

A Prototype of an Integrated Blood Pressure and Electrocardiogram Device for Multi-Parameter Physiologic Monitoring

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Abstract—We present a prototype of an integrated blood pressure (BP) and electrocardiogram (ECG) device for multi-parameter physiologic monitoring. A standard BP pressure cuff and an ordinary wristband have been modified to incorporate in them dry ECG electrodes made of thin conductive fabric. The modified BP cuff and wristband are coupled with commercially available hardware and software to harvest simultaneous arterial pulse wave and ECG data from the arm and wrist of the other hand. Software has been written for assessing multiple physiologic parameters from the harvested pulse wave and ECG signals. We provide an initial validation of the performance of our prototype by conducting a study on six healthy subjects.

Keywords—blood pressure; electrocardiogram; physiologic monitoring; heart rate variability; pulse transit time; respiratory patterns

I. INTRODUCTION

Non-invasive blood pressure (BP) measurement is one of the principle methods for assessing cardiovascular health. BP may either be measured via the auscultatory method employing a stethoscope and a classic sphygmomanometer or via the oscillometric method which involves the observation and analysis of oscillations in the cuff pressure [1]. With the advent of computers and advances in digital signal processing, the oscillometric method has gained widespread popularity for measuring BP [2-3]. Today, there are many commercially available automated oscillometric BP monitors [4-5] which are not only being used in homes but also in clinical settings for assessing BP.

However, both oscillometric BP monitors as well as classic sphygmomanometers suffer from two main disadvantages. First, they provide a limited assessment and analysis of the physiologic system. For example, the most common information provided by these monitors may only include systolic pressure, diastolic pressure, mean arterial pressure (MAP), and mean heart rate (HR). Second, these monitors might produce erroneous results in presence of conditions such as arrhythmias [6-7], obesity [8-9], and movement / postural changes [10-11].

Simultaneous BP and electrocardiogram (ECG) monitoring promises to provide a more comprehensive assessment of the physiologic system, and robustness in BP estimation. For example, ECG-based motion artifact removal from arterial pulse waves has been employed for increasing accuracy of non-invasive BP measurements [12]. The use of an ECG signal has been proposed for reconstructing an arterial pulse wave signal contaminated by artifacts for providing accurate assessment of BP [13]. The AMON system combines pulse and ECG sensors in a single wrist worn enclosure for continuously collecting and evaluating multi-parameter vital signs [14]. Inukai et al. propose an apparatus for detecting an arterial pulse and ECG waveform using a BP cuff that incorporates a flexible ECG electrode made of conductive fabric [15]. A 2007 United States (US) patent application proposes to incorporate an ECG electrode inside a wrist BP cuff for simultaneous display and measurement of ECG and arterial pulse wave signals [16].

TABLE I. COMPARISON OF THE PROPOSED BP-ECG PROTOTYPE WITH EXISTING DEVICES / METHODS

Author (s)	BP-ECG Signal Quality	Integrated	Analysis / Physiologic Information	Remarks
Revision Labs Inc. [12]	Satisfactory	No	Systolic and diastolic pressure	ECG signal used only for motion artifact removal from arterial pulse waves.
Walloch [13]	Not provided	No	Systolic and diastolic pressure	Proposes the use of an ECG signal for reconstructing a noisy arterial pulse wave signal.
Anliker et al. [14]	Poor ECG signal	Yes	ECG signal, Systolic & diastolic pressure, SpO ₂ , pulse, and QRS interval	Compact design but poor and unusable ECG signal and SpO ₂ results. Accuracy for systolic and diastolic BP measurement was < 15 mmHg for 85% of subjects.
Inukai et al. [15]	Not provided	No	None	Proposes the use of conductive fabric as an ECG electrode inside a BP cuff.
Platt et al. [16]	Not provided	Yes	Arterial pulse wave and ECG signal	Proposes an integrated BP-ECG device for measuring and displaying these two signals.
Ahmad et al.	Satisfactory	Yes	Arterial pulse wave & ECG signal, heart rate variability, pulse transit time, and respiratory patterns	We have a working prototype of an integrated BP-ECG system with satisfactory signal quality and results of multi-parameter physiological assessment.

