Towards a Broadly Configurable Wearable Device for Continuous Hemodynamic Monitoring

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***Abstract*—Hemodynamic signals, such as Electrocardiogram (ECG), Photoplethysmography (PPG), and Electrical Bioimpedance (EBI), provide a myriad of critical health indicators. For instance, ECG is utilized by cardiologists to diagnose a range of cardiovascular diseases, PPG estimates blood oxygen levels, and EBI gauges vascular flow. A multimodal device capable of continuously monitoring these signals can furnish healthcare professionals with profound insights into patient well-being. This paper introduces a hardware design for a programmable, wearable device that measures multiple physiological signals, including ECG, PPG, and EBI. Equipped with an onboard microprocessor, the device enables real-time data processing. Compact and efficient, the system measures just 60.5mm by 38.75mm, has low-power dissipation, and is powered by a coin cell battery. Using largely discrete commercial components and a custom electrode array, the open-source hardware allows for rapid configurability and broad usage applications within the health monitoring domain.**

I. Introduction

Cardiovascular disease is the leading cause of death globally, with an estimated 20.5 million lives being taken each year [1]. Heart and vascular disorders affect approximately 121.5 million adults, or 48% of individuals in the United States alone [2]. Hemodynamically based signals such as Electrocardiogram (ECG), Photoplethysmography (PPG), and Electrical Bioimpedance (EBI) offer critical health indicators essential to modern medical diagnostics for these diseases. ECG serves as a non-invasive tool for cardiologists to diagnose various heart diseases by capturing the heart's electrical activity and identifying abnormal patterns. PPG measures blood oxygen levels, providing vital data on pulmonary function and circulatory health, while EBI provides insights into body composition and fluid distribution, crucial for monitoring vascular flow [3]. Non-invasive monitoring of vascular flow can offer indispensable information for assessing the health of vascular and tissue systems without the need for intervention or surgery [4]. Collectively, these technologies enable a comprehensive evaluation of a patient's health, facilitating the early detection and management of medical conditions.

A variety of commercial and academic devices are available that support the measurement of ECG and PPG [5], [6]. Portable devices that measure EBI have also been developed, enhancing the assessment of vascular flow and body composition. Notably, devices described in Hersek's 2017 study highlight the utility of EBI in wearable technology, providing deeper understanding of fluid dynamics and tissue properties [7], [8]. Such devices are instrumental in the prompt detection of physiological abnormalities, enabling clinicians to more accurately diagnose problems and recommend suitable treatments.

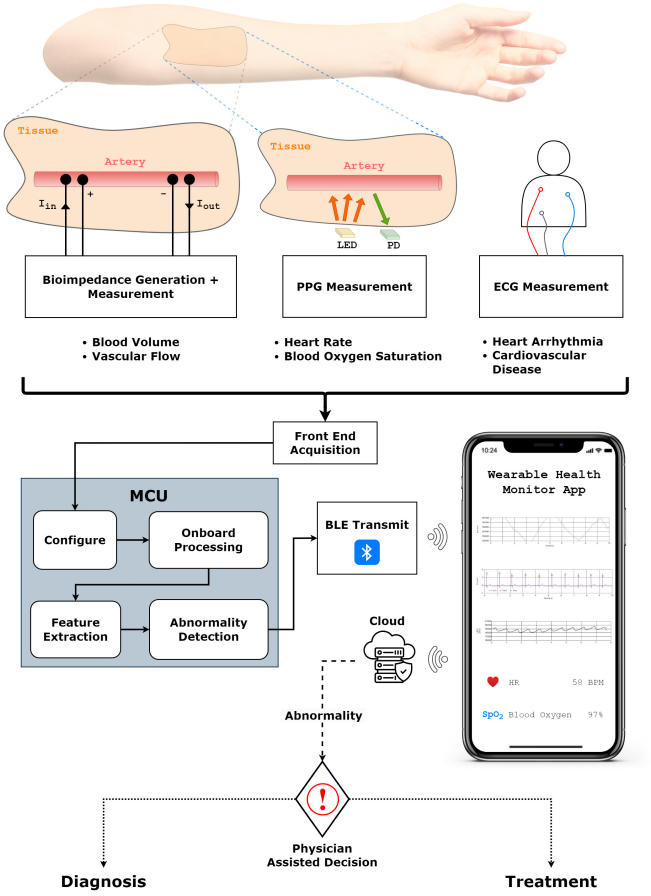

Fig. 1: Overall system flowchart displaying real-time onboard processing and configurability of the device for multiple biosignals.

