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Student's name: **Matthew T Giovanetti**

This work and its defense approved by:

Committee chair: Fred Beyette, Ph.D.

Committee member: Joshua A. Hagen, Ph.D.

Committee member: Amit Bhattacharya, Ph.D.

Committee member: Carla Purdy, Ph.D.

Committee member: Philip Wilsey, Ph.D.

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Physiological Health Assessment and Hazard Monitoring Patch for Firefighters

A dissertation submitted to the Graduate School of the University of Cincinnati
in partial fulfillment of the requirements for the degree of

Doctor of Philosophy

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by

Matthew T. Giovanetti

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Dissertation Advisor and Committee Chair: Dr. Fred R. Beyette Jr.

Abstract

Firefighting is a dangerous profession which has multiple hazards and risks that can result in injuries or death. The objective of this work is to create a novel device capable of monitoring and sending alerts with the comprehensive physiological status of the firefighter when a hazard is imminent or encountered.

The device is a small, lightweight, flexible, and wireless Band-Aid like device that can be attached to the firefighter's skin positioned on the upper arm, preventing the device from interfering with required activities. The device monitors the firefighter's physiological health including heart rate, blood oxygen level, carboxyhemoglobin, skin temperature, and core temperature. Not only will the device monitor the firefighter's physiological health, it will also monitor the environmental temperature and humidity of the firefighter's encapsulated protective suit, and sudden event hazards including burns, impacts, falls, impact from objects, or contact with objects.

The project device integrates multiple sensors together including a reflective pulse oximeter, accelerometer, environmental humidity and temperature sensor, and a contact temperature sensor. Each sensor integrated in the device shows good performance against similar stand-alone commercial devices. The pulse oximeter's heart rate and blood oxygen measurement error is 1.75% and 2.08% respectively, the skin temperature sensor's error is 2.83%, and the environmental humidity and temperature sensor's error is 4.43% for temperature and 7.44% for humidity.

The device integrated each of these sensors into a small 2.25 inches by 4 inches patch, designed to be attached to the upper arm of the subject in order to monitor and wireless report back when physiological health hazards are high or when sudden events occur.

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List of Abbreviations

WASP	Wearable Advanced Sensor Platform
PCB	Printed Circuit Board
HR	Heart Rate
Hb	Hemoglobin
HbO ₂	Oxyhemoglobin
SaO ₂	Arterial Blood Oxygen Saturation
SpO ₂	Bood Oxygen Saturation Level
R	Red Wavelength
IR	Infrared Wavelength
HbCo	Carboxyhemoglobin
CT	Core Temperature
V _t	Variance
C _{vc}	Variance Coefficient
k	Kalman Gain
IDE	Integrated Development Environment
CPU	Central Processing Unit
MIPS	Million Instructions per Second
ADC	Analog-to-Digital Converter
PWM	Pulse Width Modulation
PR _y	Timer Period register
UART	Universal Asynchronous Receiver Transmitter
I ² C	Inter-Integrated Circuit

SpCO	Blood Carbon Monoxide Saturation Level
BMI	Body Mass Index
GUI	Graphical User Interface

1. Introduction

1.1. Firefighter Responsibilities

The occupation of firefighting has multiple responsibilities, beyond fighting forest and building fires, including the following activities:

- Conduct preventive measures by inspecting locations.
- Educating the public on preventing fires.
- Continuous education and training.
- Maintaining the station and equipment.
- Respond and assist with emergencies which includes medical problems.

In order to successfully perform in this occupation maintaining peak physical health, quick decision making and problem solving, and the ability to use learned knowledge and skills is required. Fighting forest and building fires contains the highest hazards and risks which can result in injuries and deaths for the firefighters [1].

1.2. Firefighter's Risks

Workplace hazards and risks, which can lead to injuries and even death, are associated with all occupations. The severity of the injury can be significantly reduced and even death prevented when attention and care is administered immediately after an incident. Firefighters face significant increase of risks and hazards, compared to other jobs, due to the extreme environment and hazardous conditions faced when on active duty.

Firefighters, on active duty, are exposed to high physical demands from both the weight of equipment and conducting tasks for an unpredictable amount of time. Both factors contribute to large energy expenditure during active duty. In addition to the job being physically demanding, the environment leads to heat strain due to the required protective clothing. The protective clothing prevents burns by reducing the heat exchange from inside the clothing and the environment. While protective clothing allows the firefighter to conduct work in extreme environments, the downside includes increased physical strain, decreased work capacity, and heat-induced exhaustion causing thermoregulatory and cardiovascular strains [2].

Environmental conditions, which cause firefighters to experience heat strain, are classified into three categories: routine, emergency, and critical. Each of these conditions allows for limited exposure in order to reduce the risk of injury. Exposure above the predetermined time dramatically increases the chance for injury. Routine environmental condition, which is the safest, is classified as a temperature of 100°C and/or thermal radiation of 1 KW m^2 with an operation time up to 25 minutes of exposure. Emergency environmental condition is classified as a temperature of 160°C and/or thermal radiation of 4 KW m^2 with an operation time up to 1 minutes of exposure. Critical environmental condition, which is extremely dangerous and can damage both equipment and protective clothing, is classified as a temperature of 235°C and/or thermal radiation of 10 KW m^2 with an operation time less than a single minute of exposure [2].

Not only can the environmental conditions lead to burns but can also lead to the firefighter experiencing heat strain and hyperthermia, due to high physical demands, exposure to high temperatures and radiative heat, and partially or fully encapsulated protective clothing. The firefighter's core body temperature has the potential to increase from the normal core body temperature of 98.6°F to a body temperature of 102°F or even as high as 105°F [3].

The average yearly deaths for firefighters is 75 for the past decade in the United States, excluding the deaths caused by and related to the World Trade Center incident in 2001 [4]. Reported deaths were commonly caused by overexertion, stress, sudden cardiac arrests/stroke, caught or trapped by ceiling collapsing, explosions, lost in a burning building, drowning, crushed, struck by object, falling, and heat stroke [5]. Reported injuries were significantly higher at 62,085 injuries, in 2016, for active duty firefighters. Reported injuries were caused by overexertion, slips, falls, jumps, struck by objects, contact with objects, extreme weather, and exposure to chemicals, radiation and fire products [6]. The leading cause for both injuries and death was overexertion, which results in strains, sprains, and muscular pain. Statistical analysis was done on the injuries, reported between the years of 2005 and 2009, into two different categories including minor injuries, defined as no loss of time and requires only treatment of a physician/first aid, and moderate to severe injuries, defined as loss of time and potentially life threatening. The moderate to severe injuries consistent of strains or sprains (34%), thermal burns (10%), and pain (9%) with the cause being from falls, slips, or trips (28%) [7]. Currently there is no devices available which can both monitor these different risks and hazards while also alerting a monitoring personal to request for immediate assistance.

1.3. Health Assessment and Hazard Monitoring Device Motivation

Currently, devices, both on the market and under development, are designed to monitor either firefighters for risks or physiological status of a subject.

1.3.1. Commercial Devices

There are two devices on the market, which are similar to the device reported here. The, Wearable Advanced Sensor Platform (WASP) and the Worcester Polytechnic Institute's

integrated firefighter location and physiological monitor are both intended as device to enhance firefighter safety. Both devices monitor physiological status using similar sensors. However, each of these commercial devices has limitations that are addressed with the device presented in this research.

The WASP is designed to monitor heart rate, heart rate variability, respiration rate, activity levels, posture, and other physiological factors. The device requires a special shirt which holds the monitoring device that is held in place with a fastening strap. A major limitation of the WASP is the requirement of the special shirt which increases the number of clothing layers inside the protective clothing. Further, the WASP device is designed to be attached to the side on the ribcage, which leads to discomfort and can interfere with achieving a full range of motion. Unfortunately, moving the sensor location on the body (including movement of the sensor during wear) leads to inaccurate data due acquisition. This mandates the use of a tight fitting strap to hold the WASP device in place [8].

The Worcester Polytechnic Institute's integrated firefighter location and physiological monitor is designed to monitor heart rate, respiration rate, oxygen saturation level, skin and environment temperature, posture and activity level, and 3-D location. The device consists of a special shirt with sensors on the center of the chest, a head strap, and a radio sized black box. A major limitation, similar to the WASP device, is the requirement of a special shirt, which increases the number of clothing layers inside the protective clothing. In this case there is the addition of a bulky device attached to the chest which limits range of motion. Other limitations include requiring a strap for the forehead which interferes with the required air mask. Further, the strap can dig into the head leading to discomfort and motion artifacts. Beyond these limitation,

requiring the user to carry multiple components (ex. the headband or the radio unit) can lead to device failure if the extra components are dropped or misplaced [9].

1.3.2. Other Research Grade Devices

In addition to the commercially available firefighter monitoring device, there are two devices under development which have similar design goals. The SenseWear Pro and a patch for heat stress monitoring are both smaller devices used to monitor physiological health status with similar sensors. These devices also have limitations which this research is designed to address.

The SenseWear Pro Armband is designed to monitor heartbeat, acceleration, skin temperature, galvanic skin response, ambient temperature, and heat flux allowing the device to calculate the wearer's energy expenditure. The device is an armband which has not been designed for firefighters leading to major limitations. The device measures the subject's physiological health status but does not include any kind of hazard monitoring, warning system, or hazard reporting, because of these limitations the device has limited ability to provide real-time warnings that could reduce injuries or deaths [10].

The patch for heat stress monitoring patch is designed to monitor electrocardiogram, humidity, skin temperature, and acceleration in order to determine the subject's total amount of work done. Based on using a wet bulb globe temperature index the device is designed specifically for monitoring heat stress. By not monitoring other hazards, such as impacts or carbon monoxide exposure, and not having a hazard monitoring or reporting system a firefighter cannot be effectively monitored in real-time for injuries or deaths [11].

