DETECTION OF RESPIRATORY INFORMATION USING ELECTROMAGNETIC BIOSENSORS

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Abstract

Continuous respiratory activity monitoring can help predict and identify respiratory failure, and could potentially be life-saving. Recent wearable, wireless technology allows users to monitor important physiological signals, such as heart and respiratory rate, with the advantages of comfort and portability. However continuous, remote, monitoring requires regular battery replacement, which poses an inconvenience for the user, and a barrier to compliance and wide adoption of this technology. If respiratory energy is harvested, the energy can provide power for such a wearable biosensor, thus eliminating the need for regular battery replacement. The work in this dissertation demonstrates the feasibility of the zero-net energy biosensor concept. Contributions of this dissertation work to electrical engineering include the design of an electromagnetic respiratory effort harvester, the first human study data of respiratory rate and tidal volume detection using electromagnetic biosensors, and an investigation of simultaneous harvesting and sensing feasibility with a low-power system-on-chip. Respiratory effort is harvested through electromagnetic generation, while concurrently sensing critical respiratory parameters. Methodology for extracting respiratory parameters from electromagnetic generator outputs is investigated. High sensing accuracy of both respiratory rate and tidal volume is initially demonstrated using a mechanical target simulating respiratory motion. Close agreement between electromagnetic sensor outputs and a gold standard for respiratory measurements is further demonstrated through human testing.
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Chapter 1
Overview

Respiratory failure is difficult to predict and can become life-threatening in minutes, or even build up gradually. Thus, continuous monitoring of respiratory activity is important for predicting or identifying fatal or high-risk situations, and appropriate respiratory monitoring techniques could potentially be life-saving [1].

Continuous monitoring of respiratory activity is of great importance in the assessment of a patient’s physiological state [2]. Recording respiratory signals can be useful for monitoring patients with breathing problems, aiding in the diagnosis of disease, and evaluating the response to medication that affects the respiratory system. Patients with chronic diseases, elderly citizens living at home alone, and those working under heavy loads, including firefighters and military personnel, could benefit from continuous and remote respiratory monitoring.

Two of the most important parameters of respiratory activity are respiratory rate and tidal volume. Respiratory rate is the number of breaths taken over a set amount of time, usually reported as breaths per minute. The average respiratory rate in adults is 12 breaths per minute. Tidal volume is the volume of air displaced between normal inspiration and expiration, typically reported in milliliters (mL) or liters (L). The average tidal volume for adults is about 500mL, or a half-liter.

The gold standard for measuring respiratory rate and tidal volume is the spirometer, shown in Fig. 1.1. Spirometers measure the patient’s airflow, which requires the patient to breathe into a facemask, or a mouthpiece while wearing a nose clip. The apparatus can be obtrusive and uncomfortable for the patient, and can thus alter the patient’s breathing pattern. The facemask or mouthpiece also prevents the patient from communicating verbally during monitoring. The facemask or mouthpiece of the spirometer contains a differential pressure transducer, which is wired to a PC for processing. There may also be a flexible tube that connects the facemask or mouthpiece
to a gas analyzer. These connections limit the mobility of the patient, so remote monitoring is infeasible.

![Patient using a spirometer.](image)

Wireless spirometers have been recently developed, which allow the patients to move freely. They still utilize a mouthpiece, so patients may still feel some discomfort. Some wireless spirometers have digital displays on the device, as shown in Fig. 1.2, while others can transmit the data wirelessly to a PC or other remote receiver display. Batteries provide the power required for processing and transmission. Monitoring continuously and remotely would require regular replacement of batteries, which present an environmental hazard and an inconvenience to the user.
Recent wearable, wireless technology, also known as smart clothing or smart garments, allows users to monitor important physiological signals, such as temperature, heart rate, and respiratory rate, with the advantages of comfort and portability. However, like wireless spirometers, wearable technology requires batteries to power the sensors, processor, and wireless link. Even with an ultra low-power system-on-chip, such as the Texas Instruments CC430, continuous monitoring and data transmission require frequent battery replacement. Assuming the CC430 is powered by CR2025 lithium batteries, which are used in many Polar heart rate monitors, monitoring for one minute and transmitting respiratory rate and tidal volume data once every hour would require replacement every nine days [3, 4]. Frequent battery replacement presents an inconvenience for the user, as well as a barrier for compliance and wide adoption of wearable sensors.