Typically, wearable device designs focus on specific hardware systems tailored to measure distinct physiological signals. Commercially available PPG devices, for example, are engineered specifically to assess blood oxygen levels [6]. Devices dedicated to ECG recordings primarily monitor cardiac activity [5]. However, comprehensive patient health assessment often necessitates the simultaneous measurement of multiple physiological signals.

Furthermore, previous designs of hemodynamic measuring devices often feature large, discrete, front-end-only platforms rarely integrated onto a single device. Few possess both onboard processing and wireless transmit capabilities. For example, the cuff-less blood pressure monitoring wearable introduced in Rachim and Chung's 2019 study includes a small wrist worn EBI and PPG front end sensor with integrated electrodes [9]. However, only the analog front end (AFE) is integrated on the wearable; data acquisition and processing units remain offboard. Similarly, the device in Hersek's 2017 study presents a wearable EBI AFE with an onboard MCU and discrete electrodes, but important post-processing tasks such as feature extraction are performed offboard without wireless transmit [7]. Even fewer systems include the various channels of biosignal sensing capacity needed for broader hemodynamic monitoring.

To address this need, by utilizing readily available discrete commercial bio-aquisition front ends, this work aims to describe a smaller yet more comprehensive, more power efficient yet similarly capable platform more suitable for wearable applications. An open-source, programmable, highly-integrated wearable platform capable of simultaneously measuring ECG, PPG, and EBI is thereby presented. This device also features an onboard microprocessor that supports real-time processing of physiological signals, significantly enhancing its capability for continuous health monitoring. Figure 1 depicts the overall system architecture.

II. Biosignal Measurement

A. Electrocardiogram

The electrocardiogram is the most commonly used noninvasive technique for gleaning information about the cardiac muscle contraction sequence and nominal heart functioning. Physiological markers which can be monitored through an ECG's QRS complex include R-to-R heart rate, arrhythmia, pacemaker malfunction, cardiomyopathy, rheumatic heart disease, along with symptoms associated with cardiac diseases such as tachycardia and bradycardia, among others [10].

The measurement of ECG signals in a clinical setting typically involves a standard 12-lead setup [11]. However, for most wearable ECG applications, a 3-lead or single-lead setup is sufficient for basic diagnostic purposes [12]. At its core, an ECG measures the potential difference between sets of two points on the skin, otherwise known as leads, as electrical impulses from the heart traverse through the body. In its simplest form, this is achieved using two conductive electrode pads interfacing the skin, which are fed through as inputs to an instrumentation amplifier [13]. The instrumentation amplifier is composed of a difference amplifier with two input buffer amplifiers, negating the need for impedance matching and providing gain adjustability through a single resistor.

Raw ECG measurements are inherently noisy signals, and external artifacts such as baseline drift, 50/60 Hz power line coupling, electrode contact quality, respiration, and motion artifacts must be accounted for [14]. Signal noise can be mediated with both analog and digital bandpass filters. To ensure that the common-mode voltage of the leads avoid amplifier saturation, the patient's body must be voltage driven to a level such that the electrodes are DC biased to within the input range. This is commonly performed using a right-leg drive (RLD) circuit, requiring an additional "ground" reference electrode. However, 2-electrode systems offer greater hardware efficiency, and can be achieved by biasing the input electrodes to within the voltage range of the acquisition system's data converter [15]. Portable and wearable ECGs have gained in prevalence. However, accurate ECG signals must still be measured across the heart, limiting the local measurement ability of limb-worn wearables, thereby constraining many long-term monitoring use cases.

