iCalm: Wearable Sensor and Network Architecture for Wirelessly Communicating and Logging Autonomic Activity

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***Abstract*—Widespread use of affective sensing in healthcare applications has been limited due to several practical factors, such as lack of comfortable wearable sensors, lack of wireless standards, and lack of low-power affordable hardware. In this paper, we present a new low-cost, low-power wireless sensor platform implemented using the IEEE 802.15.4 wireless standard, and describe the design of compact wearable sensors for long-term measurement of electrodermal activity, temperature, motor activity, and photoplethysmography. We also illustrate the use of this new technology for continuous long-term monitoring of autonomic nervous system and motion data from active infants, children, and adults. We describe several new applications enabled by this system, discuss two specific wearable designs for the wrist and foot, and present sample data.**

***Index Terms*—Affective computing, anxiety disorders, autism, autonomic nervous system (ANS), electrodermal activity (EDA), fabric electrodes, heart rate variability (HRV), network, radio, sleep, wearable sensors.**

I. Introduction

AFFECTIVE computing has a growing number of applications in healthcare, motivated by over a decade of findings from neuroscience, psychology, cognitive science, and the arts about how emotion influences human health, decision making, and behavior. While there is still no widely accepted definition of emotion, many scientists agree that its main dimensions can be described as arousal (calm or excited) and valence (negative or positive) [1]. The ability to measure changes in arousal and valence accurately, comfortably, and continuously, without injecting cumbersome wires or boxes into people's activities, has the potential to revolutionize health therapies and services, especially through advancing personalized therapies, where a treatment or service can adapt to the patient's individual affective state and state-influenced needs. For example, a majority of relapses in smoking are linked to stress, and stress management is a strong contributing factor of successful cessation [2], [3]. A wearable, autonomic device that is unobtrusive, low cost, and comfortable enough for continuous wear, might help the wearer who wants to quit smoking to see whether and when stress is affecting his or her behavior—perhaps influencing use of tobacco or drugs, overeating, or other health problems. Help might then be customized for better managing that affective state and its causes, perhaps even at a moment well-timed to the triggering event [4]. There are also many potential uses of affect sensing in developing physiological and behavioral measures to classify emotional states associated with preclinical symptoms of psychosis, mood, anxiety, and personality disorders, as well as in monitoring physiological and behavioral reactions to tailor medications to an individual [5].

A challenging application area of interest to our research, which informed the design of the system described in this paper, is the communication and characterization of emotion in autism. People diagnosed with autism spectrum disorders (ASD), especially those who are nonspeaking, are often described as having unexpected "meltdowns," where they appear perfectly calm and yet suddenly become disrupted and may engage in behavior that is self-injurious or injurious to others [6]. Measurements have shown instances, where an autistic individual can appear outwardly calm, while having an internal state of extremely high-autonomic arousal [7], [8]. There is reason to believe that such "unseen stress" may be broadly true in ASD [9], especially, where a person may be unable to speak or otherwise communicate feelings accurately. We would like to create technologies that these individuals can use to more accurately express their internal state to people they trust.

Arousal is a dimension of emotion that occurs when there is activation in the autonomic nervous system (ANS), which has two main branches: sympathetic and parasympathetic. Generally speaking, the sympathetic nervous system (SNS) dominates in emergency conditions and initiates widespread and profound body changes, including acceleration in heart rate (HR), increased electrodermal activity (EDA), dilation of the bronchioles, discharge of adrenaline, inhibition of digestion,

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and elevation in blood pressure. The parasympathetic nervous system (PNS) contains chiefly cholinergic fibers that tend to induce secretion, increase the tone and contractility of smooth muscles, and slow HR. The SNS and PNS work together to maintain homeostasis: a dynamic equilibrium in which continuous changes occur, yet relatively uniform conditions prevail [10].

In this paper, we describe the design, construction, and evaluation of interactive continuous autonomic logging and monitoring (iCalm), a new device that is reliable, low power, low cost, and comfortable enough to wear around the clock by adults, children, and infants for logging and communicating personal autonomic data.