There is a device currently under development and designed for firefighter monitoring, which is closest device to the project device [12]. This device is designed for monitoring and reporting the

subject's temperature, heart rate, movement, global positioning system, carbon monoxide concentration of the environment, and location. The device was designed for the sole purpose of monitoring and reporting hazards for firefighters but still has limitations which the device presented here overcomes. The first limitation of the device is measuring the carbon monoxide concentration of the environment, which is not necessary if the firefighter is following procedure of wearing the oxygen mask. A better approach to monitoring the risk of carbon monoxide exposure is to directly measure the effect of carbon monoxide exposure on the subject. The second limitation is the method used for reporting of a "Man Down", which only reports when no movement is detected for a minute, leading to false reports during breaks, rehydrating, and cool down periods. The third limitation is the user feedback system requiring the subject to toggle switches in order to report "Man Down". This approach has limited utility due to potential for accidental activation, forgetting to toggle when appropriate, and unable to toggle due to incapacitation that might result during the incident [12].

1.4. Research Objectives

The objective of this research is to design and develop a device which is capable of monitoring physiological health stats and common hazards and risks which a firefighter faces while on active duty.

The research effort is divided into three aims. The primary aim of the project is to integrate multiple sensors together in a single armband device. Completing this aim requires interfacing all sensors into a single rigid wireless printed circuit board (PCB), development of the firmware for each sensor, development of thresholds for warnings of hazard events, scheduling the sensors to monitor the subject, and testing each sensor against an equivalent commercial device.

The second aim of the project is to transition the device from a rigid armband to a Band-Aid like patch. This aim requires transferring the design from a rigid PCB to a rigid-flex PCB, testing the device against the commercial devices, and attaching layers for protecting the device and attaching the device to the subject. The final aim of the project is to develop a communication protocol to transfer both warning and status updates of the subject's physiological health to monitoring personnel.

1.5. Dissertation Outline

This dissertation presents a device which can monitor the physiological health and hazard exposure of firefighters in order to report back to a monitoring personnel. The sensors used for this device was chosen in order to monitor and detect most of the hazards and risks which are present in the firefighters active duty work environment.

Chapter 1 introduces firefighter job with their job requirements, devices which are both commercially available and under research/development to help firefighters. Finally, this chapter describes the objectives of this dissertation.

Chapter 2 covers the background information for how each sensor used in the device works and why they are important to the overall project goals.

Chapter 3 shows the project device including each selected component and sensor. How the components and sensors are connected to the device, why they were selected, how they communicate to the microcontroller, and how the data can be used to monitor the firefighters.

Chapter 4 reports on the results for each individual sensor and how accurate each one is to a commercially available device. Additionally this chapter will show how the system works over all with human testing.

Chapter 5 concludes with the summary of the dissertation, conclusion of the work, and a discussion of the future possibilities of the work.

2. Literature Review

2.1. Photoplethysmography

2.1.1. Pulse Oximeter Principle

Beer-Lambert law is an important principle used in Pulse Oximeter calculations in order to calculate both heart rate (HR) and blood oxygen level. The law states that the solution's concentration of a desired absorbent can be mathematically calculated by passing light through the solution if the extinction coefficient of the absorbent, path length, and intensity for the light's wavelength are known. Pulse Oximeter's target solution is the subject's blood with the desired absorbent being both hemoglobin (Hb) and Oxyhemoglobin (HbO₂). The extinction coefficient for an infrared wavelength, at 940nm, shows that HbO₂ has more absorbent while Hb is less. The reverse is true for a red wavelength, at 680nm, with Hb absorbing more than HbO₂ but to a more significant degree. Through the use of both Hb and HbO₂ the arterial blood oxygen saturation (SaO₂), referred to as blood oxygen saturation level (SpO₂) when taken through the use of an indirect measurement as with a pulse oximeter, can be approximated [13].

Beer-Lambert law requires all light present to be accounted in the equations. Ideal conditions calls for only one light emitting source present at a time, either the red or infrared for pulse oximeter, and all the emitted light either being transmitted through the target solution or absorbed by the target solution. Unfortunately ideal conditions are nearly impossible to obtain due to the ambient light, tissue, bone, muscle, blood, and electrical interferences with the electronic components. These interferences in the body can lead to light scattering or being reflected making it difficult (or even impossible) to account for all the emitted light. Additionally

the arterial blood, which absorbs the signal, pulses throughout the body creating an AC flow throughout the body [13].

Due to these issues the pulse oximeter would be impossible if not for scaling of the signals. By scaling both the red and infrared signals a comparison can be made in order to find the SpO₂. This is accomplished by scaling the AC by the DC in order to mathematically cancel out the interferences as seen later in Equation 21 [13].

2.1.2. Calculations of Pulse Oximeter

The pulse oximeter is capable of approximating both the HR and SpO₂. HR is derived from the measured AC signal. Accomplished by first locating either the maxima's or minima's time value from the AC signal then through the use of Equation 1, which requires two sequential maxima or minima, the heart rate can be determined.

$$\text{Heart Rate (bpm)} = \frac{60}{\text{Maxima}_2(\text{sec}) - \text{Maxima}_1(\text{sec})} \quad (1)$$

The SpO₂ can be approximated through the use of the ratio between red wavelength (R) and infrared wavelength (IR). R/IR is calculated using both the AC and DC components for both wavelengths, as seen in Equation 2, as seen in Wukitsch [13].

$$\frac{R}{IR} = \left(\frac{\frac{AC_R}{DC_R}}{\frac{AC_{IR}}{DC_{IR}}} \right) \quad (2)$$

The R/IR value is used to determine the SpO₂ through the use of a well-documented graph shown in Figure 1.

Figure 1: Shows the relationship between the red and infrared ratio and the SpO₂. This graph has been reproduced from Wukitsch [13] but can also be found in other pulse oximeter reviews [32] [33] [34].

2.1.3. Process of Pulse CO-Oximeter

Based on the same principles as the normal pulse oximeter the pulse CO-oximeter uses multiple wavelengths to determine the Carboxyhemoglobin (HbCO) [14]. The principle of operation is identical with the primary difference being to evaluate at a wavelength where response to HbCO is dominant.

2.2. Accelerometer

An accelerometer is a sensor capable of measuring linear force in up to three directions. Such sensors can be found in common devices including smart watches and cell phones. Data from an accelerometer is valuable and can be used to estimate both displacement and velocity.

Acceleration, velocity, and displacement data can be used for several different event and activity

recognition algorithms. Activity recognition is limited by sensor placement on the subject, for example brushing teeth while having the recognition sensor located on the hip is not useful for detecting activity associated with teeth brushing. If the sensor was instead placed on the active wrist or arm conducting the brushing then prediction of the activity would be more accurate [15]. Prediction capabilities of the accelerometer can not only predict activities but can also approximate a subject's energy expenditure, which is not dependent on the sensor's location [16]. Another important capability is recognizing and quantifying impacts to a subject and determining if the subject requires attention [17] [18].

2.3. Humidity and Suit Temperature Sensor

The environment's humidity and temperature has important information that can effect multiple systems and both are capable of being measured with a single sensor. Moisture content of the environment, especially the environment inside a firefighter's turnout system, plays a complex role in heat transmission and can lead to injuries to the skin from burns. The time required for a burn to occur, with constant thermal radiation, changes dependent on the moisture content [19].

2.4. Skin Temperature

In order for the human body to achieve thermal equilibrium the skin releases sweat to cool off. Multiple factors, including environment conditions, effect the cooling process for the human body. One method to monitor this process is by monitoring skin temperature. By monitoring the change in skin temperature, hyperthermia can be determined [11]. A simple temperature sensor can be applied to the skin in order to obtain a real time skin temperature measurement.

2.5. Core Temperature

The core temperature (CT) of a subject can be noninvasively approximated through the use of the Kalman filter algorithm, which depends on both a time update model and an observation model. The time update model defines how a variable changes over time, in this case how the CT will change with time. The observation model defines how the main variable changes with respect to a second variable, in this case how CT changes with respect to HR. The approximation of CT requires a series of HR results measured every minute [20].

The approximating CT through the observation of HR model has been developed by Buller [20]. Buller gathered data from a total of 100 test volunteers in order to develop and validate the model. The model requires the simplified Equations 3 through 7 to be calculated for each HR value recorded every minute in order to obtain a new CT approximation, as shown by the following equations [20]:

Based upon previous values the variance (v_t) is estimated, as shown in Equation 3.

$$v_t = v_{t-1} + \text{gamma} \quad (3)$$

Variance coefficient (C_{VC}) is determined using previous CT value, as shown in Equation 4.

$$C_{VC} = 2 * b_2 * CT_{t-1} + b_1 \quad (4)$$

Kalman gain (k) is found using the variance coefficient, as shown in Equation 5.

$$k = \frac{v_t * C_{VC}}{(C_{VC}^2) * v_t + \text{sigma}} \quad (5)$$

The final estimate of the CT is calculated using the HR, Kalman gain, and previous CT value, as shown in Equation 6.

$$CT_t = CT_{t-1} + k * \left(HR - (b_2(CT_{t-1})^2 + b_1 * CT_{t-1} + b_0) \right) \quad (6)$$

Update the variance (v_t) based on variance coefficient, Kalman gain, and first variance value, as shown in Equation 7.

$$v_t = (1 - k * C_{VC}) * v_t \quad (7)$$

The Kalman filter parameters constants are shown in Equation 8.

$$\begin{aligned} \text{gamme} &= 0.022^2; b_0 = -7887.1; b_1 = 384.4286; \\ b_2 &= -4.5714; \text{sigma} = 18.88^2 \end{aligned} \quad (8)$$

Buller showed that, through the use of the Kalman filter model, approximating the CT is possible with observation of the HR. There are multiple factors which affects people's HR such as caffeine, sleep, diet, and stress but these are outweighed by thermoregulation and physical activity. This model is not perfect and other methods, such as rectal and esophageal probes, have shown more accuracy. However the more accurate methods require additional hardware and invasive sensors, which could interfere with the active duty of a firefighter. The Kalman filter model does not require additional hardware and can be easily added requiring only the results from pulse oximeter [20].