B. Photoplethysmography

Found in many health wearables, photoplethysmography offers a more compact, more localized, but less power efficient sensing solution in comparison to ECG for certain cardiac monitoring applications. In principle, PPG operates by illuminating skin tissue with specific wavelengths of light emitted from an LED. Light absorbed, reflected, or transmitted through the skin is then measured with a photodiode sensor and converted to an electrical signal, which can vary as a subcutaneous organ changes in volume. With each cardiac cycle, pressure waves sent through the cardiovascular system cause blood vessels to dilate, which can be measured via PPG. A large limitation to PPG systems are their susceptibility to motion and external light artifacts. Additional 3-axis calibration accelerometers and ambient-light cancelling algorithms are therefore necessities when such systems are used in wearables [6].

Much like ECG, the periodicity of the PPG signal can be used to determine heart rate. For heart rate monitoring, green ($\sim$530 nm) LEDs are typically used. For blood oxygen saturation monitoring (SpO_2), red ($\sim$660 nm) and infrared ($\sim$940 nm) LEDs are typically used [16]. PPG is often used in conjunction with other biosignals such as ECG and EBI. For example, to calculate pulse transit time (PTT) for systolic blood pressure (SBP) estimation, a key marker for hypertension, the time difference between the R-peak of an ECG signal and the systolic peak of a concurrently measured PPG signal must be calculated, highlighting the importance of parallel multi-biosignal measurement [17].

C. Electrical Bioimpedance

Electrical bioimpedance measurements provide another host of biomonitoring applications, namely blood volume measurements and vascular flow. For impedance plethysmography (IPG) utilizations, injected AC current from an oscillator is passed through an area of skin containing a target artery via an electrode pair. As blood flow volume pulses, the impedance of subsurface tissue is altered. These pulses are measured using an additional set of voltage detect electrodes measuring the change in potential difference across the area of skin. Similarly to ECG, this voltage is measured using an instrumentation amplifier setup. Due to the nature of skin

presenting a high complex impedance against low frequency current, high frequency low magnitude sinusoidal AC current must be used [18]. Typically, IPG injection current frequencies lie between 10kHz to 100kHz for sufficient tissue penetration [19]. Current magnitudes usually remain under the perception threshold $I_{max} = 10^{-7} * freq$ to maintain user safety under the IEC-60601 standard [20]. EBI acquisition hardware must also leverage phase-sensitive detection to extract both resistive and reactive components of skin impedance.

Following acquisition, the measured EBI waveform is used to construct the IPG signal using (1).

$$IPG = \frac{dEBI}{dt} \tag{1}$$

Subsequently, pulsatile blood volume (ΔV_{bd}) can be found using (2), where ρ is blood resistivity, L is the distance between voltage electrodes, R_{dc} is the DC resistance of tissue, Δt_{pulse} is the time difference between the minima preceding and following each EBI pulse, and $\left(\frac{dEBI}{dt}\right)_{\max}$ is the rising edge amplitude of the IPG signal.

$$\Delta V_{bd} = \rho \left(\frac{L}{R_{dc}}\right)^2 \Delta t_{pulse} \left(\frac{dEBI}{dt}\right)_{\max} \tag{2}$$

Finally, the blood flow rate (Q_{bd}) is calculated using (3), where HR corresponds to heart rate.

$$Q_{bd} = \Delta V_{bd} \cdot HR \tag{3}$$

As such, to calculate blood flow rate, EBI must be measured concurrently with either the systolic peaks from PPG or R-to-R detection from ECG [21].

III. Hardware Architecture

The wearable device hardware consists of several tightly integrated biosignal front ends capable of monitoring ECG, EBI, and PPG signals, in addition to an onboard MCU with wireless transmit capabilities and a tailored power management system. The platform measures 60.5mm by 38.75mm at its widest and is implemented on a 4-layer PCB. The overall hardware block diagram presented in Figure 2 visualizes the EBI + ECG and PPG front ends, power, and data acquisition (DAQ) module blocks. A labelled view of the custom fabricated electrodes and device is displayed in Figure 3. The system's hardware design files are open-sourced and available to the public.