The aforementioned devices / methods show promise, however, each of them have certain limitations that restrict their practical applicability in providing comprehensive and accurate physiological assessment. While some of them report no or poor BP-ECG signal quality, others don't analyze these two signals to extract pertinent physiologic information. Finally, some of them are not fully integrated as a single device. Please see Table 1 for more details.

To address the above lacuna in the field of concurrent BP and ECG monitoring, we present a prototype of an integrated BP-ECG device for multi-parameter physiologic monitoring. We modify a standard brachial BP cuff by stitching a piece of thin conductive fabric on its inner side. This renders the cuff as a dry electrode for ECG data harvest in addition to serving its usual purpose of applying pressure on the arm for sensing a mechanical pulse. Another piece of conductive fabric is stitched on the inner side of an ordinary wristband. This wristband (worn on the other hand) acts as the second dry electrode for ECG data harvest. We use commercially available hardware and software for interfacing the BP-ECG cuff and the ECG wristband with a personal computer (PC) for simultaneous pulse wave and ECG data acquisition. We have developed software for extracting multi-parameter physiologic information from the acquired BP and ECG data. Our prototype currently has the capability to: (a) acquire, clean, and synchronize arterial pulse wave and ECG data, (b) identify ECG and arterial pulse peaks to derive RR and pulse to pulse (PP) interval time series, (c) perform heart rate variability (HRV) analysis on RR and PP interval time series, (d) estimate pulse transit time between

ECG and arterial pulse peaks, and (e) extract respiratory patterns from ECG and arterial pulse wave data.

The novelty of our prototype lies in the successful design of a multipurpose BP-ECG cuff and an ECG wristband which enable it to satisfactorily harvest both arterial pulse wave and ECG data from the arm and wrist. Moreover, our prototype performs a comprehensive analysis of the acquired arterial pulse wave and ECG data to output a variety of physiologic information. To the best of our knowledge, no published work reports such a method for acquiring and analyzing arterial pulse wave and ECG data in a single integrated system (Table 1).

In this paper, we describe our integrated BP-ECG prototype and provide an initial validation of its performance by conducting a study on six healthy subjects.

II. METHODS

A. Integrated BP and ECG Device

The illustrative block diagram of our prototype is presented in Fig. 1. A brachial BP cuff is modified by stitching a strip of conductive fabric on its inner side (A). We use a medical grade silver plated 92% nylon and 8% dorlastan stretchable conductive fabric (Block EMF, HealthStores.com, Carlsbad, CA, USA). The thickness of this conductive fabric is 0.50 mm and its surface resistivity is less than $1 \Omega/\square$. The length of the conductive fabric strip is chosen to be equal to the width of the cuff (14.5 cm). The width of the conductive fabric strip is chosen to be equal to 7.0 cm, resulting in a conductive fabric area of (14.5 cm x 7.0 cm) 101.5 cm^2 .