Harvesting energy from respiration could eliminate the need for regular battery replacement, and the extraction of respiratory information from the harvester signal would realize the concept of a zero-net energy biosensor. A zero-net energy biosensor would reduce the amount of disposed batteries, and would also be useful in applications where it is difficult to replace batteries frequently or where there is no access to batteries.

Extraction of information from an energy harvester output has been done in other fields and applications [5-7]. Respiratory energy harvesters have been developed [8, 9], but primarily just for harvesting energy and not for simultaneous monitoring of respiratory activity. The proposed biosensor would be the first to harvest respiratory
energy and simultaneously sense respiratory information using electromagnetic generation.

This dissertation examines the feasibility of simultaneously detecting respiratory information and harvesting respiratory energy to realize a zero-net energy biosensor. It is organized into nine chapters. The remainder of this chapter provides background of the importance of respiratory rate and tidal volume monitoring, as well as current methods of sensing these parameters. A background of wearable technology and human energy harvesting is also provided, as well as the summary of the research objectives. Chapter Two explains the theory of using electromagnetic generation to harvest respiratory energy, as well as to detect respiratory rate and tidal volume. Chapter Three explores the design and operation of the electromagnetic biosensor, as well as the design of a mechanical chest for testing. Chapter Four presents the protocol for human testing and the measurement of human subjects. In Chapter Five, the average amount of power harvested during human testing is presented. Chapters Six and Seven describe the detection of respiratory rate and tidal volume respectively, from the electromagnetic biosensor output. Chapter Eight explores the feasibility of continuous and remote monitoring with the electromagnetic biosensor and a low-power system-on-chip. Chapter Nine provides a conclusion and offers recommendations for future work on this topic.

1.1. Importance of Respiratory Rate Monitoring and Current Monitoring Techniques

Abnormal respiratory rates and changes in respiratory rate can serve as a precursor for many health crises, such as cardiac arrest, changes and/or exacerbations of chronic illnesses, the immediate admission to an intensive care unit (ICU), respiratory dysfunction, and other major physiological instabilities [10, 11]. Respiratory function can also be affected by changes in organ systems, including the nervous system, cardiovascular system, and excretory system. Respiration monitoring can thus provide an indication of imbalance or failure in these organ systems as well.

Spirometers implement pressure transducers in the path of the airway that measure the pressure differences caused by the flow pattern, and converts those
differences into electrical signals for processing. Temperature, humidity, and carbon
dioxide sensors also directly measure the patient’s airflow, and thus must be placed in the
path of the patient’s airway. The airflow temperature (inhaled air is usually cooler than
exhaled air), humidity (exhaled air usually has higher humidity than inhaled air), and
carbon dioxide concentration (exhaled air usually has a higher CO₂ concentration than
inhaled air) differences are used to distinguish inspirations and expirations [12-15].

Other sensors have been developed that can measure respiratory rate without
directly measuring the patient’s airflow. These sensors are usually integrated into torso
bands or belts and measure changes that occur on the chest or abdomen during
respiration. Piezoelectric sensors, linear variable differential transformers (LVDT), and
strain gauges all measure the displacement change between inspirations and expirations
of the rib cage and abdomen. Respiratory inductive plethysmographs (RIP) measure the
change in inductance of a loop of wire wrapped around the rib cage and abdomen. Thoracic impedance sensors measure the impedance change of the lungs as the volume of
air changes. The impedance can be measured with the same electrodes used for
electrocardiograms (ECG). These sensors can all measure respiratory rate accurately and
are non-invasive, but still require the patient to be wired to stationary medical equipment.

Non-contact methods, such as acoustic sensors [16], ultrasonic sensors [17],
infrared motion capture, or methods employing the Doppler effect [18], require the
patient to be within the effective area of the sensor. Acoustic sensors measure the low-
frequency sound waves emitted by the flow of air during breathing. Ultrasonic, infrared,
and Doppler radar sensors all measure the chest and/or abdomen displacement during
breathing.