A. ECG & EBI Front End

The EBI front end interfaces with the patient's skin using a custom flexible dry electrode array. The electrode array can be printed on a flexible substrate, or fabricated as a flex PCB where the electrode pads are coated with Ag/AgCl. The onboard ribbon connector enables tailor designed electrodes for maximum adaptability regardless of device placement on the body. To collect data across varying lengths and current penetration depths of tissue, the default wearable device utilizes an array of four sets of current generation and voltage detect electrodes. The electrode pairs are spaced apart by distances from 1cm to 4cm in 1cm increments. The pairs can be selected using a 2-channel 4:1 multiplexer (MUX).

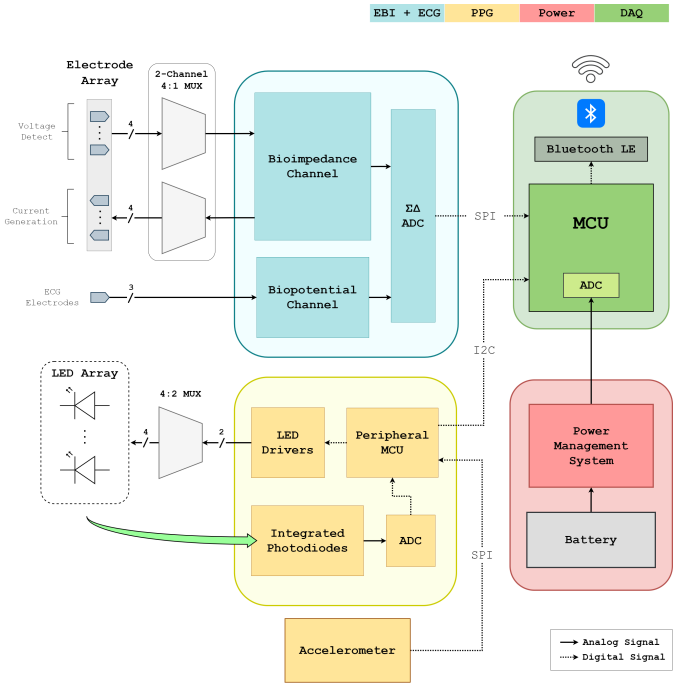

Fig. 2: Block diagram of the hardware.

The ECG front end input is accessed through a 3.5mm barrel connector jack using a set of commercial 3-lead ECG electrodes, with the third reference electrode optional. To obtain a clear signal, the positive and negative leads must cover sufficient body surface area such that the ECG signal vector can provide sufficient potential difference to be detected.

The acquired ECG and EBI signals are passed through passive low pass filters before being acquired by a MAX30001 Ultra-Low-Power Integrated Biopotential and Bioimpedance AFE (Analog Devices, Inc, Norwood, MA). Both the ECG and EBI channels have their own respective 18-bit and 20-bit $\Sigma\Delta$ analog-to-digital converters (ADC's). The MAX30001 is supplied with a constant 1.8V V_{DD} level.

For both ECG and EBI signal acquisition, internal DC lead biasing drives the connected tissue to an appropriate common mode voltage level when using a single lead (two electrode) setup. The common-mode voltage of the positive and negative electrodes is body biased to an average DC level (V_{MID}), which can be calculated as $V_{MID} = \frac{V_{DD}}{2.15} = 837mV$. The input electrode voltage range can subsequently be bandlimited to $V_{MID} \pm 550mV$ maintaining within the saturation requirements of the MAX30001's V_{DD}. For EBI signal acquisition, the magnitude and frequency of the injection current is selectable to be between 8µA to 96µA and 125Hz to 128kHz, respectively.

B. PPG Front End

The system also contains a complete photoplethysmography system capable of monitoring both pulse heart rate (HR) and pulse blood oxygen saturation (SpO_2). The PPG system contains an LED array consisting of two green (536 nm) LEDs, a red (655 nm) LED, and an infrared (940 nm) LED.