3. System Design

3.1. System Level Block Diagram

Figure 2 shows the block diagram of the device and how the results are generated and processed. The green boxes represent the hardware components and the black boxes represent the firmware functions. The accelerometer, suit humidity and temperature, and skin temperature are standalone devices whose results are generated by the sensor. The HR, SpO₂, and SpCO require the LEDs signal sent into the subject and received by the photodiodes which sends the signal to the amplifiers. All the results are then read and processed by the microcontroller before sending out important information through the Bluetooth.

Figure 2: Block diagram of the device (Based on previous publication [36]).

3.2. Microcontroller

The brain of this device is the microcontroller. This component is in charge of controlling every function of the sensors including initializing the sensors, communicating with the sensors to obtain the sensor's data, processing the data into an understandable value, and translating the values into known categories. Not only does the microcontroller control the sensors but is in charge of scheduling the sensors and communicating the status of the subject to monitoring personnel. The selected microcontroller is a general purpose microcontroller from the PIC24F family from Microchip Technology Inc. This was chosen due to the wide variety of peripheral features and large memory. The main attraction of this microcontroller is the elimination of additional hardware through the use of a single component without the need of additional modules for sensor communication or addition memory storage. MPLABX Integrated Development Environment (IDE) is used for programming Microchip's microcontroller. This IDE allows developing the firmware, using the high level C language through the support of the XC16 compiler, and programming the microcontroller, using the PICkit 2 development programmer and debugger [21].

The microcontroller used for this device is the PIC24FJ128GA010 which is 16-bit flash microcontroller with a high performance Central Processing Unit (CPU). The CPU is a modified Harvard Architecture with up to 16 Million Instructions per Second (MIPS) at 32MHz. This is obtained through the use of the internal oscillator at 8MHz with four phase locked loops (PLL) and multiple divided operations, 17 bit by 17 bit single cycle hardware multiplier, 32 bit by 16 bit hardware divider, and 16 by 16 bit working register array. The C compiler optimizes the instruction set and uses two address generation units for separate Read and Write addressing of the data memory [21].

3.2.1. Oscillator Configuration

The PIC24FJ128GA010 allows multiple clock frequencies to be tailored to the application. The oscillator functions as the source for the clock signal. The oscillator can be either internal or external and used with the 4x PLL to boost the operating clock frequency. The clocking can be changed postscaler for selective clocking for system power saving with the CPU. Four different sources can be used for the systems clock [21].

1. Primary Oscillator uses a crystal attached between pins OSC1 and OSC2.
2. Secondary Oscillator uses a crystal attached between pins SOSCI and SOSCO.
3. Fast Internal RC Oscillator built inside the microcontroller.
4. Low-Power Internal RC Oscillator built inside the microcontroller.

Only two sources, primary oscillator and FRC, can use the four internal PLLs. FRC clock can also use a clock divider in order to reduce the frequency. These different clock sources is used for not only the processor but also the peripheral clock sources. The internal instruction cycle clock is created by dividing the processor clock in two. Three registers are used in order to configure the operation of the oscillator [21].

1. Oscillator Control register controls and monitors the clock sources.
2. Clock Divider register controls the postscaler used with FRC oscillator.
3. Oscillator Tune register finely tunes the FRC oscillator frequency.

3.2.2. Pin Description

The PIC24FJ128GA010 used in this device is a 100 pin microcontroller with multiple peripherals. Figure 3 shows the pin diagram of the microcontroller with a color coding. The color

coding shows which pins are needed for the device, including the sensors, Bluetooth, power, ground, and microcontroller required components. Table 1 explains the color coding used in Figure 3 by listing each pin used for the device and provides a description of the function of the pin [21].

Figure 3: Pin diagram of the PIC24FJ128GA010. The color coding indicated the pin connections to the various other components. This image was taken from Microchip Technologies Inc. [21] but modified with color coding.

Table 1: Pin Description table for each pin used for the PIC24FJ128GA010

Color Code	Pin #	Description
	13	Master Clear used when programming
	26	Programming Clock
	27	Programming Data
	63	Oscillator Clock
	64	Oscillator Clock
	24	Analog input for DC Single
	25	Analog input for AC Single
	58	Digital Output for Red LED
	59	Digital Output for IR LED
	60	Switch to turn on Pulse <u>COx</u> circuit
	61	Digital Output for CO LED
	72	Offset Signal for DC circuit
	39	UART RTS
	40	UART CTS
	49	UART RX
	50	UART TX
	22	Analog input
	56	I ² C Data
	57	I ² C Clock
	66	Interrupt for Impact
	67	Interrupt for Inactivity

3.2.3. I/O Ports

Most the PIC24FJ128GA010 pins are shared between the peripherals and parallel I/O ports. The only pins which are not are the V_{DD} , V_{SS} , MCLR, and OSC1/CLK1 due to these pins being needed for the microcontroller to function. Schmitt Trigger inputs are used on all the I/O input ports in order to improve the noise immunity. A pin which is share between the I/O port and a peripheral is controlled through the use of multiplexers. One of the shared options will be allowed to be enable at a time with the other disable, in order to prevents both I/O port and a

peripheral sharing a pin to operate at the same time such as a ports digital output from driving the input of the peripheral [21].

All I/O ports have three important registers that determine their operation. These three registers are TRISx, LATx, and PORTx. TRISx is the Data Direction register in which determines is the pin is an input or output. LATx is the latch location the used for reading and writing. This location is used in order to prevent any reads or writes which are not allowed. PORTx are for the port pins. Reading is allowed with the port pins while writing needs to be done to the latch which later writes to the port if allowed [21].

3.2.4. Analog-to-Digital Converter (ADC)

The PIC24FJ128GA010 contains multiple 10-bit Analog-to-Digital Converters (ADC). The ADC on for this microcontroller has the following key features [21]:

- Successive Approximation conversion
- Conversion speeds of up to 500 ksps
- 16 analog input pins
- External voltage reference input pins
- Automatic Channel Scan mode
- Selectable conversion trigger source
- 16-word conversion result buffer
- Selectable Buffer Fill modes
- Four result alignment options
- Operation during CPU Sleep and Idle modes

The ADC allows the sampling of an analog voltage signal up to VDD which will translate to a decimal value of 1023 [21].

3.2.5. Timers

The PIC24FJ128GA010 has five timers each can be set to different frequencies. Timer 1 is a 16-bit timer while Timers 2 through 5 can be used as either independent 16-bit timers or 32-bit timers when combining Timers 2 and 3 or Timers 4 and 5. These timers have features for gate operation, selectable prescaler, and operating when in idle or sleep mode. Timer 1 can be configured either as synchronous or asynchronous counter. Time 1 also allows the interrupting on external gate signal or 16-bit period register match. Timer 2/3 and 4/5 only support synchronous counter and 32-bit period register match interruption. Only Timer 2/3 allows for ADC event trigger [21].

3.2.6. Interrupts

The PIC24FJ128GA010 has both internal and external interrupts. Interrupt sources can be set up to 118 different sources which are controlled by the Interrupt Vector Table. These interrupts have seven priority levels, configured by the program, in order to determine which goes first. Besides the priority level the interrupts are controlled by the flag status bit, shows if the interrupt has occurred and needs attention and should be cleared after processed, and enable bit, says that the interrupt is turned on or off [21].

3.2.7. Pulse Width Modulation (PWM) Module

The PIC24FJ128GA010 has a Pulse Width Modulation (PWM) which allows the generation of a square wave signal. This signal can be customized by setting both the period and duty cycle. The

PWM period is configured by writing to the Timer Period register (PRy), is set by using Equation 9 [21].

$$\begin{aligned} PWM \text{ Period} &= 1 / PWM \text{ Frequency} \\ &= (PRy + 1) * TCY * (Timer \text{ Prescaler Value}) \end{aligned} \quad (9)$$

The PWM duty cycle is configured by writing to the OCxRS register. The duty cycle can be changed at any time during the programming by writing to this register, which will latch at the end of the PWM period. The register value is set by Equation 10. If the register is set to 0 then the signal will remain low, 0% duty cycle, while on the other hand if the register is set greater than the PRy value then the signal will remain high, 100% duty cycle [21].

$$PWM \text{ Duty Cycle} = (Duty \text{ Cycle Percentage} * (PRy + 1)) / 100 \quad (10)$$

The resolution at which the duty cycle can be set is determined by the frequency at which the PWM is set, as shown by Equation 11 [21].

$$\begin{aligned} &Maximum \text{ PWM Resolution (bits)} \\ &= \frac{\log_{10} \left(F_{CY} / F_{PWM} * (Timer \text{ Prescale Value}) \right)}{\log_{10}(2)} \end{aligned} \quad (11)$$

3.2.8. UART Module

The PIC24FJ128GA010 has multiple serial communication modules including Universal Asynchronous Receiver Transmitter (UART). UART is an asynchronous system that is used for communicating information with other peripheral devices. UART contains multiple features to help with the communication including [21]:

- Full-Duplex, UxTX and UxRX pins allow 8 or 9-bit data transmission.
- Optional, only for 8-bit data, Even, Odd, or No Parity.

- Stop bits that is one or two long.
- Optional UxCTS and UxRTS pins allowing hardware flow control
- Generating of a fully integrated baud rate with the 16-bit prescaler.
- Baud rates can range from a minimum of 1Mbps to a maximum of 15bps at 16 MIPS.
- Transmit and Receive Data Buffer that is 4-deep First-In-First-Out (FIFO).
- Interrupts triggering from either transmit or receive.
- Diagnostic support through the use of loopback mode.
- Break characters and support fir sync.
- Automatic baud rate detection support.
- IrDA logic for encoding and decoding.
- IrDA support requiring a 16x baud clock output.

The baud rate is the speed at which the UART module can communicate. The 16-bit baud rate generator is preset by the UBRGx register and is controlled by a 16-bit free-running timer. This register is set by the Equation 12 [21].

$$UBRGx = \frac{F_{CY}}{16 * Baud Rate} - 1 \quad (12)$$

There are four main pins that are used for the UART communication:

- UxTX Transmit pin used for sending the data from the microcontroller to the peripheral.
- UxRX Receiver pin used for receiving the data from the peripheral.
- UxCTS Clear-to-Send sends an interrupt to the peripheral alerting that data is ready to be received.
- UxRTS Request-to-Send sends an interrupt to the peripheral alerting that data wants to be sent.