II. Engineering and Design Challenges

Over the past few years, several commercial sensor systems have begun to emerge in the sports, fitness, and home healthcare markets that are comfortable to wear for short periods of time and capable of wirelessly transmitting autonomic data to a nearby computer. For example, the Polar and FitSense HR monitors transmit and log average HR and activity using a chest-worn strap and pedometer information [11], [12]. While these systems are relatively low cost and comfortable to wear compared to the bulky A–D converters used by psychophysiology researchers, they do not capture EDA (sometimes called galvanic skin response), a signal of particular interest in monitoring SNS activation since the skin is the only organ purely enervated by the SNS [13]. The BodyMedia armband measures EDA, motion and thermal information, and wirelessly transmits this data to a wristwatch [14]. However, these commercial systems are still relatively large, do not support customization or multiple sensor nodes, and they employ proprietary software and protocols, making them impractical for widespread use in affective computing and medical research. We have also found that long term (weeks) of continuous wear using rubberized electrodes is uncomfortable, as is long-term use of standard metal medical electrodes and the adhesive pads used to apply them: both of these have caused us skin irritation when the skin does not breathe for many days of use. Other wireless, wearable recording systems developed for research, e.g., MITes accelerometers [15], and the Handwave skin conductance sensor [16] are specialized to sense and transmit only one physiological parameter. Commercial general-purpose systems, such as FlexComp [17], Brainquiry [18], or Vitaport [19] enable high-quality recordings and are nicely customizable; however, they have cumbersome form factors and cost thousands of dollars each, limiting their use outside the laboratory for large-scale and long-term (24/7) studies. Vivometrics' LifeShirt is perhaps the most comfortable, flexible, and ambulatory ANS monitoring system available; however, its high price ($10 000–$15 000) makes researchers reluctant to let participants take it home and puts it out of the price range of most long-term multisubject studies.

In addition to performance, form factor, and cost, another important concern with existing systems is battery life. In chronic conditions (e.g., autism, sleep disorders, epilepsy, posttraumatic stress disorder, bipolar disorder, etc.), there is a need to collect physiological data continuously over weeks and months. Given a typical coin cell battery with a capacity of a few hundred milliampere hours, this requires that the average power consumption of the wearable system be less than 1 mW. This level of power consumption cannot be achieved by the radio hardware design alone; it also requires proper design of the sensing hardware and controller firmware.

In the remainder of this paper, we present the design of a compact, comfortable, low-cost, low-power wireless wearable system for autonomic sensing and communication that is optimized for outpatient and long-term research studies. Section III presents the design and operation of the sensor hardware. In Section IV, we discuss the wireless hardware and network architecture. Section V describes the form factor and software interface. In Section VI, we illustrate the sensor data.

III. Sensor Hardware

A. Design Objectives

Our primary objective was to design a low-cost, comfortable, and robust sensor module that provided the necessary set of measurements needed for affective sensing. In addition, the sensor hardware needed to be small and low power.

B. Choice of Sensors

For sensing autonomic changes due to the SNS, we chose EDA, measured as small changes in conductance across the surface of the skin. For sensing changes due to both the PNS and SNS, we measure peaks of photoplethysmograph (PPG) signals, also known as blood volume pulse (BVP), and compute features of heart rate variability (HRV). Details of the PPG and EDA circuits can be found elsewhere [20].

Because motion and environmental temperature can influence a person's electrodermal and cardiovascular signals, a low-power temperature sensor and motion sensor were also included. For temperature measurement, we used the National Semiconductor LM60 sensor IC, and for motion sensing, we used an analog motion sensor (Signalquest SQ-SEN-200) with an integrator circuit. The latter sensor was used instead of a three-axis accelerometer because it draws less than 1 μA of current and costs only US$1.50. A version of the sensor band with three-axis accelerometer is also available for more precise motion data at the expense of increased cost and power.