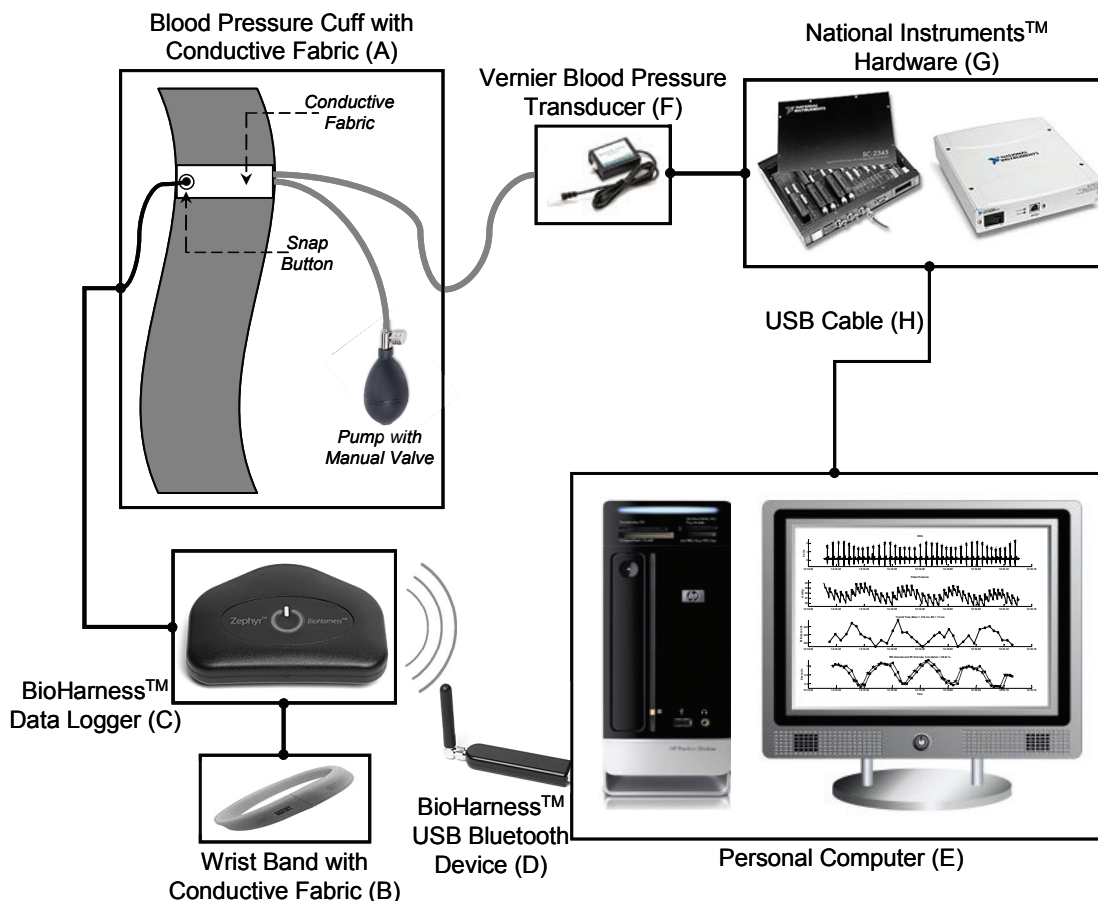


Figure 1. Illustrative block diagram of the integrated BP-ECG prototype

The conductive fabric strip is stitched on the inner side of the cuff over the portion where tubes from the bladder come out. The orientation of the conductive fabric strip is kept such that its length is parallel to the width of the cuff. The conductive fabric area, location, and orientation on the cuff ensure that it makes proper and unbreakable contact with the skin when the cuff is wrapped around the arm and when it inflates or deflates. The female part of a metallic snap button of diameter 15 mm is riveted on the conductive fabric strip. This ‘female’ snap button acts as a terminal for connecting the conductive fabric strip (BP cuff) to other hardware. A wire of desired length is soldered to the male part of the metallic snap button which is then snapped on to the female part of the metallic snap button on the conductive fabric strip. A manual pump with a manual pressure valve is connected to the BP cuff. The manual pressure valve allows us to apply desired pressure on the arm. In addition, it gives us the choice of maintaining a constant pressure for harvesting continuous arterial pulse wave data. The BP cuff with conductive fabric (A, Fig. 1) acts as a dry ECG electrode in addition to its usual function of applying pressure on the arm for sensing a mechanical pulse.

Another strip of conductive fabric is stitched on the inner side of an ordinary wristband (B, Fig. 1). The area of this conductive fabric strip is (12.0 cm x 4.0 cm) 48.0 cm² and it is stitched along the length of the wristband. Again, male and female parts of a metallic snap button are used to connect the conductive fabric strip (wristband) to other hardware. When worn on the wrist of the other hand, the wristband with conductive fabric (B, Fig. 1) acts as the second dry ECG electrode.