3.2.9. I²C Module

Inter-Integrated Circuit (I²C) is another serial communication module supported by the PIC24FJ128GA010. I²C is a master slave communication method allowing multiple peripherals to be attached to the same I²C pins. I²C contains multiple features to help with communication including [21]:

- Logic for independent master and slaves.
- Addresses for 7 and 10-bit devices.
- General addressing.
- Built in delays to the clock allowing for the microcontroller to respond to a slave request.
- Bus specification for both 100kHz and 400kHz.
- Address masking that can be configured.
- Multiple master modes in order to prevent the loss of messages when in arbitration.
- When set as a slave the microcontroller can be set to bus repeater mode, allowing accepting all messages regardless of the address.
- Automatic SCL.

When an I²C device is set to master mode the device is in control of the communication between the master and all slave devices. To start the communication process the master sends the Start condition which alerts all slaves that the master needs attention. In order for the master to select the correct slave an address is sent and the slave with the matching address sends back an Acknowledge command to indicate the message has been received and waiting a command. Each device can have a different command protocol and different commands. After the command is received and all acknowledgements have been received a Stop condition is generated. Similar to

UART, I²C requires a baud rate generator to operate. The baud rate is set by writing to the I2CxBRG register using Equation 13 [21].

$$I2CxBRG = \frac{F_{CY}}{F_{SCL}} - \frac{F_{CY}}{10,000,000} - 1 \quad (13)$$

3.2.10. Programming the PIC

In order to program the PIC24FJ128GA010 a couple requirements are needed including IDE, compiler, programmer/debugger, and the microcontroller with the programming pins correctly connected. There are a couple different methods that can be used to program the different PICs, this document will cover the method used for the device talked about in this document. The IDE used for this microcontroller is MPLAB X V2.00 IDE which is provided free from Microchip

Figure 4: The circuitry need to program the PIC24FJ128GA010. The black arrow lines up with the arrow on the PICKit2.

Technology Inc. This IDE requires a compiler to translate the program into firmware for the microcontroller. The compiler used is the MPLAB XC16 in order to code the program using the high level C programming language. This C program will be translated with the compiler into a firmware to be loaded into the microcontroller. Using the PICkit 2 connected to the microcontroller, as shown in Figure 4, the PIC can be loaded with the firmware.

3.3. Photoplethysmography Signal

This section covers the hardware and software used for the Pulse Oximeter and Pulse CO Oximeter. The theory of the both the Pulse Oximeter and Pulse CO Oximeter can be seen in 2.1. Photoplethysmography.

3.3.1. LEDs

Three LEDs were used for operation for both the Pulse Oximeter and Pulse CO Oximeter. Each LED was selected based off its wavelength. The packaging of the LEDs allowed the light to be directed into the skin and not shining light through the sides, there for reducing the amount of light detected before entering the skin. The Pulse Oximeter LEDs were selected based on the theory covered in 2.1.2. Calculations of Pulse Oximeter. The wavelength for the Red LED is 645nm and IR LED is 940nm. The pulse CO-oximeter uses the pulse oximeter LEDs with one additional LED, from now on will be referred to as Other LED. The Other LED was selected to be 830nm. This wavelength was selected because it is near the point which both Hb and HbO₂ have similar extinction levels. The pulse oximeter finds the both Hb and HbO₂ allowing the HbCO more easily derived from each LED results. These LEDs are controlled by the microcontroller's digital output pins.

Figure 5: The spectral sensitivity of the photodiodes used on the device. The lines are showing the location of each LED lined up on the graph with red for the Red LED, green for the Other LED, and purple for the IR LED. This graph was taken from Fairchild but modified with the red, green, and purple lines [35].

3.3.2. Photodiodes

The selection of the photodiode was based off the ability to detect all three LEDs' wavelengths, as shown in Figure 5. In addition the photodiode's package allows a wide angle of detection allowing more reflected light to be detected.

Figure 6: Shows the AC waveform traveling from the photodiodes on the left to the Microcontroller on the right with each step of the amplifier and filter in between.

3.3.3. AC Signal

3.3.3.1. AC Circuit

The AC component is used for deriving HR, SpO₂, and blood carbon monoxide saturation level (SpCO). The AC component is generated from the photodiodes then is amplified and filtered before being recorded by the microcontroller, as shown by Figure 6.

Figure 7: High Pass Filter.

Noise reduction through filtering and amplification of the signal is key to obtaining a readable signal by the microcontroller. The AC signal from the photodiodes contains noise as large as or larger than the desired signal, therefore multiple filters and amplifiers are required. A high pass filter, as shown in Figure 7, allows for blocking frequencies lower than the targeted corner frequency. Six high pass filters were used to reduce the low frequency noise and block the DC signal from entering the AC circuitry through the use of the capacitor. The cut off frequency for all of the high pass filters is 0.5Hz. The reason that this cut off frequency was chosen is to filter out a HR of less than 30BPM.

An active low pass filter, as shown in Figure 9, is used for both filtering the higher frequencies and amplifying the AC signal. The cut off frequency for the low pass filter used in the circuitry is 3.4Hz. The reason for this frequency is to filter out a HR of greater than 204 BPM. The gain of the active filter used is -100, as calculated using Equation 14.

$$Gain = -R_2/R_1 \quad (14)$$

Figure 9: Active Low Pass Filter. Can be used for filtering and amplification.

Inverting amplifier, as shown in Figure 8, is used to further signal amplification. This amplifier allows not only the signal to be increase but also inverts the signal in order to get the signal out of the negative voltage into the positive range. The gain of the active filter used is -10, as calculated using Equation 14.

Figure 8: Inverting Amplifier

Figure 10: Non-Inverting Amplifier

Non-inverting amplifier, as shown in Figure 10, is used to amplify the signal. Unlike the inverting amplifier this amplifier does not invert the signal. The gain of the active filter used is 201, as calculated using Equation 15.

$$Gain = 1 + (R_2/R_1) \quad (15)$$

The final part of the AC circuit is the inverting level shifter, as shown in Figure 11. This circuit is a combination of a basic differential amplifier with a gain of 1, such that no amplification, and a voltage divider, in order to get a uniform voltage increase by supplying a constant DC voltage for which the signal is subtracted. This constant voltage level shift for this circuit is increase by 2V, as caluated with Equation 16.

$$Voltage\ level\ shift = \frac{R_2}{R_1 + R_2} V_{DD} \quad (16)$$

Figure 11: Inverting Level Shifter

Figure 12: The amplification and filtering circuit used for the AC signal.

The AC circuitry is a combination of these filters and amplifiers, as shown in Figure 12, along with two Schottky diodes, in order to protect the microcontroller from negative voltage.

3.3.3.2. Box Car Averaging

Even after filtering the signal with the hardware, the signal still can show up with some noise. In order to further reduce the noise Box Car Averaging is implemented in the software. Box Car Averaging involves averaging the current data point with a number of data points before and after the current data point, which will be referred to as N . This allows the sharp changes caused by noise to be smoothed out. The higher the N number the smoother the line becomes, this can also negatively affect the data if N is too high. Figure 13 shows how data containing noise is affected by Box Car Averaging, with variable N values of 1, 2, and 3. For clarification Three Point is when $N=1$ because a total of three data points are averaged together.

Figure 13: The results found by varying the number of points included in the box-car averaging to reduce noise.

3.3.3.3. *Maxima and Minima Detection*

After the filtering is complete the signal is now ready to be processed. Processing the AC signal requires detection of all the maxima and minima of the waveform. The data from these locations will be used for calculating both the HR and SpO₂.

The algorithm traverses the data by starting at the beginning and steps forward one data point at a time until it reaches the end. First the algorithm searches for a minima, this is accomplished by storing the first value then checking if the next value is lower, if lower than the stored value is replaced. This process is repeated until higher values are observed. Since anomalies are in the data and the data is not a perfect sine wave a threshold is required to know when the minima is detected. The threshold starts as a set number but is updated as maxima and minima are detected by being set as one forth the difference between the previous maxima and minima. If the current value minus the threshold is greater than the lowest stored value then the microcontroller saves

the minima and starts to locate a maxima. The maxima is the same process as the minima except the algorithm searches for the highest value and the threshold must be added and less than the stored highest value.

This algorithm goes back and forth between detection the minima and maxima until the end of the data has been reached. Both the relative time value and ADC value are stored each time a maxima and minima are detected. An example of the final result for this process is shown in Figure 14.

Figure 14: A snapshot of the results of the microcontroller locating the maxima and minima in the AC signal. The blue line is the waveform reconstructed by the microcontroller from the sensor. The green and red points are the maxima and minima, respectively, located by the microcontroller using the algorithm.

3.3.4. DC Circuit

The DC component used for deriving the SpO₂ and SpCO and contains less noise than the AC component. The DC component is generated from the photodiodes, processed through a differential amplifier along with a PWM signal generated by the microcontroller, and recorded by the microcontroller as shown by Figure 15.

Figure 15: Shows the DC waveform traveling from the photodiodes on the left to the Microcontroller on the right with an amplifier between.

Though the use of a PWM signal and a simple parallel RC Smoother circuit, which contains a 22uF capacitor and 10kΩ resistor, the microcontroller generates a variable voltage signal, which can be set between ground and VDD. This variable voltage signal is adjusted by setting the duty cycle of the PWM. Figure 16 shows the relation between setting the duty cycle and the expected

Figure 16: Shows the relationship between the decimal levels of the microcontroller's ADC against the Duty Cycle percentage of the PWM.

input into the differential amplifier, obtained by sending the PWM into the RC Smoother circuit then read by the microcontroller's ADC.

Both the photodiode and variable voltage signal are used with a differential amplifier, as shown in Figure 17, in order to amplify the DC component while restricting the signal between the rails of the amplifier. The differential amplifier's output voltage is determined through the use of Equation 17.