C. Circuit Design

Our sensor board implements an exosomatic measurement of EDA, such that a small voltage is applied to the skin and the resulting potential drop is measured. The primary technical challenge in creating this circuit was to achieve a low-power design while still maintaining good dynamic range. It is well known that baseline skin resistance can vary over a few orders of magnitude from 100 kΩ to approximately 10 MΩ; yet, it is necessary to detect minute changes in this value, which is somewhat challenging with low-voltage power rails (2.5 V).

In order to preserve good precision over a wide dynamic range, an automatic gain control circuit was implemented using two op-amps with nonlinear feedback. Using an op-amps with low-leakage current (such as the AD8606), it was possible to achieve a measurement circuit with sufficient dynamic range and low-power consumption ($<$1 mA at 2.5 V).

In order to maximize battery life and maintain a stable voltage rail for the op-amps and sensors, a low-power low-noise regulator was added (LM1962). This regulator has a power enable pin to turn-OFF the power to the entire sensor module in between sensor readings, thus reducing the power consumption of the entire sensor module to less than 20 μW and enabling several days of continuous use on a single charge.

For measuring HR and HRV information, a special version of the sensor board was constructed, which included an optional PPG circuit for measuring blood volume pulse. The PPG circuit consisted of a Honeywell SEP8706-003 800 nm LED and an advanced Photonix PDB-169 photodiode configured for a reflectance measurement from the perfused skin. At present, only the single 800 nm wavelength was used, since it is an isosbestic point with respect to blood oxygen saturation, however, an additional measurement at a second wavelength (e.g., 680 nm) could readily be added for the purpose of measuring relative blood oxygen level at the expense of greater power consumption.

We also designed our system to use rechargeable batteries. This not only eliminates the need to purchase hundreds of batteries that may be needed for each study, but enables the battery to be completely embedded inside the wearable package, such as a shoe or sock, for weatherproofing and safety reasons (e.g., for use in infant monitoring).

IV. Wireless Platform

A. Design Objectives

The overall design objective for the wireless network was to enable wireless data collection and easy sharing, and access to the physiological data across a variety of devices, including personal computers, mobile phones, and the Internet. In addition, we were interested in supporting data collection from multiple (e.g., dozens) radio modules simultaneously. Additional primary design objectives for the radio module were to minimize size and maximize battery life.

B. Network Architecture and Wireless Protocol

To meet the needs of most health applications and medical research, we chose the wireless network architecture shown in Fig. 1 for communication between the radio modules, base station, mobile phones, and personal computers.

Several radio protocols and open standards are now available, however, most of them do not support multiple sensors and are not compatible with low-power radio hardware. For this reason, we chose IEEE 802.15.4, which in recent years has emerged as the dominant wireless protocol for low-power sensor networks, and is also the physical-layer protocol for Zigbee [21]. UltraWideBand standard IEEE 802.15.4a is a possible alternative in the future, but chipsets are not yet available. Although higher level transport protocols, such as Zigbee support multi-hop routing and mesh networking, we chose instead to adopt a star topology for our network in order to minimize processing overhead and power consumption.

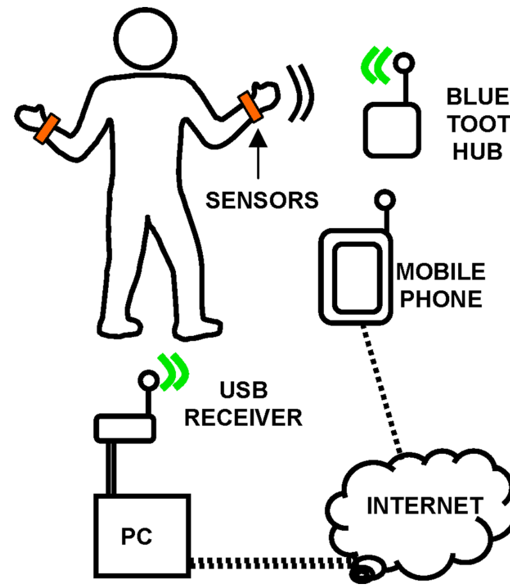

Fig. 1. Wireless Network Architecture for iCalm.