We use a chest-belt telemetry system (BioHarness™, Zephyr Technology Ltd., Auckland, New Zealand) for ECG data acquisition. The BioHarness™ telemetry system comprises a data logger which transmits ECG data sampled at 250 Hz to a PC via a universal serial bus (USB) Bluetooth device using proprietary software. The BioHarness™ system has a user

interface which apart from other functionality can display time stamped ECG data in real time. The BioHarness™ system uses the PC time for time stamping ECG data. Once monitoring is complete, the user can click on an export button on the user interface to export the acquired time stamped ECG data to a comma-separated values (CSV) file. As shown in Fig. 1, we connect our BP cuff with conductive fabric (A) and wristband with conductive fabric (B) to the BioHarness™ data logger (C). The BioHarness™ data logger (C) transmits the harvested ECG data via the BioHarness™ USB Bluetooth device (D) to a PC (E).

For acquiring arterial pulse wave data, we use a Vernier pressure transducer (SenSym SDX05D4) and National Instruments™ (Austin, TX, USA) hardware and software. As shown in Fig. 1, the pressure output from the BP cuff with conductive fabric (A) is fed to the Vernier pressure transducer (F). This transducer converts mechanical vibrations in the cuff (produced due to arterial BP) to an analog voltage signal in the range of 0-5 V. This voltage range of 0-5 V corresponds to a pressure range of 0-200 mmHg. The analog voltage output from the pressure transducer (F) is fed to a National Instruments™ (NI) signal conditioner carrier (SC 2345) (G). Inside the SC 2345, the analog signal passes through a low pass filter module (NI-SCC-LP04) to reduce noise. This filter is a fourth order Butterworth low-pass filter (cut-off frequency = 1 kHz, gain = 0.5, maximum input range = ± 10 V). The output from the SC 2345 is connected to the input port of a NI data acquisition (DAQ) device (NI USB-6251, 16-Bit, 1.25 MS/s, M Series Multifunction DAQ) through a 68-pin shielded cable. The DAQ device (G) digitizes the analogue pressure signal to a digital signal and transmits the quantized samples to a PC (E) via a USB cable (H). We use NI LabVIEW development environment for acquiring and controlling the acquisition of arterial pulse wave data. We have developed a customized LabVIEW user interface which in addition to other functionality can display the acquired arterial pulse wave data in real time.

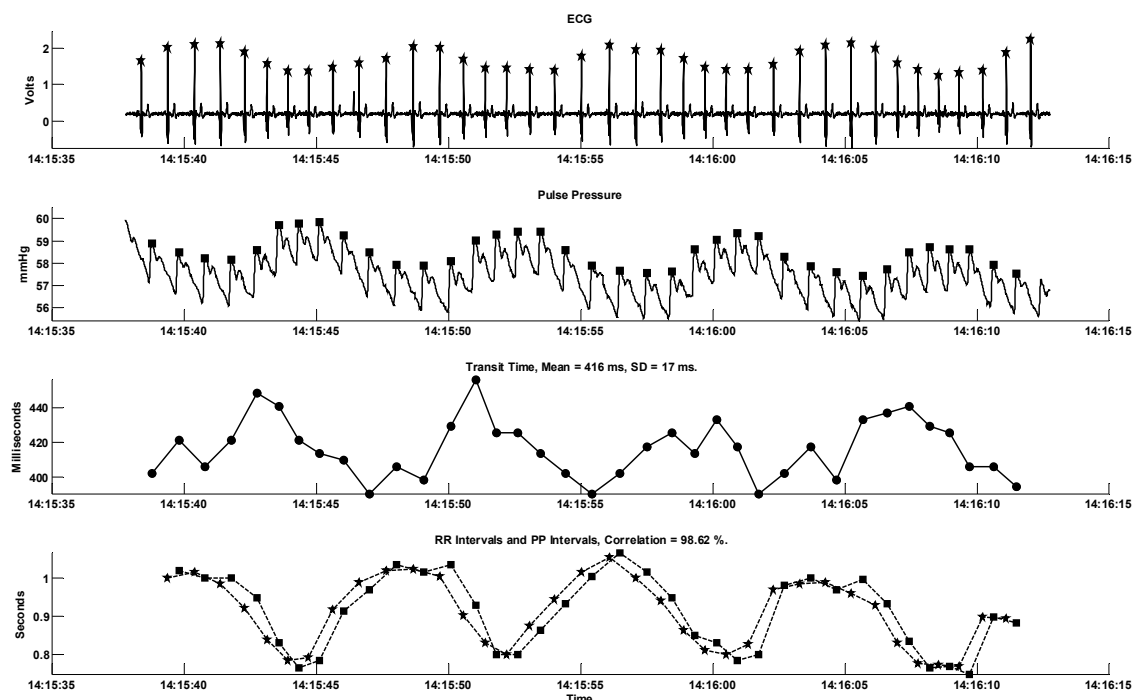


Figure 2. First 30 s of data harvested and analyzed by the integrated BP-ECG prototype for Subject # 1