Figure 17: Differential Amplifier used to determine the DC component.

$$V_{out} = \frac{R_2}{R_1} (V_{Photodiodes} - V_{PWM}) \quad (17)$$

Equation 18 shows how the DC values is calculated, with the $ADC_{Average\ DC}$ being multiple ADC reading average together from the differential amplifier and the $ADC_{Duty\ Cycle}$ being the duty cycle's ADC value located from Figure 16.

$$DC\ Component = ADC_{Average\ DC} + 19(ADC_{Duty\ Cycle}) \quad (18)$$

3.3.5. Heart Rate Calculation

HR can be calculated using the AC component from any of the three LED signals, either the maxima or minima time values are required. Equation 1 shows the calculation required to find the HR, for simplicity the equation will only be shown using maxima but can be replaced with

minima values. All HR values calculated during a single gathering cycle are averaged together to obtain a more stable result. Any outlier HR values, which are determined by either being not possible or too far away from previous results, are thrown away to compensate if an unstable AC waveform was recorded.

3.3.6. SpO₂ Calculation

SpO₂ is found using the relationship between Equation 20 and Figure 1. Equation 20 shows how the undesired variables, such as difference in sensitivity of photodiodes between each LED, are cancelled out, through scaling of AC by DC and R by IR. Equation 20 shows how the R/IR ratio is calculated from use of both Red and IR readings, which includes the AC and the DC components. The AC component is the peak-to-peak value of the waveform, difference between the maxima and minima, as shown by Equation 19.

$$AC\ Component = Maxima_1 - Minima_1 \quad (19)$$

Since the waveform is not a perfect sin wave the AC component is an average of each peak-to-peak value with the outliers thrown away. The DC component is given by Equation 18.

S = Sensitivity of Photodiode; G = Gain of the amplifiers

$$\begin{aligned}
 \frac{R}{IR} &= \left(\frac{AC_R}{DC_R} / \frac{AC_{IR}}{DC_{IR}} \right) = \left(\frac{\left(\frac{S_R (V_{PPR_AC} / G_{AC})}{S_R \left(\left((V_{DCR} - V_{DCR_Offset}) / G_{DC} \right) + V_{DCR_Offset} \right)} \right)}{\left(\frac{S_{IR} (V_{PPIR_AC} / G_{AC})}{S_{IR} \left(\left((V_{DCIR} - V_{DCR_Offset}) / G_{DC} \right) + V_{DCIR_Offset} \right)} \right)} \right) = \\
 &= \left(\frac{\left(\frac{S_R}{S_{IR}} \right)}{\left(\frac{S_{IR}}{S_{IR}} \right)} \right) \left(\frac{\left(\frac{V_{PPR_AC} / G_{AC}}{\left((V_{DCR} - V_{DCR_Offset}) / G_{DC} \right) + V_{DCR_Offset}} \right)}{\left(\frac{V_{PPIR_AC} / G_{AC}}{\left((V_{DCIR} - V_{DCR_Offset}) / G_{DC} \right) + V_{DCIR_Offset}} \right)} \right) \\
 &= \left(\frac{V_{PPR_AC} / G_{AC}}{\left(\frac{(V_{DCR} - V_{DCR_Offset})}{G_{DC}} \right) + V_{DCR_Offset}} \right) \left(\frac{\left(\frac{(V_{DCIR} - V_{DCR_Offset})}{G_{DC}} \right) + V_{DCIR_Offset}}{V_{PPIR_AC} / G_{AC}} \right) \quad (20) \\
 &= \left(\frac{V_{PPR_AC} / G_{AC}}{V_{PPIR_AC} / G_{AC}} \right) \left(\frac{\left(\frac{(V_{DCIR} - V_{DCR_Offset})}{G_{DC}} \right) + V_{DCIR_Offset}}{\left(\frac{(V_{DCR} - V_{DCR_Offset})}{G_{DC}} \right) + V_{DCR_Offset}} \right) \\
 &= \left(\frac{\left(\frac{1}{G_{AC}} \right)}{\left(\frac{1}{G_{AC}} \right)} \right) \left(\frac{V_{PPR_AC}}{V_{PPIR_AC}} \right) \left(\frac{(V_{DCIR} - V_{DCR_Offset}) + G_{DC} (V_{DCIR_Offset})}{(V_{DCR} - V_{DCR_Offset}) + G_{DC} (V_{DCR_Offset})} \right) \\
 &= \left(\frac{V_{PPR_AC}}{V_{PPIR_AC}} \right) \left(\frac{(V_{DCIR} - V_{DCR_Offset}) + G_{DC} (V_{DCIR_Offset})}{(V_{DCR} - V_{DCR_Offset}) + G_{DC} (V_{DCR_Offset})} \right)
 \end{aligned}$$

The R/IR ratio requires the use of Figure 1 in order to approximate the SpO₂. Unfortunately figures cannot be programmed into a microcontroller without an equation. The equation for Figure 1 was not freely available, as such the equation had to be derived from the graph. Through

the use of excel the approximated equation for the relationship between R/IR and SpO₂ was determined, as shown by Equation 21, allowing the microcontroller to calculate the SpO₂.

$$SpO_2 = -3.1796 \left(\frac{R}{IR} \right)^2 - 21.152 \left(\frac{R}{IR} \right) + 109.02 \quad (21)$$

SpO₂ is found by recording and processing results from a single gathering cycle. A single gathering cycle, for the pulse oximeter, involves sampling the AC waveform, for both Red and IR, once every 10 millisecond for ten seconds, allowing multiple peaks, then sample the DC waveform, for both Red and IR, once every 10 millisecond for half a second, due to multiple tests showing that recording for half a second or ten seconds leads to the similar result. Figure 18 shows a single SpO₂ gathering cycle with the calculated variables and results displayed.

Figure 18: A single gathering cycle of the device (Based on previous publication [36]).

3.4. Accelerometer

The 3-axis accelerometer selected for the device is the ADXL345; due to the ultralow power capable of 23μA while active and 0.1μA in standby mode, high resolution up to 13-bits, force

measurements up to $\pm 16g$, and multiple interrupt capabilities including both activity and inactivity monitoring [22].

The ADXL345 communicates with the microcontroller through I²C protocol. The I²C is configured to run at 100 kHz for 7-bit addressing. The accelerometer is the slave in the communication protocol with an address of 0x1D with the microcontroller as the master. The exact read and writing protocol used for the accelerometer is shown in the datasheet, through the use of these protocols and the register the microcontroller can configure and access the data of the accelerometer [22].

The main registers for configuring the accelerometer is both the BW_RATE, DATA_FORMAT, and FIFO_CTL. BW_RATE controls the communication speed which is set for 100 kHz. DATA_FORMAT controls multiple variables including setting the accelerometer to I2C, resolution mode which is set for the full resolution of 4 mg/LSB, justify which is set to right-justified mode with sign extension, and the force range which is set to $\pm 16g$. FIFO_CTL controls how the accelerometer gather, stores, and handles the data. The data is set to trigger mode, gathers a data point before trigger event and fill an array of 30 data points [22].

The trigger event, which informs the accelerometer to gather data, is set by the activity interrupt. The activity interrupt allows the detecting of falls and impacts by alerting the microcontroller when the accelerometer detects a force above the set threshold on any axis, which can be set at increments of 62.5 mg/LSB. The threshold is set to 3 g, this value was chosen in order to detect any impact above the normal movement impacts.

After the trigger event occurs the data is required to be read before another trigger event can occur. In order to read the data a multiple byte read is required of registers 0x32 to 0x37 in order

for no data to be lost. After the register is read the data is deleted and the next data set in the array is loaded. As such, after a trigger event occurs, these register are read, in a loop, 30 times in order to empty the filled data array. The microcontroller will process each data set by calculating the impact force through the use of Equation 22, while storing the maximum impact force calculated. Once the microcontroller finished processing the impact forces an alert is sent to the supervisor with the stored value.

$$\begin{aligned} & \text{Impact force}(mg) \\ & = \sqrt{(Data\ X * 4mg/LSB)^2 + (Data\ Y * 4mg/LSB)^2 + (Data\ Z * 4mg/LSB)^2} \end{aligned} \quad (22)$$

Another interrupt, which is not a trigger event, which alerts the microcontroller is inactivity. Unlike the previous interrupt, the inactivity interrupt requires readings under a set threshold for a set amount of time before interrupting the microcontroller. The threshold force can be set at increments of 62.5 mg/LSB and is set a little above 1 g, in order to account for gravitational force. The threshold time can be set at increments of 1 sec/LSB and is set to four minutes.

3.5. Humidity and Suit Temperature Sensor

The combination digital humidity and temperature sensor selected in this device is the HIH8120; due to the low power of 650 μ A while measuring and 1 μ A in sleep mode, high resolution of 14-bits for both humidity and temperature, wide temperature and humidity range of 125°C and 100% respectively, and built in hydrophobic filter allowing protection in multiple environmental conditions [23].

The HIH8120 communication is similar to the accelerometer using the I²C protocol and same pins. The I²C is configured to run at 100 kHz for 7-bit addressing. The HIH8120 is the slave in the communication protocol with an address of 0x27 with the microcontroller as the master. The

microcontroller can only read the humidity and temperature data. Unlike the accelerometer the HIH8120 does not have any configurable registers [24].

Reading the values from the HIH8120 requires four data bytes to be read. These data bytes contain the status of the data, the humidity data, and the temperature data. Each of these are required to be processed in order to acquire and verify the data [24].