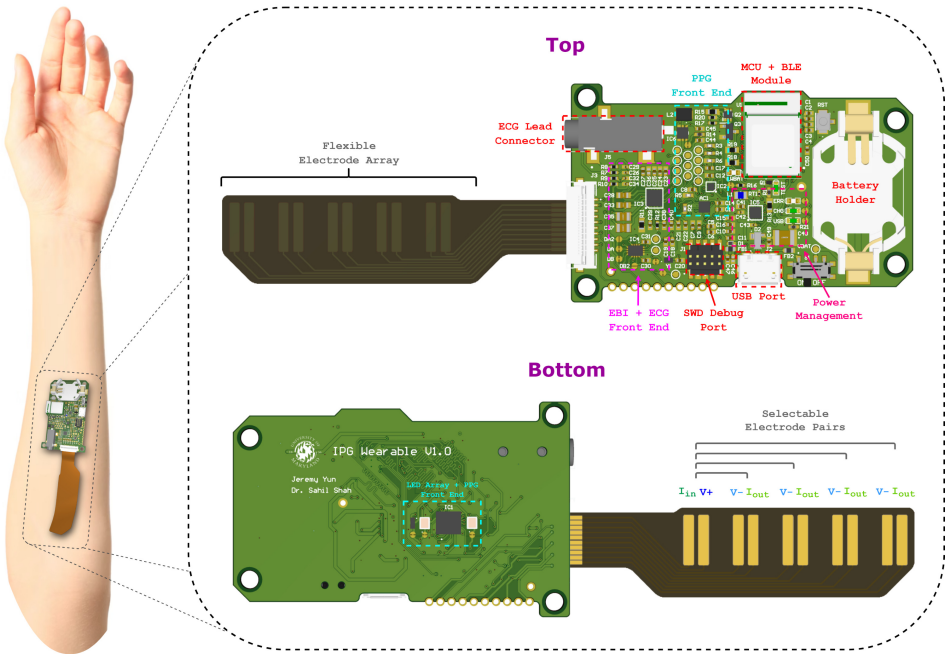

Fig. 3: Custom printed circuit board and flexible electrodes used for measuring the physiological signals.

The assorted wavelengths allow for both heart rate (dual green LEDs) and SpO_2 (Red and IR LEDs) configurations, which can be toggled with a peripheral 4:2 MUX. The reflected PPG light is sensed by two photodiodes integrated on a MAXM86146 Optical Biosensing Module with an internal processor (Analog Devices, Inc, Norwood, MA). A peripheral accelerometer communicates with the MAXM86146 via Serial Peripheral Interface (SPI), enabling motion artifacts to be removed with embedded algorithms in the internal processor. The PPG front end has internal LED drivers with configurable PWM and LED supply current, along with built in ambient light cancellation.

C. Data Acquisition Processing Unit

The device's primary microcontroller unit (MCU) contains an nRF52840 Multiprotocol Bluetooth 5.4 System-on-Chip (SoC) (Nordic Semiconductor ASA, Trondheim, Norway). The MCU is built around a 32-bit ARM M4 Cortex CPU with floating point unit running at 64 MHz and an internal ADC. The MCU also holds 1 MB of internal flash memory and 256 kB of RAM. The nRF SoC is encapsulated within a NINA-B302 Bluetooth Low Energy (BLE) module with an onboard PIFA antenna (u-blox, Thalwil, Switzerland). The primary MCU receives data from the MAX30001 EBI and ECG front end via SPI and the PPG data from the MAXM86146 via the I2C protocol. The MCU pinout configuration is designed for compatibility with an Arduino Nano 33 BLE bootloader and library in an unmodified state to allow for easy embedded software bring-up.

