SCENARIOS AND SIMULATIONS

Studying routing protocols and identify their behavior, is important to forward data, and to gather information from people in health monitoring. This part gives brief descriptions about the methods followed to compare the four mentioned MANET protocols. The approach includes two fundamental scenarios. The first one is used to study the behavior of each protocol according to the density and scaling in numbers of nodes in three different cases. The initial case contains 50 nodes, which it has six nodes send traffic by TCP and UDP transport layer protocols. While case two created by hundred nodes, besides, to raising the packet size and number of sending nodes, to send more flow in network. The last topology has two hundred nodes as well as increasing in packet size. All topologies have same set of some nodes moving around in different times, and some others are static for whole time of simulation. In each case study, the four protocols have been tested in the same environment and all the parameter is same except the type of protocols. The second one is used to examine how these routing protocols will react in different mobility models, this study analyses two models, Random Walk and Manhattans Grid Mobility (MGM). In random walk mobility model, nodes moves randomly by any speed or direction [8], such as movement of users in public parks and playgrounds or large hypermarkets. While the grid represents transitions of mobile nodes in streets, every node permitted to move horizontal or vertical. Further, the two topologies of the second scenario are set up with same parameters and size of network. Figures 1 and 2 show these topologies configured in the simulation. The connections inside the figures are just to demonstrate the intersections between nodes, addition to sources and destinations of network flow.

In order to measure the performance of the routing protocols in each scenario, data are collected from the simulation. Then throughput and normalized routing overhead metrics are chosen for the performance assessment. The throughput in communication networks is the average of successful message delivery over a communication channel and it is measured by the number of bits delivered per second or data packets per time unit. While normalized routing overhead (also called normalized routing load) is the total number of routing packet transmitted per data packet, it examines the cost of routing vs. success receiving of application data.

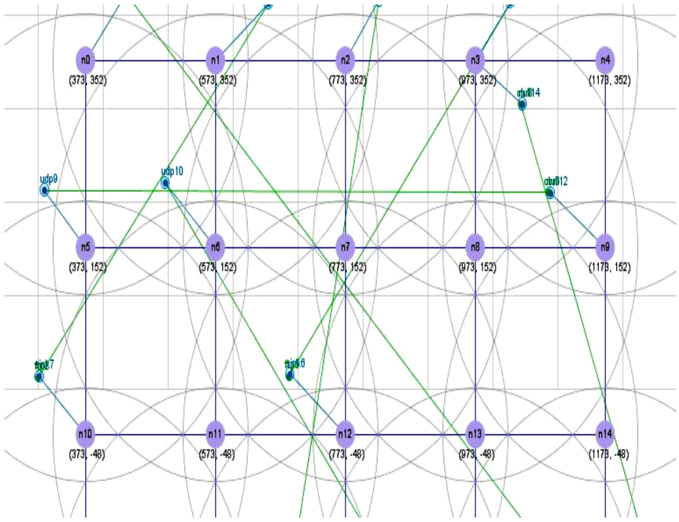

Figure. 1. The Grid Mobility Model

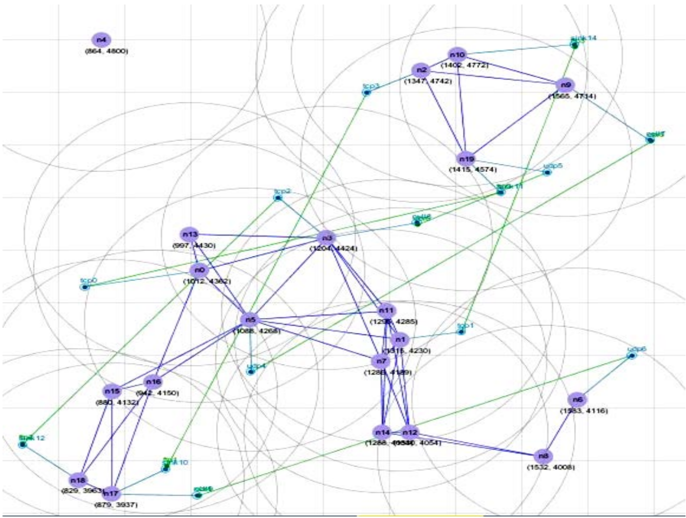

Figure. 2. The Random Mobility Model

A. *Simulation environment and flow parameters*

The experiments have been performed using NS2 simulator, NS2 is a discrete event open source simulator [15]. It supports different types of wired and wireless network protocols. In addition it has been used in many networking researches. NS2 use TCL scripting language to set and draw the network topology, NS2 scenarios generator 2 (NSG2) is a Java based tool used to create TCL programs of examined scenarios. NSG2 reduces the time of writing TCL scripts by using an easy GUI interface [16].Table 1 shows the parameters applied in the simulation tests. Awk scripting language is used to calculate the throughput and routing load. The graphs of the data gathered have been created using Microsoft Excel.