First is the Status Bits which are used to ensure that the data is current. The Status Bits are processed by bit shifting the data by six places to the right, as shown in Equation 23. Only three statuses are used for the HIH8120, which are Normal Operation, Stale Data, and Command Mode. Normal Operation status indicates valid data, which is data that has not been read. Stale Data status indicates repeated data, which is data that was already read. Command Mode status indicates that the HIH8120 is not in normal operation mode and therefore the data is invalid [24].

$$\text{Status Bits} = \text{Data Byte 1} \gg 6 \quad (23)$$

Once the valid data is retrieved, by the microcontroller, processing begins. First the temperature requires combining Data Byte 3 and 4 together, as shown by Equation 24, then calculate the temperature through the use of Equation 25 in order to have the temperature value in degrees Celsius [24].

$$\text{Temperature Data} = (\text{Data Byte 3} \ll 6) | (\text{Data Byte 4} \gg 2) \quad (24)$$

$$\text{Temperature } (^{\circ}\text{C}) = 165 \left(\frac{\text{Temperature Data}}{(2^{14} - 2)} \right) - 40 \quad (25)$$

Second the humidity requires combining Data Byte 1 and 2 together, as shown by Equation 26, then calculate the humidity through the use of Equation 27 [24].

$$\text{Humidity Data} = (\text{Data Byte 1} \ll 8) | (\text{Data Byte 2}) \quad (26)$$

$$Humidity (\%RH) = \left(\frac{Humidity\ Data}{(2^{14} - 2)} \right) (100\%) \quad (27)$$

3.6. Skin Temperature

The linear active thermistor selected for this device is the MCP9701; due to the low power of 6 μ A, high resolution of 19.5mV/ $^{\circ}$ C, and a wide temperature measurement range up to 125 $^{\circ}$ C. MCP9701 has a linear output response to the temperature measurement [25].

The active thermistor uses ADC to measure the temperature, allowing for single pin to read the voltage from the thermistor. Equation 28 shows the calculation required to find the temperature from the ADC value.

$$Temperature (^{\circ}C) = \frac{\left(\left(3.6 \left(\frac{Temperature_{ADC}}{1024} \right) \right) - 400mV \right)}{19.5\ mV/^{\circ}C} \quad (28)$$

3.7. Bluetooth

The Bluetooth module selected for this device is the bluegiga BLE112 Bluetooth V4.0 due to the low power of 27mA when transmitting and 0.4 μ A when in sleep mode. The Bluetooth module communicates with the microcontroller through a UART protocol [26].

3.8. Power Supply

Each component requires power requiring both power source, in this case a battery, and a voltage regulator, in order to stabilize the voltage. In order to simplify the design and reduce the device

Figure 19: The voltage requirements for each peripheral used in the device. All components share a voltage range between 3.5 and 3.6 volts.

size a single voltage regulator is used. Figure 19 shows the voltage requirements for each component, the common voltage for all components is between 3.5 and 3.6 volts.

The voltage regulator, which powers all components, selected is the TPS63070. This high efficiency and low quiescent current regulator has a variable output voltage, from 2 to 9 Volts, buck-boost converter with 3.6 amps switching current [27].

3.9. Schematic of Circuit Board

Figure 20 through Figure 23 shows the schematic, which includes all electrical connections, for the device.

Figure 24 through Figure 26 shows the footprint of the device. The device contains four layers, two rigid and two flexible. The rigid layers, which the components are attached, sandwich the flexible boards, which holds the rigid board electrically and physically together.

Microcontroller

Figure 21: The schematic of the microcontroller and required components for the microcontroller.

Transmit Bluetooth

Programmer

Figure 20: The schematic of the transmission of data circuitry including the Bluetooth and microcontroller programmer.

Figure 24: The top layer of the device. This layer is four separate rigid boards which holds the components.

Figure 25: The top layer of the flexible interconnection layer connecting the rigid boards.

Figure 27: The bottom layer of the flexible interconnection layer connecting the rigid boards.

Figure 26: The bottom layer of the device. This layer is four separate rigid boards which hold the components.

3.10. Program Flow

This section contains an abstract program flow of the firmware. More details on equations, algorithms, and threshold values can be seen in their appropriate sections.

The main function, as shown in Figure 28, for the device starts out by initializing the peripherals then jumps into the main loop which will loop forever. The main loop starts by gathering the pulse oximeter results then calculates the HR and SpO₂ of the subject, if the SpO₂ is below 90% an alert is issued. Afterwards the SpCO is calculated and determined if the percentage has risen, if in an upward trend is occurring then an alert is issued. Next the CT of the subject is calculated, if above the threshold for hyperthermia to occur then an alert is issued. Skin temperature is measured next, if conditions for a burns to occur then an alert is issued. Finally the suit temperature and humidity is determined, if the environmental conditions are in the danger zone for burns then an alert is issued.

Each alert is accompanied by the current status information which triggered the alert, for example if the SpO₂ is below 90% then the alert would send the following message “ALERT: SpO₂ is at ##%”. As such the alerts are only a guide designed to be read by the monitoring personnel who will be responsible for deciding if additional attention is required for the subject. The alerts are designed as such due to the fact that people are more likely to listen to the chain of command than a device, in addition the subject might be too busy on active duty to interact with a device.

Figure 28: The main function of the device.

Figure 29: The order in which the device gathers Pulse Ox Results.

multiple sets, as shown in Figure 29. The first step required gathering the AC values for the red LED and using the algorithm to find each maxima and minima contained in the AC waveform. Next the red DC signal is read and calculated. This process is then repeated for the IR led. The important information gathered during this process is AC components, both the average peak to peak value and frequency, and

The device initialized the peripherals, as shown in Figure 30. The order only matters with accelerometer due to the use of I²C required for communication. The pulse oximeter requires both the ADC and digital initialization, while the skin temperature requires only the ADC initialization. The Bluetooth requires UART initialization. Both the accelerometer and suit temperature and humidity sensor requires I²C initialization. Finally the Accelerometer's interrupts are set before the main loop can occur.

Gathering pulse oximeter results involves

Figure 30: Initializing order for each of the peripherals.

Figure 31: (A) The accelerometer interrupting the device when a force above the threshold is detected. (B) The accelerometer interrupting when the device does not detect any force for a set amount of time.

the DC component for both the red and IR LEDs. With these values the SpO_2 , HR, SpCO , and body temperature can be approximated.

The accelerometer's important information is only when two different interrupts occur. First when a force is detected above a threshold. Second when no activity force is detected for a set period of time. Each of these cases have the ability to interrupt the device to obtain immediate attention. If an impact occurs the value is recorded and the alert is issued, as shown in Figure 31 A. If the subject is not moving the device will either start a counter if it is a new event or increase the counter if old event then issue the alert, as shown in Figure 31 B. The counter is used to record the amount of time the subject has not been moving and is reset once movement is detected.

4. Results

Each sensor, in the device, has been tested against a similar commercially available device.

Testing against commercially available devices allows both the hardware and software to be verified for both functionality and accuracy.

4.1. Pulse Oximeter

4.1.1. Experimental Setup

The selected commercial device, which the device was tested against, was the Masimo RAD-57 Pulse CO-Oximeter, which attaches to a finger and is capable of measuring both HR and SpO₂. Detection of SpO₂, with the RAD-57, has a resolution of 1% and an accuracy between $\pm 2\%$ to $\pm 3\%$, with minimal accuracy difference with motion. Detection of HR, with the RAD-57, has a resolution of 1bpm with an accuracy of ± 3 bpm without motion and ± 5 bpm with motion. Both the devices' results were gathered simultaneously. The dissertation device result values, which was compared to the RAD-57's results, were calculated from a single gathering cycle [28].

4.1.2. Results

The results, shown in Table 2, compares the dissertation device against the Rad-57 with a low error of 1.75% for HR and 2.08% for SpO₂. The dissertation device's results were calculated from a single gathering cycle when comparing devices. Averaging multiple data sets can reduce error by detection of incorrect or outlier results.

Table 2: The results for the comparison testing of the pulse oximeter sensor. The average error of the HR and SpO₂ was 1.75% and 2.08% respectively.

Device		Rad-57		Percent Error	
HR (BPM)	SpO ₂ (%)	HR (BPM)	SpO ₂ (%)	HR (%)	SpO ₂ (%)
65.89	95.43	67	97	1.66	1.62
67.57	94.99	69	97	1.63	2.07
69.56	94.84	71	97	2.33	2.23
72.49	95.30	74	97	2.04	1.76
76.48	94.33	77	97	1.10	2.75

4.2. Pulse CO-Oximeter

4.2.1. Experimental Setup

The selected commercial device, which the device was tested against, was the Masimo RAD-57 Pulse CO-Oximeter, which attaches to the finger and is capable of monitoring SpCO. Detection of SpCO, with the RAD-57, has a resolution of 1% and an accuracy of $\pm 3\%$. Both devices' results were gathered simultaneously. The dissertation device's gathered data, which were compared to the RAD-57's results for correlation, was gathered from a single gathering cycle and involved recording the AC and DC parts for all three LEDs [28].

4.2.2. Results

The sample distribution of the SpCO measurements from the RAD-57 is shown in Figure 32.

Using the SpCO samples from the RAD-57 and the gathered data from the dissertation device, which consists of AC and DC for all three LEDs, the correlation was calculated. The correlation

Figure 32: The sample distribution for the SpCO results from the Masimo RAD-57. was calculated for each value and combination of values to determine which could be used as an indicator for SpCO change in the subject, as shown in Table 3.

Both AC and AC/DC for the Red LED has the highest correlation to SpCO. Both Figure 33 and Figure 34 shows the relationship and corresponding linearly fitted equation. Results from this section require further investigation with more data points and greater SpCO range, which is beyond the goals of the presented work.

Table 3: The correlation for the each of the different AC and DC values for each LED to the SpCO results. The two highlighted values have the highest correlation.

Red			IR			Other LED		
AC	DC	AC/DC	AC	DC	AC/DC	AC	DC	AC/DC
0.8134	-0.4141	0.8405	0.3931	0.0179	0.4880	0.2914	-0.1075	0.4444

R/IR	R/O	IR/O	R/IR/O	R/O/IR	O/R/IR	IR/R/O	IR/O/R	O/IR/R
0.7202	0.6683	0.6683	0.2697	0.2697	-0.6854	-0.7338	-0.7338	-0.6854

Figure 33: The graph showing the correlation between the red LED AC value and the SpCO.

Figure 34: The graph showing the correlation between the red LED AC/DC value and the SpCO.

4.3. Motion Artifacts in AC Signal

Motion artifacts occur when a signal is being read and the object under observation is in motion. This can easily occur when a sensor is attached to a subject conducting physical activity, such as a firefighter on active duty. Motion artifacts will distort the desired signal and create either distort data or create unusable data. One cause of these artifacts occur due to a loose connection between the subject and the sensor allowing the sensor to move, with respect to the subject, and cause such distortion. The dissertation device tries to overcome this cause through the use of double sided adhesive tape, which allows for a more uniform connection to the subject.