TABLE II. PULSE TRANSIT TIME, RR-PP INTERVAL CORRELATION, AND HRV ANALYSIS FOR ALL SIX SUBJECTS

Subject #	Gender	Age (yrs)	Signal Duration (s)	Pulse Transit Time (ms)		RR-PP Correlation	Measure	Heart Rate Variability (HRV)				
				Mean	SD			Mean HR (BPM ^a)	SD	RMSSD	FFT LF/HF	MODWT AUC
1	M	23	90	417	14	98%	ECG Based HRV	69	0.084	0.0566	6.68	-10.59
							Pulse Wave Based HRV	69	0.090	0.0649	6.64	-10.17
2	M	32	90	547	24	90%	ECG Based HRV	71	0.068	0.0466	8.53	-11.07
							Pulse Wave Based HRV	71	0.075	0.0729	7.47	-10.22
3	M	37	90	464	34	84%	ECG Based HRV	65	0.079	0.0658	5.95	-10.16
							Pulse Wave Based HRV	65	0.084	0.0960	5.06	-10.05
4	F	18	90	602	24	92%	ECG Based HRV	79	0.074	0.0478	7.75	-10.90
							Pulse Wave Based HRV	79	0.073	0.0656	6.70	-10.59
5	F	25	90	510	39	82%	ECG Based HRV	84	0.073	0.0486	4.85	-10.92
							Pulse Wave Based HRV	84	0.085	0.0886	3.85	-9.83
6	F	27	90	470	19	99%	ECG Based HRV	77	0.163	0.1085	9.71	-7.56
							Pulse Wave Based HRV	77	0.164	0.1158	10.01	-7.34

a. Abbreviation BPM stands for beats per minute.

Once monitoring is complete, the user can click on an export button on the user interface to export the acquired time stamped arterial pulse wave data to a Technical Data Management Streaming (TDMS) file. The arterial pulse wave data is time stamped based on the PC time. We have set the sampling frequency of the arterial pulse wave data acquisition to 250 Hz.

B. Arterial Pulse Wave and ECG Data Analysis

We have written software in Matlab[®] (The MathWorks Inc., Natick, MA, USA) for reading, processing, and analyzing the concurrently acquired arterial pulse wave and ECG data. TDMS files containing arterial pulse wave data are read using an adapted C-header file and Matlab[®] code available from the NI website [17] while CSV files containing ECG data are read using Matlab[®] code that we have written. The quality of the acquired ECG data sampled at 250 Hz is good and as such requires no cleaning. However, the acquired arterial pulse wave data sampled at 250 Hz has 60 Hz and other noise and requires cleaning. To clean the arterial pulse wave data, we pass it through a fifth order linear-phase finite impulse response (FIR) band pass filter with a pass band between 1-15 Hz. We then up sample the cleaned 250 Hz arterial pulse wave data to a sampling frequency of 1000 Hz for synchronization with ECG data. The 250 Hz ECG data is also up sampled to a sampling frequency of 1000 Hz. The 1000 Hz arterial pulse wave and ECG data are then synchronized. The up sampling of data to a sampling frequency of 1000 Hz ensures that least count of the timestamps is 1 ms, resulting in accurate synchronization.

ECG R-peaks are identified using MIT / PhysioNET Matlab[®] QRS onset detector software [18-19], which we have modified to augment its speed and accuracy. Arterial pulse peaks are detected by finding the maximum amplitude of each pulse peak between two R peaks. Based on the identified R and pulse peaks, corresponding RR and pulse to pulse (PP) interval time series are derived. Five HRV parameters, namely mean heart rate (HR), standard deviation (SD), root mean square

successive difference (RMSSD), fast Fourier transform (FFT) low frequency to high frequency (LF/HF) ratio, and maximal overlap discrete wavelet transform (MODWT) spectral density area under curve (AUC) are computed on the RR and PP interval time series. Details about these HRV characterization techniques can be found in [20-23]. Pulse transit time [24] is computed by finding the difference between the temporal locations of each arterial pulse peak and R-peak.