C. Operating Frequency

Many wireless sensors operate in the UHF range, e.g., 433 and 915 MHz, however, we chose to operate with 2.4 GHz in order to enable a smaller antenna size and achieve much better indoor radio propagation in typical buildings and homes due to the smaller wavelength.

D. Command Layer

In order to dynamically configure various parameters on the iCalm sensor band (such as sampling rate, transmit rate, and transmit power), a command protocol was added using the "command and response" paradigm. To send a command to a specific radio module, the user first sends the command to the radio base station. The command is then stored in the base station "command queue" until the specific radio module wakes up and transmits its data packet to the reader (base station). The radio module then receives and immediately executes the command before going back to sleep.

E. Radio Module

The radio module (see Fig. 2) consists of an Atmel Atmega328 microcontroller and a Chipcon CC2420 radio module. The radio module was designed to expose six 10-bit A/D ports on the microcontroller for interfacing with the sensor module. The reference voltage on these inputs can be configured via wireless commands from the radio base station.

The IEEE802.15.4 protocol was implemented in firmware with independent sampling and transmission intervals that can be set via wireless commands from the base station. Every transmission cycle, the radio module wakes up, and then, in turn

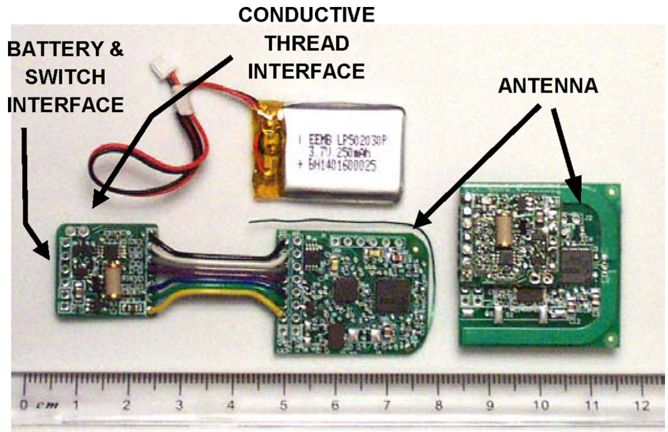

Fig. 2. (Left) Electronic modules (sensor + radio + battery) used for embedding in a shoe, and (right) module with integrated PCB antenna for use with wrist strap.

activates the power enable pin on the sensor module to power up the sensors. After a 10 ms delay, the radio module captures a 10-bit A/D sample from each of the sensors, transmits the data packet to the base station, and then goes back to sleep.

F. Radio Module Antenna

For better omnidirectional performance, at the expense of increased module size, we also used a version with an integrated printed circuit board (PCB) antenna. A bent-dipole, horseshoe-shaped antenna (see Fig. 2) was designed using Ansoft high-frequency structure simulator resulting in a compact design having a nearly isotropic radiation pattern.

G. Data Transport and Sampling Rate

Although the IEEE 802.15.4 communication hardware supports a 250 kbps data rate, the data packet length and transmission rate were reduced in order to minimize power. Since EDA data has a relatively low rate of change, we selected a slow rate of 2 Hz for sampling and packetized data transmission to enable a very low-operational duty cycle and long battery life.

For the PPG version of the sensor board, the standard sampling rate of 2 Hz was obviously insufficient. However, rather than using a much higher sampling rate and transmitting a long stream of raw analog values—which would have significantly increased power consumption—a completely different measurement method was employed. The filtered PPG waveform was hard-limited and fed to an interrupt-driven ICP capture pin on the radio module microcontroller that precisely measures the elapsed time between successive rising edges on the PPG waveform. Thus, instead of wirelessly transmitting raw analog-to-digital converter values, the PPG information was succinctly encoded in the transmitted data packet in the form of two numbers: the moving average of the past ten pulse intervals, and the difference between the current pulse interval and the running average. The PPG version of the sensor board could be programmed to run continuously or to only measure BVP periodically (e.g., once per minute) in order to save power.