Wireless spirometers and other respiratory monitoring systems allow for comfort
and full mobility, but the constant need for battery replacement make these systems
infeasible for continuous and remote monitoring.
1.2 Importance of Tidal Volume Monitoring and Current Monitoring Techniques

Tidal volume is a key index of the mechanical status of the ventilatory system, and can forecast lung disease [19]. Abnormal tidal volumes or changes in tidal volume are useful in assessing conditions such as asthma, chronic obstructive pulmonary disease, cystic fibrosis, pulmonary fibrosis [20].

Spirometry is the most common method of measuring tidal volume. The pressure difference measured by the transducer is linearly related to the volume of air inhaled or exhaled. A spirometer can be uncomfortable for the patient, prohibits the patient from communicating verbally, and limits their mobility. Breathing exclusively through the mouthpiece requires cooperation from the subject. Therefore, the spirometer is often not suitable for infants, children, and heavily sedated or unconscious patients. Effective medical assessments require that measurement techniques neither impede unrestricted chest motion nor influence subject’s respiration pattern.

Studies have shown there is a linear relationship between tidal volume and chest wall displacement during unobstructed breathing. Chest wall displacement has been measured via lasers [21], piezo sensors [22] and Doppler radar [23].

Tidal volume has also been demonstrated to be directly proportional to chest (rib cage) circumference change [24, 25]. A few methods have been developed to measure chest circumference dynamically. Pneumatic belts utilizing air [24] or mercury [25] use transducers to relate the change in length of the chest belt to the change in pressure within the belt. Another type of chest belt uses an ultrasound emitter and receptor on opposite ends of the belt, and relates the time between sent and received signals to the length of the belt [26]. A magneto-resistive sensor system has also been developed, where circumference change is determined from the magnetic polarity or flux density change detected by the sensors, depending on the polarity setup of the magnets [27]. Circumference change can also be measured with transthoracic inductance and impedance plethysmographs, strain gauges, and piezoelectric belts. Three-dimensional body scanning is a non-contact method that can dynamically measure chest circumference change, but this method immobilizes the patient.
While these methods can measure chest circumference change dynamically and accurately, the electromagnetic biosensor has the added benefits of low cost, portability, simplicity, and energy harvesting capability.

1.3 Wearable Technology and Human Energy Harvesting Background

Wearable health monitoring systems are a promising new way of collecting physiological data, without immobilizing or inconveniencing patients [28, 29]. Physiological signals are measured with the garments’ embedded sensors, processed locally, and the resulting data is then wirelessly transmitted to an accompanying wristwatch, smartphone, or PC. Fig. 1.3 shows a smart shirt, the Under Armour E39, that can measure heart rate, respiratory rate, and G-force, which can help evaluate an athlete’s performance and assess possible concussions or other injuries [30].

Fig. 1.3: Under Armour E39 smart shirt.

However, continuous health monitoring with such systems would require frequent battery replacement, which may not be feasible or desirable. The need for frequent battery replacement could also create a barrier for compliance as well as for wide adoption of this technology. In military and firefighting applications, users cannot easily replace batteries in the field, or they may not have access to batteries at all.

Human energy harvesting could be used to provide power for the sensors and wireless link, thus creating a wearable, zero-net energy biosensor. Human energy harvesting for wearable and portable electronics was proposed in the mid 1990’s [31].
While a number of potential human energy sources were identified in [31], human energy harvesting has mostly been focused on kinetic energy [32-34], particularly on the locomotion during walking [29].

In [34, 35], the concept of self-powered biosensors through sensing and harvesting methods for respiratory effort was introduced. A Piezo Systems D220-A4-503YB lead zirconate titanate (PZT) bending sensor was fastened to the chest of a human subject, as shown in Fig. 1.4, and the voltage output was recorded during normal breathing. Piezoelectric respiratory effort harvesting has shown to output high voltages (~8.5 volts peak-to-peak) as shown in Fig. 1.5. While piezoelectric materials are sensitive and feasible as a respiratory sensor, they also have low inherent capacitance (232nF for this PZT sample), resulting in a low power output (~0.4µW during normal breathing). This low power output makes piezoelectric materials infeasible as respiratory energy harvesters.

![Piezoelectric generator on human subject.](image)
It has been demonstrated that electromagnetic scavenging is more efficient than piezoelectric scavenging [33]. While electromagnetic generators may be feasible for harvesting respiratory energy, their ability to detect respiratory information needs to be explored.