Respiratory patterns are extracted from arterial pulse wave and ECG data based on the amplitude and frequency modulation methods [25-28]. Four different algorithms namely ECG amplitudes, ECG intervals, pulse wave amplitudes, and pulse wave intervals are employed for extracting respiratory patterns. To remove the effects of noise and long term drift, the extracted respiratory signals are cleaned utilizing the difference of two moving averages of different window sizes [29]. Instantaneous and overall respiratory frequencies or breathing rates are estimated from the cleaned respiratory signals using the zero crossing method [30].

C. Study on Healthy Subjects

As an initial validation of our integrated BP and ECG prototype, we performed a study on six healthy subjects ($N_T = 6$, Age Range: 18-37 yrs) out of which three were males ($N_M = 3$, Age Range: 23-37 yrs) and three were females ($N_F = 3$, Age Range: 18-27 yrs). These subjects are part of our research group and they volunteered to participate in this study. To the best of our knowledge, no subjects had any history of cardiovascular or respiratory disease.

For each subject, we harvested and analyzed 90 s of arterial pulse wave and ECG data using our prototype. We used the manual pressure valve of our device to maintain a constant cuff pressure in the range of 40-60 mmHg for each subject. This pressure range (40-60 mmHg) ensured that we operated below subjects' diastolic pressure, thus enabling comfortable monitoring. The subjects were asked to deep breathe at a constant rate during the entire monitoring period. A metronome

was used to help subjects regulate their breathing. The subjects were asked to inspire for four seconds and then to expire for four seconds, so as to complete one breathing cycle in eight seconds. This resulted in a constant breathing rate of 7.5 breaths per minute for the entire 90 s of monitoring for each subject.

For each subject, for the entire 90 s of monitoring, we assessed: (a) the correlation between RR and PP intervals, (b) mean and SD of transit time, (c) five HRV parameters (Mean HR, SD, RMSSD, FFT LF/HF, and MODWT AUC) measured from RR and PP interval time series, and (d) overall breathing rates employing the four ECG and arterial pulse wave amplitude and frequency modulation algorithms.

III. RESULTS

Fig. 2 shows the first 30 s of data harvested and analyzed by our prototype for Subject # 1. Panel one shows the ECG signal along with the identified R-peaks (stars). The arterial pulse wave signal with the identified pulse peaks (squares) is shown in panel two. Pulse transit time (line and dots) is shown in panel three while the RR (dotted line and stars) and PP (dotted line and squares) interval time series are shown in panel four. The quality and fidelity of the acquired ECG (Panel # 1) and arterial pulse wave (Panel # 2) signals are satisfactory. The effects of regulated deep breathing are eminent in the periodically changing amplitudes of the ECG and arterial pulse wave signals (amplitude modulation due to breathing). Similarly, the prominent temporal cyclicity of RR and PP intervals (Panel # 4) also results from regulated deep breathing (frequency modulation due to breathing). The RR-PP interval correlation is 98.6% and the pulse transit time is 416 ± 17 ms.

Table 2 shows results for pulse transit time, RR-PP interval correlation, and HRV analysis for the six subjects. For these subjects, the pulse transit time is in the range of 417 ms to 602 ms with a median of 490 ms. The correlations between RR and PP interval time series are high and in the range of 82% to 99%. Moreover, for each of the six subjects, there is good agreement between ECG-based and arterial pulse wave based HRV parameters.