4.3.1. Experimental Setup

The motion artifact testing involved exposing the dissertation device to different levels of motion and sending the raw AC and DC signals from the dissertation device to the computer for motion artifacts to be identified and processed.

The motion artifacts occurred when the dissertation device was under intense motion. A snapshot of the raw data is shown in Figure 35 which best shows the two different motion artifacts observed. The first motion artifact is when the AC signal hits the rails of the amplifier, when the data is above 650 or below 0, theorized to occur due to the increase blood flow through the body during extreme physical activity. The second motion artifact occurs when multiple false peaks are detected and unreliable results are calculated. The first motion artifact is corrected by estimating the peaks above and below the rails, through software. The second motion artifact is corrected by comparing the calculated heart rate, for the current peak, to previous results and determining if it is an outlier. The data point is an outlier if it is significantly higher or lower than previous result. If determined to be an outlier data point the peak is discarded.

Figure 35: Motion artifacts seen when measuring the AC signal while the device is under intense motion. Both type of motion artifacts can be seen. The first is before 3 seconds in which the signal hits the rails of the amplifiers. The second is after 3 seconds where there are additional peaks in the signal.

Two motion artifact correction methods were implanted on the raw data. These correction methods were used in order to determine the impact and recoverability for the first type of motion artifact. These two correction methods involve determining the peaks beyond the limits of the rails, through continuing the wave in either a triangle or parabolic shape. The triangle correction method involves determining the slope for two points on each side, then extending each side until they meet at a single point, which is the peak location for the triangle correction. The parabolic method involves finding three points, two on one side and one on the other side of the rail limit, then calculating the vertex of the parabola.

Figure 36: Displays the two correction methods used for determining the peak beyond the rails of the amplifiers.

4.3.2. Results

The peak was determined, for a flattened signal, by calculating average x-axis position and the rail limit for the y-axis. Figure 36 shows the peak location difference for both correction methods and the flattened signal, for a single minimal that exceeded the limit of the amplifier rails.

Both of these methods predicted different peak locations on both the x and y axis. The x-axis location is used for HR while the y-axis is used for SpO₂. Figure 37 visually shows the difference in HR for different times when the AC signal was beyond the limits of the amplifier rails.

Table 4 shows the percent error between the measured value and the correction methods and uncorrected method. Each method has roughly the same error with the triangle method having the highest, this is due to the x-axis position having roughly the same values with and without correction.

Figure 37: Shows the difference between HR when using different correction methods compared with the measurement device and without correction.

SpO₂ on the other hand does use the y-axis, which was limited by the amplifier rails has the most impact from correction. Theoretically, the chosen correction method, used on both signals, should not matter since the SpO₂ is a ratio of signals allowing the error to be reduced. Figure 38 visually shows the calculated SpO₂ using the correction methods, uncorrected data, and measured SpO₂.

Table 4: Shows the difference and error of each correction method and without correction for HR.

Time (Mins)	Measured (BPM)	Flattened (BPM)	Triangle (BPM)	Parabolic (BPM)	Flattened Error (%)	Triangle Error (%)	Parabolic Error (%)
16	172	167.9	163.3	167.9	2.39	5.03	2.39
17	175	167.1	173.7	166.6	4.50	0.72	4.79
18	180	170.5	167.5	173.1	5.29	6.94	3.84
19	180	173.3	169.4	171.9	3.72	5.90	4.51
20	179	174.4	172.9	174.0	2.59	3.42	2.78
Average Error (%)					3.70	4.40	3.66

Figure 38 – Shows the difference between SpO₂ calculation when using different correction methods, uncorrected data, and SpO₂ measurement device.

Table 5 shows the percent error between the correction methods and uncorrected data against the desired SpO₂ measurement device results. Both correction methods have roughly the same error, with the parabolic correction having slightly higher error.

Both correction methods show significant improvement, when compared to uncorrected data results, but neither approach can completely correct the problem. As such the parabolic method was chosen to be incorporated into the dissertation device. This decision was based on the decreased computation time of $O(1)$ for the parabolic method, due to only requiring use of five

Table 5: Shows the percent error for correction method and uncorrected data against the measured SpO₂.

Time	Measured	Flattened	Triangle	Parabolic	Flattened Error	Triangle Error	Parabolic Error
(Mins)	(%)	(%)	(%)	(%)	(%)	(%)	(%)
16	98	77.2	88.1	86.2	21.27	10.11	12.03
17	97	77.3	87.2	81.0	20.27	10.13	16.52
18	98	78.2	87.7	87.7	20.24	10.50	10.56
19	100	77.3	84.0	83.7	22.68	16.02	16.30
20	100	77.1	87.9	85.9	22.93	12.14	14.11
Average Error (%)					21.48	11.78	13.90

equations, versus the computation time of $O(n)$ for the triangle method, due to looping through each flattened data point. A better method, which requires hardware adjustment, for fixing the issue is discussed in 5.3. Discussion.

4.4. Accelerometer

4.4.1. Experimental Setup

The accelerometer was tested by verifying the sensor is able to interrupt the device and report back with the impact force. In order to complete this test, the device was dropped roughly six feet onto a padded surface. Then the data was gathered and used for calculating the impact force vector from the values in the x, y, and z axis.

4.4.2. Results

The results shown in Figure 39, indicate that the accelerometer is successful in detecting, recording and reporting an impact from a fall or other forces, which may act upon the subject.

Figure 39 – Three separate drop tests of the device in order to verify that the device is capable of interrupting and recording not only the peak but the force before and after the event.

4.5. Humidity and Suit Temperature Sensor

4.5.1. Experimental Setup

The selected commercial device in, which the dissertation device was tested against, was the AcuRite 00613 Humidity Monitor with Indoor Thermometer, which is capable of detecting both relative humidity and temperature of the environment. Detection of the relative humidity, with the AcuRite, has resolution of 1% with an accuracy of $\pm 3\%$ to $\pm 5\%$ depending on relative humidity range. Detection of the temperature, with the AcuRite, has a resolution of 1°F with an accuracy of $\pm 2^{\circ}\text{F}$, the temperature can easily display Celsius instead of Fahrenheit [29].

4.5.2. Results

The results shown in Table 8 compare the dissertation device against the AcuRite with a percent error of 4.43% and 2.73% for temperature, when measured in Celsius and Fahrenheit respectively, and 7.44% for humidity. The dissertation device's results were calculated off a single data set,

Table 6: The accuracy measurement of the suit humidity and temperature sensor. The average error of the temperature for Celsius and Fahrenheit is 4.43% and 2.73% respectively, and for humidity is 7.44%.

AcuRite			Device			Percent Error		
Humidity	Temperature		Humidity	Temperature		Humidity	Temperature	
(%)	(°C)	(°F)	(%)	(°C)	(°F)	(%)	(°C)	(°F)
43	23	73	49.88	23.35	74.04	16.00	1.54	1.42
48	23	73	52.10	23.85	74.92	8.54	3.68	2.64
36	25	77	38.96	26.44	79.58	8.23	5.74	3.35
37	26	79	40.29	26.11	79.00	8.89	0.43	0.00
40	26	79	42.76	26.62	79.91	6.90	2.37	1.15
36	26	79	39.48	27.62	81.72	9.66	6.24	3.45
53	29	84	54.57	30.45	86.82	2.97	5.01	3.35
53	29	84	53.36	30.50	86.91	0.69	5.19	3.46
60	30	86	53.88	32.82	91.08	10.21	9.40	5.90
53	32	90	51.77	33.52	92.33	2.32	4.74	2.59

when determining both the temperature, in both Celsius and Fahrenheit, and humidity. Averaging multiple data sets can help with detection of incorrect or outlier results.

4.6. Skin Temperature

4.6.1. Experimental Setup

Similar to the Humidity and Suit Temperature the selected commercial device, which the dissertation device was tested against, was the AcuRite 00613 Humidity Monitor with Indoor Thermometer.

4.6.2. Results

The results, shown in Table 3, compare the dissertation device against AcuRite, shows a percent error of 2.83% and 1.49% for Celsius and Fahrenheit respectively. The dissertation device's results were calculated off a single data set when determining the temperature, in both Celsius and Fahrenheit. Averaging multiple data sets can help with detection of incorrect or outlier results.

Table 7: The accuracy measurement of the skin temperature sensor. The average error of the temperature for both Celsius and Fahrenheit is 2.83% and 1.49% respectively

Acurite		Device		Percent Error	
(°C)	(°F)	(°C)	(°F)	(°C)	(°F)
23	73	21.13	70.04	8.11	4.05
23	73	22.04	71.66	4.19	1.83
25	77	24.92	76.86	0.32	0.19
25	77	24.38	75.88	2.48	1.45
26	79	26.54	79.78	2.09	0.98
27	81	28.53	83.35	5.65	2.90
29	84	29.07	84.32	0.23	0.38
29	84	29.07	84.32	0.23	0.38
30	86	31.05	87.89	3.50	2.20
32	90	32.49	90.49	1.54	0.54

4.7. Core Temperature

4.7.1. Experimental Setup

Core temperature was tested against the oral ANKOVO Medical Digital Thermometer. ANKOVO has a resolution of 0.1°F and an accuracy of $\pm 0.2^{\circ}\text{F}$ when measuring between the temperature of 95.9°F and 107.6°F . Oral thermometer was chosen due to the ease of use. Different HR intensities were used for the calculations to approximate the core temperature. Testing started with no activity for the first 5 minutes. Next medium activity for next 60 minutes. Followed by 20 minutes of light activity. Finally ended with no activity [30].

Figure 40 – The results of the core temperature testing. The dissertation device’s HR Temp Approx, red line, show the results of the device approximating the core temperature based off HR readings. While the ANKOVO, blue line, is an oral temperature results.

4.7.2. Results

Figure 40 shows the results from the core temperature testing. The dissertation device's approximated core temperature follow the similar trend set by the ANKOVO, with a greater gap between the two during the peak core temperature readings. The highest difference between the two results was found to be 1.657°F with a percent error of 1.674%.