### 1.4 Specific Aims for Electromagnetic Biosensing

Wearable technology has allowed for accurate physiological sensing without sacrificing the comfort or mobility of patients, but its reliance on batteries makes continuous and remote monitoring inconvenient or difficult. The main goal of this study is to design an electromagnetic biosensor capable of detecting respiratory rate and tidal volume accurately while simultaneously harvesting significant respiratory energy. The energy harvested would provide the power for a low-power system-on-chip (SoC) for respiratory monitoring, resulting in a zero-net energy biosensor. The architecture for this system is shown in Fig. 1.6.
Fig. 1.6: Zero-net energy biosensor architecture.

Respiratory effort would be simultaneously harvested and sensed by the electromagnetic biosensor. The voltage output of the biosensor would be rectified and stored in a capacitor or battery to power the SoC. The voltage output would also be the respiratory signal, to be digitized and processed by the SoC. The respiratory rate and tidal volume will be calculated by the SoC, and transmitted wirelessly to a wristwatch, smartphone, or other remote receiver display. The four main objectives of this study are:

1. To design an inexpensive, ergonomic electromagnetic biosensor capable of harvesting and sensing respiratory effort.

2. To determine respiratory rate from the electromagnetic biosensor voltage output.

3. To determine tidal volume from the electromagnetic biosensor voltage output.

4. To investigate the feasibility of continuous and remote respiratory monitoring with the electromagnetic biosensor and a low-power SoC.
Results from this study can help in the optimization and integration of an electromagnetic biosensor into clothing. The theory for respiratory rate and tidal volume detection can be applied to any type of circumferential electromagnetic generator. Other physiological signals, such as heart rate, may be integrated into the zero-net energy biosensor as well.
Chapter 2

Electromagnetic Generation Theory

Electromagnetic (EM) generation is based on Faraday’s Law, shown in (2.1), which states that a changing magnetic field induces an electric field, resulting in an electric current through a conductor.

$$\nabla \times E = -\frac{dB}{dt}$$  \hspace{1cm} (2.1)

For a coil of wire, the electromotive force (emf), or voltage, can be determined from the number of coils, N, and the rate of change in magnetic flux, $\phi_B$. This relationship is shown in (2.2).

$$V = -N \frac{d\phi_B}{dt}$$  \hspace{1cm} (2.2)

A voltage can be induced across a coil of wire by moving the coil toward or away from a permanent magnet (or vice versa), or rotating the coil relative to the magnet (or vice versa).

2.1 Electromagnetic Respiratory Energy Harvesting

To harvest energy from the movement of the chest walls during respiration, the expansion and retraction of the chest circumference can be used in conjunction with a coil-and-magnet chest belt to achieve the relative motion needed. The chest belt can move the coil toward and away from the magnet (linear EM generation) or rotate the magnet relative to the coil (rotational EM generation) during respiration.

A linear EM generator chest belt can be designed with a magnet plunging into and out of a coil, as shown in Fig. 2.1, or with a magnet and coil shearing against each other, as shown in Fig. 2.2. For both configurations, the generator oscillations would be achieved from the movement of the chest walls during inspiration and expiration.
A rotational EM generator chest belt can be achieved with a DC brushed motor, as shown in Fig. 2.3. The chest belt can be configured so that during inspiration, the rotor of the motor is turned. An added spring can rotate the coil back in the opposite direction during expiration.
2.2 Respiratory Rate Detection From EM Generator

Output

Respiratory information, including respiratory rate and tidal volume, can be extracted from the voltage induced by the generator. Both parameters can be derived from Faraday’s Law. The assumption is made that the number of coils, N, remain constant, and that permanent magnets are used (magnets are not changed dynamically, and magnetic field is constant). For linear EM generation, the magnetic flux is directly proportional to the rate of change of the linear displacement, m, as shown in (2.3). For rotational EM generation, the magnetic flux is directly proportional to the rate of change of the angular displacement, θ, also shown in (2.3). The coefficient L will be used in the derivation for linear EM generators, and the coefficient R will be used in the derivation for rotational EM generators.

\[
\frac{d\phi_B}{dt} = L \cdot \frac{dm}{dt} = R \cdot \frac{d\theta}{dt}
\]  