TABLE I. SIMULATION PRAMATERS

Prompters	Topology _50	Topology _100	Topology _200	Random and Grid
Number of nodes	50	100	200	200
TCP packet size/bytes	1500	3000	4000	2000
UDP packet size/bytes	1000	1500	2000	2000
Application level data	FTP and CBR	FTP and CBR	FTP and CBR	FTP and CBR
Simulation time /s	0-15 s	0-15 s	0-15 s	0-10s
Others	Others parameters such as channel type, MAC protocol, etc are same for all topologies			

V. RESULTS AND DISSECTION

This section explains and discusses the results gathered from simulation in term of throughput and routing overhead for the two scenarios stated above.

A. *First Scenario Dissections*

Figure 30 shows the throughput of the protocols when the number of nodes is 50. In this graph the DSR protocol obtains very good throughput in comparison with other protocols. It has average throughput of 761.0 kbits/s. While DSDV takes a lot of time in control before it began to deliver applications data and get 130.3 kbits/s as average. As observed in Figure 3, AODV has an average throughput of 434.8 kbits/s, which places it between the others. In this case, TORA is the worst one because its throughput is zero; furthermore, during the simulation it is just sending information to discover neighbors. TORA behave as same with scaling number of nodes and the data traffic dropped in the intermediates nodes.

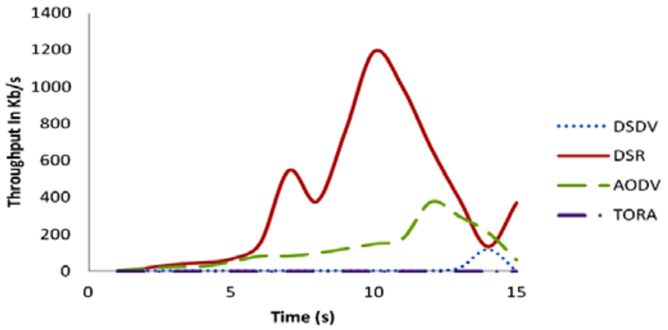

Figure. 3. Throughput in 50th nodes topology.

The throughput given when there are 100 nodes is illustrated in Figure 4. In this situation the flow of application data also increased. As shown in the figure, AODV has a good throughput with the expansion of the network than others protocols. It has 369.8 kbits/s of average throughput. In spite of the growth of the network, DSR is still well behaved which has an average throughput of 351.2 kbits/s. While DSDV witness some reduction in its average throughput and has 99.0 kbits/s. TORA has the worst average throughput.

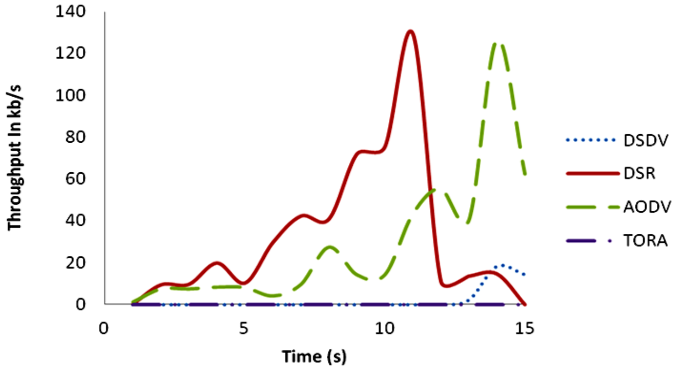

Figure. 4. Throughput in 100-nodes topology.

As noticed in Figure 5, the throughput is degraded for all protocols with the network growth. In this case, TORA situation not changed at all. In contrast, DSDV had some improvements compared with previous results with an average throughput of 118.5 kbits/s, while DSR and AODV has similar result, their throughput is 355.2 and 235.0 kbits/s in average respectively.