H. Time Synchronization and Collision Mitigation

For medical applications that monitor data from multiple radio modules (e.g., on both left and right wrists), it is necessary to synchronize time between multiple sensors. The *ad hoc* asynchronous nature of the network does not automatically provide a common time base; thus, we programmed the radio base station to time stamp each arriving data packet in order to generate a proper time base of the measurements.

As part of our firmware implementation of IEEE 802.15.4 MAC layer, we also implemented the carrier sense multiple access (CSMA) algorithm, which provides exponential back off in the case of colliding transmissions between two or more radio modules.

I. Transmission Power and Operating Range

The CC2420 radio IC has a maximum transmission power of 1 mW (0 dBm), which provides a wireless detection range of 50–75 m in free space using a 5 dBi gain receiver antenna. Indoor range is significantly less and depends on the building layout, but is approximately 15–20 m for the module with integrated antenna and 8–10 m for the version with external antenna. These wireless operating distances are sufficient for our current health and medical research needs. The radio module also has controllable output power, therefore, the operating distance can be reduced to less than 1 m as one of several ways to address data privacy (see next).

J. Security and Privacy

Privacy and data security are important concerns in all of our research. In addition to controlling the radio output power, the CC2420 radio IC includes hardware support for the 128-bit advanced encryption standard, which can be turned on as an option. Our sensor devices also contain a user controlled ON/OFF switch, therefore, the user can choose to turn-OFF the data transmission when desired.

K. Radio Base Station

We have developed several different types of radio base stations or readers that are used to collect data from multiple radio modules and sensors. The most popular base station is the ZR-universal serial bus (USB), which has a USB interface to plug into PC's and laptops. This base station comprises the Atmega168V microcontroller, CC2420 radio IC, and the FTDI232BQ USB interface chip. A 50-Ω antenna port permits a variety of commercially available 2.4 GHz antennas to be used.

To enable applications that require sending physiological data to a remote Web site without a PC, we also installed a radio base station (TagSense ZR-HUB), which includes an embedded Linux computer that is programmed to upload data automatically to a remote Web server, and supports a JAVA API capable of running application-specific programs.

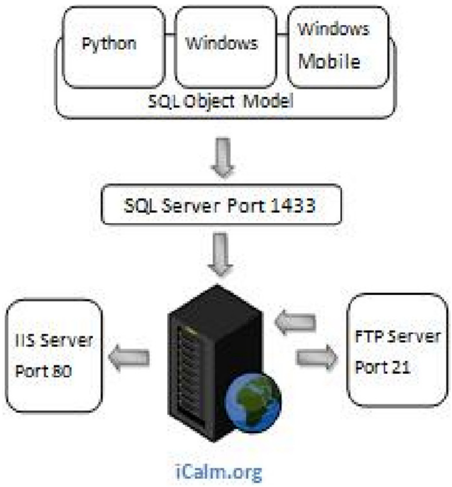

Fig. 3. Web server architecture used for iCalm showing connection to PC as well as mobile phone platforms.

L. Bluetooth Gateway

For certain applications involving a healthcare worker or researcher, it is desirable to receive and display physiological data directly on a mobile phone. For this purpose, we developed a Bluetooth gateway device that contains two radios (Bluetooth + 802.15.4) and is capable of bridging an IEEE 802.15.4 network and Bluetooth network. We named this device "personal-area-network Hub" (PAN-HUB). The PAN-HUB is powered by a 1760 mA·h rechargeable battery and contains a microSD card slot for expandable data storage. Custom software installed on the phone enables the phone to connect automatically when a given PAN-HUB is in range and gracefully manage disconnection when the user goes out of range.