Rectal measurement would have obtained a better measurement of the core temperature, while being more invasive and difficult to record. Oral and rectal measurements differ depending on the amount of physical activity. A study found that for a stepping rate of between 3 and 6 steps a minute would have an average difference of less than 0.55°C, while 24 steps a minute would average a little more than 1.0°C. In addition, some individuals experienced a difference of 2.0°C. These difference have been theorized to the legs, which conduct the work, being closer to the rectal measurement site then the oral measurement site [31].

4.8. System Level Testing

University of Cincinnati IRB study # 2017-6102 “System Level Testing of the Physiological Health Assessment and Hazard Monitoring Patch” was submitted and accepted as a minimal risk study. This study was conducted in order to test the device as a system and not designed to test individual sensors. The dissertation device was set to gather the subject's physiological status during testing. The sensors were active, with the exception of the SpCO, and could send alerts. The study was not designed for any alerts to be set off, for example impact alerts was purposely designed not to be set off due to the danger that the subjects would be exposed. SpCO was inactive during the system level testing due to additional development of the sensor required.

4.8.1. Protocol

A total of 10 people were tested on a treadmill. The study involved first a questionnaire, which gathered basic information from the subject. The information requested involved age, gender, height, weight, smoker status, required physical fitness for a career or team, amount of cardiovascular exercise per week, and duration of cardiovascular exercise.

The physical activities, in which the subjects were requested to complete, while having the dissertation device applied to their upper arm, involved the following:

1. Walking for 5 mins
2. Jogging for 5 mins
3. Running for 5 mins

Each activity's speed was set by the test subject, in order to obtain a comfortable speed for the amount of time. The subjects were allowed to rest after each activity before continuing to the next activity, which no subjects chose to rest. These activities were seen as a way to test the device at different activity levels while not forcing the subjects to overexert themselves. The only perceived risks of the study was a minimal risk of fatigue, respiratory discomfort, and the possibility of slips and falls from physical activity.

4.8.2. Results

A total of 10 subjects, ranging in both age and body mass index (BMI) as seen in Figure 41, participated in the system level testing. No subjects experienced any harm during the study.

Environmental conditions, for the study, had a recorded temperature between 76.2°F and 82.4°F, and humidity between 35.1% and 47.9%.

Figure 41: (a) The age range and (b) the BMI range of all the test subjects who participated in the study.

The dissertation device showed constant environmental temperature and environmental humidity throughout each subject's test period. The reason for the constant environmental temperature and environmental humidity is because the subjects' local environmental conditions are the same as the global environmental conditions. These readings are important because the firefighters' local environment would be the environmental conditions inside the protective clothing.

The physical status results of the test subjects are shown in Figure 43 through Figure 45. Most of the results show an increasing trend as the physical activity intensity increase. The only results that shows a downward trend is SpO_2 , which is due to motion artifacts as explained in section 4.3. Motion Artifacts in AC Signal. Only one test subject set off the impact alert, with an impact force of 6g, due to bouncing on the treadmill when switching from a jogging pace to a running pace.

Figure 43 – Shows the HR results of the system level testing. The line shows the average HR with error bars. As expected the HR starts out lower and show increase with activity.

Figure 42 – Shows the SpO₂ results of the system level testing. The line shows the average SpO₂ with error bars.

Figure 44 - Shows the CT results of the system level testing. The line shows the average CT with error bars. The increasing error trend in CT is due to the different levels of intensity from the subjects.

Figure 45 - Shows the skin temperature results of the system level testing. The line shows the average skin temperature with error bars. The increasing error trend in skin temperature is due to different levels of intensity from the subjects.

5. Conclusion and Future Work

5.1. Summary

This dissertation describes a device, which integrates multiple sensors into a small, lightweight, wireless device. The dissertation device is capable of monitoring both physiological status and sudden events. The physiological status including heart rate, blood oxygen level, carbon monoxide in the blood, body temperature, skin temperature, and environmental temperature and humidity. The sudden events include falls, impacts, and burns.

The idea behind the device was to help reduce the number of injuries and deaths among firefighters. The dissertation device is designed to accomplish this by immediately alerting a monitoring personal when conditions indicate an injury is likely to occur or when a sudden event happens. The additional information, from the dissertation device, will allow a monitoring personal to recall the firefighter in order to reduce the chance of injury; such as resting, rehydrating, or cooling down before resuming duty; or if a sudden event does occur the monitoring personal can send immediate help to retrieve the firefighter and provide immediate care.

5.2. Conclusion

This dissertation presented the research work on a potential safety device. The results from each sensor's test along with a system level test was presented, as shown in section 4. Results . Each sensor was individually tested against an equivalent commercially available device, in order to determine the error of the dissertation device's sensor and the ability to be used for determining the physiological condition of the subject. Most of these sensors were found to be comparable results with their commercially available counterparts, with only the SpCO requiring

in depth testing. Due to dangerous required conditions, some sensors were not able to be fully tested. For example, the SpCO sensor and accelerometer could not be fully tested due to the fact such conditions had the potential for injuries or death. As such these sensor were tested to the best of their ability without exposing anyone to potential injuries or death. Alerts are meant to be issued when physiological conditions were nearing the danger zone in which injuries are likely to occur, therefor testing the alerts activation on subjects was not conducted.

The dissertation device was then tested on human subjects, with the results shown in 4.8. System Level Testing. The system level testing was designed to show how the device's sensors functioned together to monitor the subject. The testing was designed such that the subjects were not at risk of injury or harm. The system level testing showed that the dissertation device is capable of monitoring subject's physical status at multiple activity levels. The only sensor which had trouble with the system level testing was SpO₂, which is due to motion artifacts as explained in section 4.3. Motion Artifacts in AC Signal.

In summary, this dissertation presented proof-of-concept for a small, lightweight, flexible, and wireless device, which can monitor the physiological status and sudden event that are present for a firefighter on active duty. Through use of the dissertation device the number of injuries and deaths occurring every year can be reduced. Completion of this work has the potential to greatly the impact the way we monitor hazards.

5.3. Discussion

This research integrates technology from multiple devices into a discreet, lightweight, and wireless device. The dissertation device was designed for monitoring the dangers present to firefighters, when on active duty. The dissertation devices, which contains a generic set of

sensors, can be implemented into a multitude of other high risk professions, where minutes between incident and treatment can be the difference between life and death. Examples of high risk professions which could benefit from a safety monitoring device include but not limited to construction laborers, power plant workers, coal miners, and etc. The dissertation device and similar safety devices are ideal for professions where a person has the potential to be alone in a dangerous situation.

In addition the dissertation device can be used to gather data without interfering with mobility, useful for scientific studies or monitoring performance. This dissertation device has the potential to be used for research purposes by replacing the commercially available devices, such as the pulse oximeter which requires attachment to a finger or earlobe. Through attaching the dissertation device to an athlete, or other person of interest, both physiological data and environmental data can be recorded and wirelessly transmitted. Allowing the subject to continue activities without stopping to gather the desired data.

5.4. Future Work

The presented dissertation device is a proof of concept device, which has areas that can be improved and areas that must be improved before the device can be used in the field. The presented device is in the first steps towards creating a lightweight and wireless device which is capable of monitoring a firefighter on active duty.

The areas in which the dissertation device must be improved include correction of the AC signal flattening due to motion artifacts, further SpCO development, the device's packaging, method of attaching to the subject, and interference of sweat. Each of these areas must be addressed before the device can be used in the field. Correction of the AC signal flattening will decrease the SpO₂

error, with activity, allowing for a more reliable measurement. This correction requires hardware change by replacing the amplifier's gain resistor with a variable resistor controlled by the microcontroller in order to change the gain of the AC amplification. The microcontroller could either detect when the AC signal is above the amplifier rails, which would require the microcontroller to re-measuring the AC signal, or use the accelerometer to detect physical activity intensity and scale the amplifier gain accordingly, which would require determining the correlation of accelerometer activity to required amplifier gain. Further development of the SpCO would allow for more complete equation across a wider range of values. The device's packaging and method of attaching to the subject allowed for a system level testing on human subjects but lacks reusability and longevity, as such both would require a redesign for use in the field. The adhesive used on the presented device did protect the circuitry from sweat but with the packaging and method of attachment being reevaluated sweat also must be taken into consideration. Even though the device was designed for firefighters the selected components require additional screening and testing under extreme environmental conditions and encapsulated environments which the firefighters are exposed.

Areas in which the dissertation device can be improved in order to further expand the capability of the device include optimizing the results for individuals, increased accelerometer data usage, use the data to help predict events before occurrence, and verification of core temperature.

Transition from generic physiological data collection and alerts by moving toward optimizing the results for individuals would allow the system to record and predict the limits for each individual. Increasing the accelerometer data usage can allow the device to recognize events including walking, running, and fall prediction along with detecting a man down situation due to the change of gravity acting upon the chosen axis. Using not only the accelerometer but also all

the other sensors' data, the device could be expanded upon allowing more events to be predicted before occurrence by recognizing results outside the norm which will require further testing and development. The core temperature approximation requires verification on a subject who is not only in an extreme environment but also an encapsulated suit comparable to the conduction firefighters experience while on active duty.

Further improvements which are required for the proposed device to be used in the field but do not directly affect the device's gathered results include creation of a GUI interface, allowing multiple devices to run at once, and an improved communication method. Creation of a computer or phone based graphical user interface (GUI) would allow subject's results and alerts to be neatly displayed for the monitoring personal. The GUI's and device's communication method must be expanded in order to allow multiple devices to be active at the same time. It would require adding an identification method for each patch and a list of active devices whose data needs to be gathered, stored, and displayed. The GUI interface would either need to be able to switch between subjects or display all subjects, while being able to draw attention to any subject whose patch has issued an alert.

Bluetooth was chosen as the wireless communication protocol but is not able to communicate through obstacles encountered during active firefighting due to range and penetration limitations. A change in the wireless protocol for both the device and monitoring interface is required for transmitting on the field across longer distances and through buildings, allowing the important information to be transmitted while the firefighters are on an active fire.

When all these requirements are met then the device can be tested and used on firefighters while on active duty.

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