(2.3)
The circumference change of the chest, $\Delta C_{\text{Chest}}$, is directly proportional to the linear or angular displacement as well, shown in (2.4).

$$\Delta C_{\text{Chest}} = L_2 \cdot m = R_2 \cdot \theta$$ \hspace{1cm} (2.4)

By substituting (2.4) and (2.3) into (2.2), the magnitude of the voltage is also shown to be directly proportional to the rate of change of the chest circumference.

$$V = -\frac{N \cdot L_1}{L_2} \cdot \frac{d\Delta C_{\text{Chest}}}{dt} = -\frac{N \cdot R_1}{R_2} \cdot \frac{d\Delta C_{\text{Chest}}}{dt}$$ \hspace{1cm} (2.5)

The sign of the voltage depends on the direction of the magnet poles, the motor terminals’ polarity, and the sign of the chest circumference change. A positive chest circumference change indicates an inspiration, while a negative chest circumference change indicates an expiration. By correlating the sign of the voltage output to a positive or negative chest circumference change, inspirations and expirations can easily be identified. Respiratory rate can then be measured by detecting the number of inspirations or expirations over time.

### 2.3 Tidal Volume Detection From EM Generator Output

As mentioned previously, tidal volume has been demonstrated to be directly proportional to chest (rib cage) circumference change [24, 25]. Thus tidal volume can be detected by calculating the quantitative chest circumference change during respiration. By integrating (2.5), we can solve for the chest circumference change, $\Delta C_{\text{Chest}}$.

$$\int V \cdot dt = \int -\frac{N \cdot L_1}{L_2} \cdot \frac{d\Delta C_{\text{Chest}}}{dt} = \int -\frac{N \cdot R_1}{R_2} \cdot \frac{d\Delta C_{\text{Chest}}}{dt}$$ \hspace{1cm} (2.6)
\[
\Delta C_{\text{Chest}} = \left( \int V \cdot dt + C_0 \right) \left( -\frac{L_2}{N \cdot L_1} \right) = \left( \int V \cdot dt + C_0 \right) \left( -\frac{R_2}{N \cdot R_1} \right) \quad (2.7)
\]

Let \( -\frac{L_2}{N \cdot L_1} = L \), and let \( -\frac{R_2}{N \cdot R_1} = R \)

\[
\Delta C_{\text{Chest}} = L \cdot \left( \int V \cdot dt + C_0 \right) = R \cdot \left( \int V \cdot dt + C_0 \right) \quad (2.8)
\]

The constants \( C_0, L, \) and \( R \) can be calculated by computing the voltage integral at known chest circumference changes. Thus, regardless of the EM generator being linear or rotational, the chest circumference change can be calculated from the integral of the voltage output and two measured constants.

Tidal volume can then be calculated as a linear function of chest circumference change, as shown in (2.9).

\[
TV = K \cdot \Delta C_{\text{Chest}} \quad (2.9)
\]

The coefficient \( K \) can be found by correlating known chest circumference changes with corresponding tidal volume measurements.
Chapter 3

Electromagnetic Biosensor Design

Choosing between a linear or rotational EM generator for the biosensor involves a balance of cost, ergonomics, and power output.

A common example of linear EM generators is shake flashlights. Shaking the flashlight oscillates a coil around a magnet, which generates power for the bulb. Implementing this concept in a chest belt requires custom fabrication, which can be costly. Rotational EM generators can be implemented by operating a motor in reverse. Turning the rotor causes a coil to rotate within a set of magnets, thus generating energy. Small servo and DC brushed motors are readily available off-the-shelf (no custom fabrication needed), are inexpensive, and can easily be implemented into a chest belt.

The circumferential movement of the chest is parallel to that of the motion of generation in linear EM generators. For rotational EM generators, the chest belt would have to be attached to the rotor of the motor in order to achieve rotation during respiration. In any case, the motor will be perpendicular to the chest circumference, as opposed to linear EM generators being parallel to the chest circumference. Thus, linear EM generators would have a lower profile than rotational EM generators, and thus an ergonomic advantage.