D. Power Management System

All onboard device components run on an energy efficient 1.8V with the exception of the LEDs, which require a 5V step-up boost converter. A highly power efficient battery management system with charging capability allows for long-term continuous usage of the device without battery replacement or frequent charging. The power management system contains an nPM1100 power management IC (PMIC) with an integrated DC/DC Buck regulator configured to 1.8V (Nordic Semiconductor ASA, Trondheim, Norway). The device may be powered either by a 3V single-use CR2032 coin cell battery or a 3.7V LIR2032 rechargeable coin cell. The PMIC contains Li-ion battery charging internals set to a 35mA charge current limit for LIR2032 battery compatibility. The device is rechargeable via Micro-USB if an LIR2032 is used. Custom battery monitoring and temperature tracking circuitry with discrete components are also included. Sampled battery voltage data is processed using the MCU for low battery indication. With all device systems continuously active, including PPG LEDs, the device would consume a maximum load current of $\sim$125mA, depleting a CR2032 battery in 2 hours. However, in most scenarios a few seconds of measurement every few hours is sufficient for continuous monitoring purposes, in which case a typical CR2032 would last over $2\frac{1}{2}$ months, and an LIR2032 would require recharge only every 2 weeks.

IV. Results & Discussion

The platform's biosignal channel modules were individually tested using commercially provided evaluation boards provided by the biosignal front end vendors. For the ECG and EBI data acquisition, a MAX30001EVSYS evaluation platform (Analog Devices, Inc, Norwood, MA) was used to test the MAX30001. Figure 4 illustrates the real-time concurrent acquisition of an ECG and EBI signal collected from a healthy subject. The ECG signal was acquired using a single lead setup with electrodes placed on both forearms. Active

R-to-R detection is demonstrated on the output waveform displayed via a connected GUI. For the EBI demonstration, current with a 16kHz excitation frequency and 8 µA magnitude was injected across the subject's chest and recorded normal respiratory behavior using voltage detect electrode pairs. As the subject inhaled, internal volume of the thoracic cavity increased; during exhale, the volume decreased, registering corresponding changes in measured EBI. As shown in Figure 4, the resulting waveform reflects the captured respiratory cycles. For both cases, all signal filtering, conditioning, and artifact removal was performed onboard the front end.

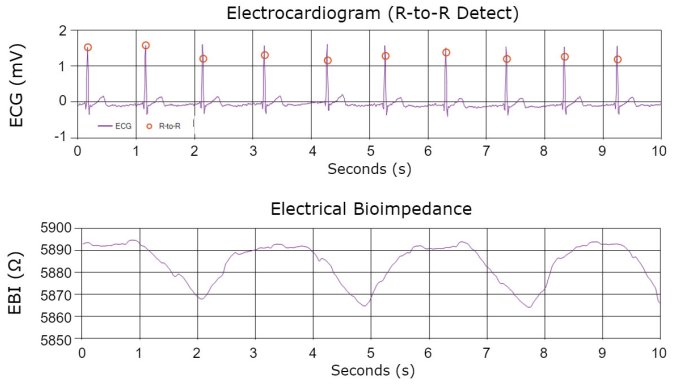

Fig. 4: **(Top)** ECG with real time R-to-R detection using the onboard biopotential front end. **(Bottom)** EBI demonstration with electrodes placed across the thoracic cavity during respiration.

V. Conclusion & Future Work

In this paper, a compact, low-power, open-source, and configurable continuous health monitoring wearable is presented. The platform contains multiple biosignal channels, including ECG, PPG, and EBI, and is capable of onboard processing and wireless transmit. Proof-of-concept efficacy of the acquisition channels are also demonstrated. To our knowledge, such an integrated platform has thus far not been explored in the literature. We envision that this open-hardware platform can be referenced to aid future rapid prototyping of wearable monitors for similar applications. Future steps include further miniaturization, the design of a mechanical packaging, and fabrication on a flexible substrate. Additional directions include the addition of ultrasound-based sensing capabilities and deeper investigations into implementing software algorithms for monitoring specific health applications, such as arteriovenous fistula (AVF) IPG monitoring.

VI. Code

The GitHub repository containing the hardware design files and documentation can be found at: https://github.com/JermYeWorm/Hemodynamic-Monitoring-Wearable.

VII. Acknowledgement

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