M. Web Server, iCalm.org

A Web site was created to store, analyze, and share collected data. The server architecture is shown in Fig. 3. The site iCalm.org, runs ASP.net (on IIS Server) and draws data from an MSSQL database backend. Clients enter data into tables in a parsed format via an SQL object model, greatly reducing client-side table row insert errors and server load. This data is dynamically read by an ASP.net client via asynchronous postbacks to deliver database content to end-users. Because data transfer from front-end to back-end is asynchronous, live data can be read quickly and updated in near-real time.

iCalm.org has a front end ASP.net client. iCalm online was designed both to manage the iCalm Server and to act as a client itself. The key feature of the Web client is that all data inserted into the MSSQL databases is available in real time and in graphical form to data consumers. Users may select and view both current and previously recorded files from the database and see if any other users are sharing public data on the server. Additionally, iCalm.org allows users to filter through archived or live data by turning off data channels viewed on the graphical display. This is particularly useful for analyzing data from a specific channel (i.e., acceleration).

Fig. 4. Photograph of several form factors used: (upper left) baby sock with sewn electrodes, (upper right) wrist worn band, and (bottom) foot/ankle sensor band.

V. Form Factor and User Interface

A. Packaging and Electrodes

Great attention and effort were dedicated to the wearable form factor design and choice of materials to ensure user comfort and good signal integrity. Instead of the typical silver-silver chloride electrodes commonly used in medicine, we chose a washable, conductive material that allows the skin to breathe, maintains elasticity, and provides consistent contact with the skin: the medical grade silver plated Stretch Conductive Fabric ("a251") Nylon (92%) and Dorlastan (8%) sold by Textronix. We also used stainless steel electrically conductive thread made by Baekart to connect the electrodes to the sensor module circuit board. This enables greater comfort and greater durability since the conductive thread does not exhibit strain fatigue like traditional metal wires.

B. Form Factor and Fitting

The use of washable electrically conductive fabric and thread facilitates integration into wearable garments, and enables the use of metal clothing snap connectors. Using these materials, it is possible to create a variety of comfortable form factors. For example, in Figs. 4 and 5, we show various form factors that can easily be put on and taken off, and which are fully washable after slipping the electronics out of a small pouch. In order to maintain good fit and continuous contact with the skin, the conductive fabric electrodes are sewn onto a stretchable fabric mesh substrate with a Velcro tab, which enables the band to

Fig. 5. Sample textile electrode configurations used for wrist sensor.

remain snug at all times while still providing ventilation to the skin.

The wrist or ankle is not a standard location for measuring EDA, since the sweat glands there tend to be less densely distributed than those on the palm or fingers, where EDA is traditionally measured. This issue, coupled with our use of dry electrodes, means that it usually takes at least 15 min (depending on humidity and the individual's temperature) before the moisture buildup between the skin and electrodes is sufficient to show a range of responsiveness. The main advantage of sensing EDA from the wrist or ankle is that the sensor can be comfortably worn for long periods of time (days and weeks) by adults and by small children (ages 3–6) without interfering with daily activities, such as sleeping, washing hands, or typing.

In Fig. 4, we also show a version customized to wear on the feet of newborn infants (whose hands and wrists are often in their mouths, and thus, undesirable for sensing). The sole of the foot has eccrine sweat glands like the palm of the hand, and this site provides a more traditional measurement of EDA [14]. In this form factor, soft electrodes are sewn into socks and attached via small magnetic clothing snaps to the sensor, which is safely sealed into the top of the shoe so the infant cannot get to the small parts. This foot-worn sensor will be used in an upcoming study at Massachusetts General Hospital with infants born into families where there is a child diagnosed with autism. These siblings have an elevated likelihood of developing autism [22], so inclusion of our sensors in this research could help determine if reliable autonomic patterns can be detected in infancy to assist with diagnosis [23] and if these patterns change in response to early intervention [24]. Currently, EDA, temperature, and motion sensors are included in the infant shoe/sock package, while the adult wrist sensor package also includes an optional PPG sensor for HR and HRV measurement.