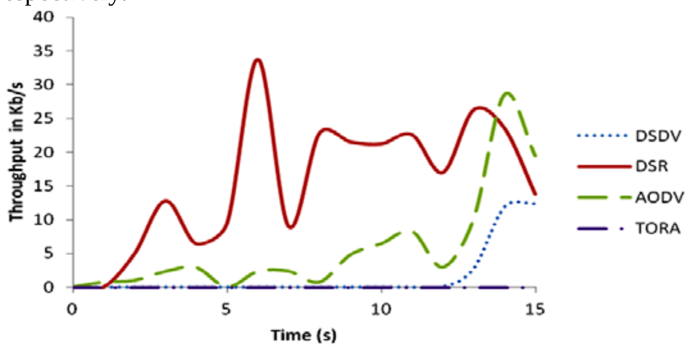

Figure. 5. Throughput in 200-nodes topology.

Figure 6 shows the normalized routing load for the case of 100 nodes. We can observe in this graph that TORA has higher routing load than the others when it starts forwarding routing packets and remains that the highest one until it finishes. Moreover, DSDV achieves high routing overhead in the beginning of the simulation time, after that, it acquires few overhead. DSR and AODV have similar moderate routing overhead.

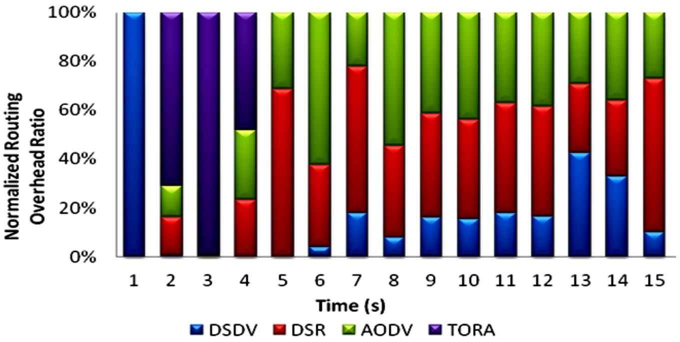

Figure. 6. Normalized Routing Overhead in 100-nodes topology.

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B. Random and Grid Mobility model Dissections:

The next graphs explain the throughput according to the two mobility models. Figures 7 and 8, show that DSR protocol performs better than other protocols in both models, and it has an average throughput of 681.0 kbits/s in random model and 615.5 kbits/s in Grid model. AODV's throughput is better in Random (where average throughput is 668.3 kbits/s) than Grid, which has 598.4 kbits/s. DSDV and TORA are the worst protocols, DSDV is bad one with average throughput because the number of nodes is small. As shown earlier, it achieves better performance with the increase of nodes, but this does not guarantee it in case of massive intensity of nodes. TORA protocol gains better throughput in Grid than Random model. However, it very trivial to be represented in the graphs with other protocols, actually it works bad even with small number of nodes. It forwards negotiable bytes in contrast of others. It has from 0.00 to 0.28 kbits/s in Random, and 0 to 32.6 kbits/s in Grid.

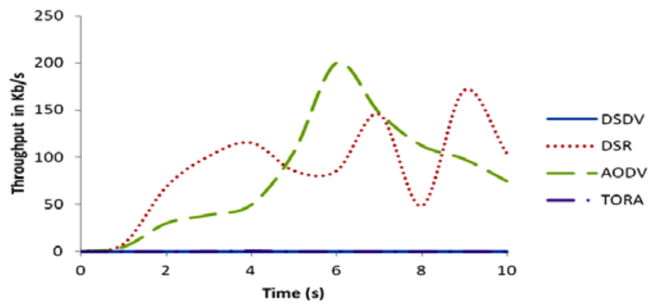

Figure. 7. Random Mobility model.

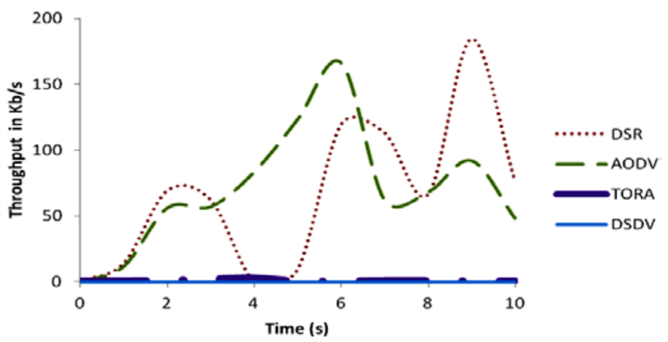

Figure. 8. Grid Mobility model.