The overall breathing rates for all six subjects derived using the four respiratory pattern algorithms are presented in Table 3. The performance of each of the four algorithms is assessed by computing the mean absolute percentage error, $MAPE = \frac{1}{N} \sum_{i=1}^N \left| \frac{A_i - D_i}{A_i} \right|$, where A_i is the actual breathing rate for the i th subject, D_i is the derived breathing rate for the i th

subject, and N is the total number of subjects, that is, six. The ECG intervals algorithm (Algorithm # 2) achieves the best performance with the lowest MAPE of 0.942%, followed by the ECG amplitudes algorithm (Algorithm # 1, MAPE = 1.016%), arterial pulse wave intervals algorithm (Algorithm # 4, MAPE = 1.036%), and arterial pulse wave amplitudes algorithm (Algorithm # 3, MAPE = 1.109%).

IV. DISCUSSION

We presented a prototype of an integrated BP and ECG device for multi-parameter physiologic monitoring. We performed a study on six healthy subjects using our prototype and showed that it can satisfactorily harvest simultaneous arterial pulse wave and ECG signals and extract accurately a variety of physiologic information from these signals.

Our prototype, inspired by a conventional BP monitor, offers certain important advantages over existing BP and multi-parameter monitoring devices. The modification of a brachial BP cuff to serve the dual purpose of arterial pulse wave and ECG data harvest offers compactness to our device by reducing the number of physiologic sensors that need to be attached to a subject. This compactness of design makes our device easier and more convenient to use. Moreover, the modified BP cuff and the wristband which act as dry electrodes make the process of ECG data harvest less intrusive and more comfortable as compared to systems that harvest ECG data from the chest. The variety of physiologic information that our prototype extracts from arterial pulse wave and ECG signals promises to provide a more comprehensive assessment of the physiologic status of an individual. For example, HRV analysis can be utilized for studying conditions such as arrhythmias and heart failure, pulse transit time can be used to assess blood pressure and arterial stiffness, and breathing patterns along with HRV analysis can provide insights into the study of conditions such as sleep apnea.

We are currently working on enhancing the performance and functionality of our prototype. Future work would involve further improving the quality and fidelity of the arterial pulse wave signals in particular. We observed that our arterial pulse wave signals were contaminated with 60 Hz and other noise. Moreover, these signals were prone to more noise at cuff pressures lower than 40 mmHg. We are exploring the use of newer pressure transducers, circuitry, and filtering approaches to improve the quality and stability of the acquired arterial pulse wave signals at pressures between 30-40 mmHg. Clean pulse wave signals acquired at pressures lower than 40 mmHg would

TABLE III. OVERALL BREATHING RATES FOR ALL SIX SUBJECTS DERIVED USING THE FOUR RESPIRATORY PATTERN EXTRACTION ALGORITHMS

Subject #	Gender	Age (yrs)	Signal Duration (s)	Actual Breathing Rate (Breaths Per Minute)	Derived Breathing Rate (Breaths Per Minute)			
					Algorithm # 1 (ECG Amplitudes)	Algorithm # 2 (ECG Intervals)	Algorithm # 3 (Pulse Wave Amplitudes)	Algorithm # 4 (Pulse Wave Intervals)
1	M	23	90	7.5	7.511	7.436	7.512	7.436
2	M	32	90	7.5	7.422	7.350	7.464	7.392
3	M	37	90	7.5	7.693	7.611	7.736	7.654
4	F	18	90	7.5	7.552	7.446	7.568	7.502
5	F	25	90	7.5	7.535	7.473	7.449	7.388
6	F	27	90	7.5	7.588	7.518	7.596	7.526
Mean Absolute Percentage Error (MAPE)					1.016%	0.942%	1.109%	1.036%

allow for more comfortable long term monitoring. We are also working on the development of algorithms that would enable future versions of our device to output newer physiologic information such as respiratory sinus arrhythmia (RSA), systolic, and diastolic blood pressure. We are also working on the development of algorithms that would enable future versions of our device to output newer physiologic information such as respiratory sinus arrhythmia (RSA), systolic, and diastolic blood pressure.

In this initial investigation, we tested the performance of our prototype by analyzing one 90 s recording from each of the six healthy subjects. Further trials are being planned whereby we will validate our prototype based on multiple recordings from each subject and a larger subject database.

The ultimate vision for our prototype is to provide a compact, portable, easy to use, cheap, and robust multiparameter physiologic monitoring device with universal applicability. We anticipate a cost of \$ 500 for the final developed version of our device.

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