C. Software Interface

Several versions of data collection software and interfaces were designed for platforms, including Microsoft Windows, Macintosh, and Linux. The software displays either a live plot of sensor data or displays prestored data, and also records data for future use and upload to the Internet. A photograph of the platform (here, using OLPC's XO laptop) for data collection for our medical research on infants is shown in Fig. 6.

Fig. 6. Sample data collection platform, consisting of laptop and USB receiver module.

Often it is important to synchronize annotations with data. The software enables the user to insert data markers to annotate the data as it is being collected. In medical experiments, a separate pushbutton "remote control" wireless device can be used by the person conducting the experiment to insert wireless data packet indicating specific events or timestamps.

VI. Data Collection and Testing

One challenge in designing sensors is determining the best way to compare new sensors to state-of-the-art, Food and Drug Administration-approved medical sensors that gather the same signal information. Below we present results from both quantitative and qualitative laboratory tests, as well as provide sample data.

A. Testing and Calibration of HR Sensor

The PPG sensor used in our system was validated and tested using the commercial ECG device made by Alive Technologies as well as by manual pulse measurements. We found that when subjects were at rest, the HR readings from the PPG and ECG devices correlated within 10% of each other. However, since the PPG sensor hardware did not include any error correction algorithm or compensation for motion artifacts, the PPG signal was not reliable during motion of the hands. If necessary, a signal processing algorithm can be applied to correct for motion artifacts [25], furthermore, it should also be noted that there are many times during the 24 h day or night when a person's wrists are still, thus allowing for reliable snapshots of HRV.

B. Testing and Calibration of EDA Sensor

The process of validating and calibrating a new EDA sensor is complex given that many parameters contribute to the raw EDA signal, including circuit design (precision, dynamic range, and noise level), electrode design (shape, contact area, materials selection, placement, and contact pressure), and concentration of eccrine and other sweat glands at each specific measurement site.

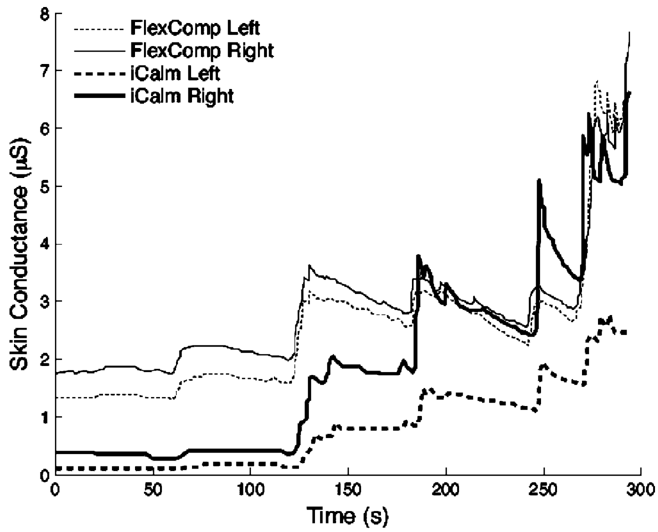

Fig. 7. Sample data showing comparison between an FDA-approved system on fingers and iCalm on wrists.

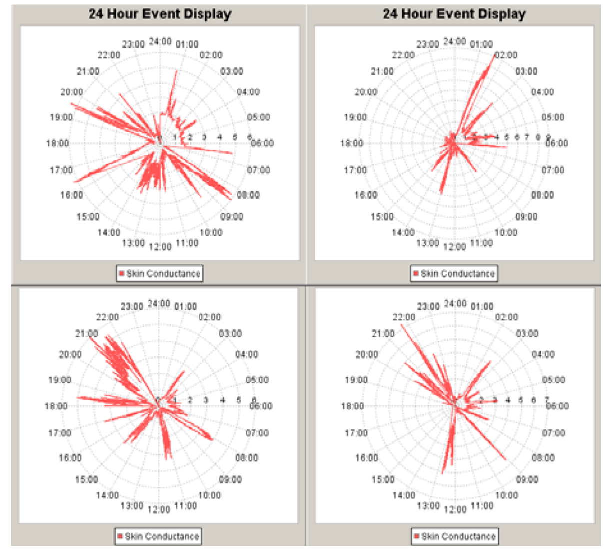

Fig. 8. Radial 24-h plots of EDA data from four days of wearing iCalm sensor bands on wrists. Skin conductance level is proportional to distance from the origin.