Finally, Figure 9 shows the evaluation of normalized routing load for the four routing protocols in the grid topology. Routing overhead in random model is studied in the first scenario. The mobility model is the same although there is different number of nodes. The normalized routing load represented in Figure 9 confirmed what declared before: DSR and AODV have lowest and moderate load, while DSDV has very huge overhead when it starts, after that it preserves small load. Lastly, TORA has the vast normalized routing overhead.

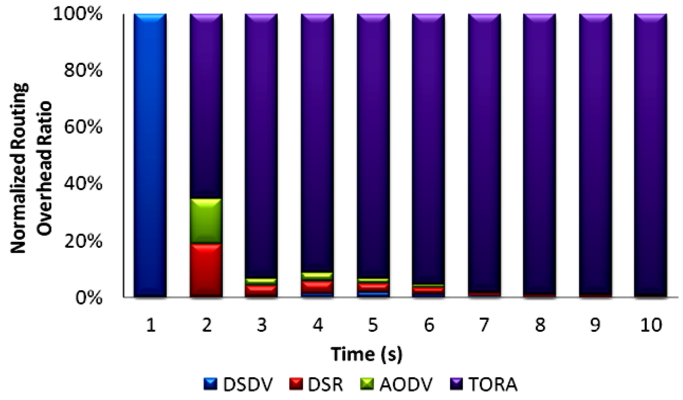

Figure. 9. Normalized Routing load in Grid Mobility model.

VI. CONCLUSIONS

The behavior of MANET routing protocols in different mobility models, and their scalability have been an issue of concern in many applications areas. This increases with the type of application area, such as people monitoring in healthcare systems. This paper studies DSDV, DSR, AODV, and TORA protocols, with different number of nodes and packets size. In addition, their performance in two mobility models is assessed. As summary, DSR obtains better throughput in comparison with others, while AODV achieves higher throughput with the growth and intensity of the network. TORA protocol is the worst one in terms of routing overhead and throughput.

Finally, in our future work, these four MANET routing protocols will be analyzed by moving all nodes around in order to show which one has better performance. Furthermore, we will continue this work, by adding Optimized Link State Routing Protocol (OLSR) [17], as well as we will study other mobility models. .

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Wearable Technologies for Multiple Sclerosis

The future role of wearable stress measurement in improving quality of life

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***Abstract*— Multiple sclerosis (MS) is the most common autoimmune disorder affecting the central nervous system. Most often diagnosed in young adults, MS runs a chronic, unpredictable course, often leading to severe disability: 50% of MS patients are unable to perform household and employment responsibilities 10 years after disease onset, and 50% are nonambulatory 25 years after disease onset. While it is not clear what factors influence the prognosis of MS, exposure to stress has long been suspected as a factor that can aggravate its progression. In this paper, we discuss the opportunities for wearable sensors in the management of stress in multiple sclerosis patients.**

Keywords–multiple sclerosis; stress; wearable; biosensor; electrodermal activity; heart rate variability.

I. INTRODUCTION

Multiple sclerosis (MS) is a chronic inflammatory, demyelinating and neurodegenerative autoimmune disease of the human central nervous system, affecting more than 2.3 million people worldwide. The median clinical onset of MS is approximately 29 years of age, with affected women outnumbering men with a ratio of almost 3:1 [1].

The clinical disease course is variable. In most cases, MS begins with a relapsing-remitting (RRMS) course characterized by periods of exacerbation, lasting from days to weeks, followed by periods of substantial remission, often with some residual disability. In a later stage, anytime between 5-35 years after onset, most patients develop a secondary progressive course (SPMS), marked by a continuous and irreversible neurological decline. A small proportion of patients have a primary progressive course (PPMS) from the onset.

Common symptoms of MS include, but are not limited to, loss of function or feeling in limbs, loss of bowel or bladder control, sexual dysfunction, debilitating fatigue, blindness due to optic neuritis, loss of balance, pain, cognitive dysfunction, and emotional changes [2].

While little is known about its cause or the factors that contribute to its unpredictable course, evidence indicates that both genetic susceptibility and environmental factors play a role [3]. One such factor is psychological stress, which has been implicated repeatedly as a determinant of disease activity ever since MS was first described in the 19th century.

In this paper, we discuss the impact of stress on MS progression and the opportunities for wearable biosensors and and autonomic activity measurement in the management of stress in MS, with the subsequent improvement in quality of life.