We carried out two types of tests to address these factors. First, independent of human configuration, we tested the isolated EDA circuit. We used a series of 1% fixed resistors ranging from 100 kΩ to 4.6 MΩ in place of the electrodes and human body. This range approximates the normal range of human skin resistance, allowing for assessment of the performance over the normal operating range of the sensors. The measured resistance from our sensor was compared to a laboratory multimeter, accurate to better than 1%. The resulting error across all resistance values ranged from 0.603% to −0.630%, with a mean of −0.148%.

In order to evaluate wrist electrodes on a person, we carried out a series of classic startle tests designed to produce an arousal response and collected data simultaneously from our EDA sensor (iCalm) attached to left and right wrists and from an FDA-approved "gold standard" system (FlexComp by Thought Technologies) connected to electrodes on fingertips of left and right hands. During the experiment, participants were seated and asked to relax for 5 min. At every one-minute interval, a loud noise was generated using an air horn to startle the participants. Data was collected from 12 people, one of which is shown in Fig. 7. As expected, skin conductance values were usually higher on the fingertips given the higher density of sweat glands, however, the data from the wrist electrodes on the iCalm device faithfully reproduced all the phasic features found in the data from the FlexComp system. The right skin conductance level was higher than the left in both the Flexcomp and iCalm recordings, which could be due to lateral differences following strong acoustic stimuli as reported in the literature [26], [27]. We also observed a slow increase in the tonic value of the iCalm data over time (particularly evident in the recording from the right hand iCalm unit in Fig. 7), which is likely due to the fact that the test subjects did not wear the sensor band for sufficient time before the experiment began in order to allow the skin electrode contact and tonic value to stabilize. From other longitudinal EDA measurements, we have observed that the sensor band should be worn for approximately 15 min or more to achieve a stable skin–electrode interface.

In addition to these controlled tests, we have successfully collected data 24/7 from several people wearing the sensors as they go about their natural life activities. Fig. 8 illustrates EDA data collected around the clock from one of several adults who wore the wrist sensor for weeks without disruption to daily activities (sensor removed only for showering). It is interesting to note that these plots include data at night: EDA has been shown to be of interest to sleep research [14] and sleep disorders are common in ASD [28].

VII. Conclusion and Future Work

As demonstrated in this paper, recent advances in low-power radio electronics and wireless protocols are enabling the development of new technology for long-term, comfortable sensing of autonomic information in new areas of health and medical research. New wearable materials, coupled with small long-lasting batteries, now provide the means to collect data over much longer time scales and in nonclinical settings, and the means for individuals to control the collection and communication of data by easily putting on or taking off the sensor (not needing the help of a researcher, and not having data sensed from them if they do not want to be sensed). We have shown data and evaluations in this paper to indicate that these new sensors, while nontraditional in their placement and design, are capable of gathering data comparable to data gathered with traditional sensors of EDA and HR. Thus, the sensors we have developed provide an important contribution over existing systems for gathering data in long-term naturalistic settings. It is our goal to help make lightweight portable sensor platforms such as the ones presented here accessible to a wider number of researchers and to individuals who wish to have help understanding and communicating their internal state changes. We envision that the strong connection between affective computing and health will also lead to new forms of understanding, diagnosing, and supporting the growing number of people who suffer from autonomic and affective disturbances.

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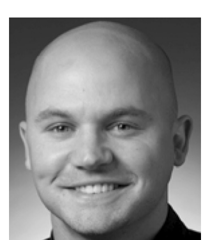

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