II. THE ROLE OF STRESS IN MULTIPLE SCLEROSIS

Stress is a broad concept used to describe conditions ranging from environmental threats to psychological responses relevant to anxiety. Despite the great variability in how people perceive and experience stress, both patients and clinicians note that increased stress increases relapses. In fact, the most recent meta-analysis of studies linking stress and disease exacerbations found that out of the 17 studies reviewed, 15 showed a significant association [4]. Furthermore, acute stressful events or chronic stressful situations have also been associated with the onset of the disease [4]

While relapses are the primary measure in these studies, the count of new lesions on magnetic resonance imaging (MRI) has proven to be a far more sensitive outcome as a measure of disease activity. Although only two studies have studied the relationship of stress and MRI lesion activity, both found that stress increases disease activity on MRI. In one study, the risk development of new lesions 4–9 weeks after a major negative life event was increased, while positive stressful events reduced this risk [5]. A previous smaller study had a similar finding [6].

These findings are not surprising, since neuroendocrine hormones triggered during stress may lead to immune dysregulation or to altered or amplified cytokine production, resulting in atopic autoimmune activity or decreased host defense [7].

III. WEARABLE STRESS MEASUREMENT

Researchers have studied a wide variety of approaches to measure stress, such as self-report measures, collected through retrospective surveys and/or experience sampling, and hormone analysis. However, stress also produces a well-studied set of physiological changes that can be continuously monitored using wearables for the purpose of stress measurement. These physiological changes are controlled by the autonomic nervous system (ANS), which regulates important bodily functions including digestion, thermoregulation, cardiac output, regional blood flow, ventilation, and many aspects of emotional behavior: feelings of fear, anger, happiness, and sadness have characteristic

 ISBN: 978-1-61208-480-0

autonomic manifestations. The ANS is divided into the sympathetic nervous system (SNS) and the parasympathetic nervous system (PNS). While SNS mobilizes the body's resources in response to a challenge or a threat (e.g., quickens the pulse, deepens the respiration and tenses the muscles), the PNS works antagonistically to control this process.

In the past, to accurately gather measurements of autonomic responses in the midst of daily activity, cumbersome electronics such as sticky electrodes on the chest were usually required. Therefore, so far studies of stress in MS that have looked beyond self-report surveys have relied on single time-point measurements of autonomic responses to a standardized stressor, such as public speaking or a cognitive task [8]. However, the most recent commercially available wearable biosensors can accurately and unobtrusively measure autonomic responses through electrodermal activity and heart rate variability data collected from the wrist:

A. *Electrodermal activity*

Electrodermal activity (EDA) biosensors measure electrical conductance changes in the skin reflecting eccrine sweat-gland activity. Unlike other bodily functions, EDA is controlled exclusively by the SNS, making it an ideal physiological signal for stress measurement. Using EDA, stress can be distinguished from other similar responses such as cognitive load. As an example of an EDA application, Hernandez et al. discriminated stressful and non-stressful calls at the call center environment using EDA features [9].

B. *Heart rate variability analysis*

Heart rate variability (HRV) analysis describes the variability of heart rate (HR) over time. In general, HRV analysis takes into consideration the frequency power of low-frequency (LF, 0.01−0.08 Hz) and high-frequency (HF, 0.15−0.5 Hz) bands, which reflects SNS and PNS modulation of HR respectively. HRV can be performed from blood volume pulse (BVP) signals obtained from wrist photoplethysmograph (PPG) sensors and, similar to EDA, used as a biomarker for psychological stress.

IV. Opportunities for Wearable-based Stress Management in MS

Although the studies examining interventions to reduce stress in people with MS are few, it has been shown that stress management programs can result in a significant reduction of both MS symptoms [10] and new brain lesions [11]. While these programs offer great promise, their effect disappear rather quickly after the therapy is stopped. Therefore, there is a need for alternative ways of delivering stress management therapy that can result in longer-lasting effects.

Recent advancements in wearable stress measurement technologies offer an opportunity for the development of biofeedback-assisted stress management therapies that could potentially satisfy this need, resulting in a significant improvement in the quality of life of multiple sclerosis patients. Specifically, we believe the more scalable, adaptive, customizable and interactive behavior interventions that can be achieved with wearable-based behavior intervention technologies tailored to the specific needs of MS patients may result in increased patient engagement, improved stress reduction, and longer-lasting effects.

Acknowledgments

This work has been supported by La Caixa Foundation.

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