According to Faraday’s Law for a coil (2.2), the voltage output is proportional to the number of coils and the rate of change of magnetic flux. For linear EM generators, the rate of change of magnetic flux is directly related to the chest circumference change during respiration, the respiratory rate, and the magnetic field strength of the magnets. Normal respiration is about 12 breaths/min (0.2Hz), and normal chest circumference changes are less than 3cm. Off-the-shelf neodymium magnets usually have magnetic fields with a strength of about 1 – 1.4 Tesla. For rotational EM generators, the rate of change of magnetic flux is directly related to the angular velocity of the rotor, i.e. how fast the rotor is spinning. The rate of change of magnetic flux for rotational EM generators is still directly related to the circumference change, rate and magnets. However, an armature and gear train can be used to increase the power output efficiency.
of the EM harvester. By adding gears to the rotational EM generator, a small angular displacement of the armature translates to a large angular displacement of the rotor, at the cost of a greater force required to turn the armature. As long as the force from respiration can turn the armature without any discomfort, higher power can be generated from the rotational EM generator with the added gear train.

The limitations of linear EM generators include costly custom fabrication and the need for a high-frequency oscillation or large linear displacement for significant generated power. The limitations of rotational EM generators include that of extra weight from the gears and housing, as well as additional static friction (stiction) from the gears that the force from respiration must overcome. Cogging also occurs in permanent-magnet brushed motors, due to pole construction [36]. A cog or detent is the point at which the center of the magnets line up with the ideal magnetic path through the poles, and extra torque is required to break that attraction. Considering the advantages and disadvantages for both types of EM generators, rotational EM generators were chosen due to off-the-shelf availability, cost, and potential for higher power output with a gear train.

### 3.1 EM Biosensor Design with a Servo Motor

A servo motor is comprised of a permanent magnet DC motor, a gear train, a potentiometer, and some control circuitry [37]. The potentiometer is attached to an armature, which is connected to the gear train, followed by the rotor of the motor. The rotor, which is basically an iron core with coiled wire, rotates within the permanent magnets inside the motor. A current is induced and collected at the motor’s terminals. A servo motor was initially chosen for the EM biosensor due to easy modification into a wearable chest belt, as well as for the built-in gear train.

The selected servo motor is a Futaba S3003 standard servo motor [38], as shown in Fig. 3.1a. The servo motor weighs about 37g, and its dimensions are 40mm x 20mm x 36mm. The armature has four blades with a diameter of 3.8cm, and the gears are nylon. The off-the-shelf servo motor was modified into a wearable chest belt, as shown in Fig. 3.1b.
Plastic spacers are attached to one side of the housing to prevent the protruding armature from hitting the chest. A string is fixed at one end of the armature using a screw. The string is then wrapped tight once around the chest, and the other end of the string is fixed at the opposite end of the armature. A rubber band is also attached between the housing and another blade of the armature. The rubber band provides a restorative force to return the armature to its original position after each breath.

The string attached to the armature is wrapped tight around the chest, with the motor placed right on the sternum, as shown in Fig. 3.2. Another wire is wrapped around the housing and chest to stabilize the motor, and ensures that only the armature turns during breaths.
When the servo motor is worn around the chest, the armature is positioned as shown in Fig. 3.3a. During inhalation, the chest circumference expands. The fixed ends of the string are pulled in opposite directions, thus turning the armature as well as stretching the rubber band, as shown in Fig. 3.3b. During exhalation, the rubber band pulls the armature back to its original position.

![Fig. 3.3: Top view of servo motor before inspiration (a) and during inspiration (b).](image)

The servo motor apparatus was tested on a single human subject during three different scenarios – normal breathing, fast breathing, and holding breath and exhaling. The subject is standing stationary. A Pneumotrace II piezoelectric chest belt was worn simultaneously as a respiratory effort reference. The voltage outputs of both the servo and the piezoelectric belt were recorded for all three scenarios, shown in Figs. 3.4-3.6 respectively. The output of the servo is the open-circuit voltage across the motor terminals. The positive half-cycles occur during inhalation, and the negative half-cycles occur during exhalation.
Fig. 3.4: Piezoelectric vs. servo motor output, normal breathing

Fig. 3.5: Piezoelectric vs. servo motor output, fast breathing
Fig. 3.6: Piezoelectric vs. servo motor output, holding breath then exhaling

The pulse-like waveform output of the servo-motor is due to static friction (stiction) in the motor’s gears. The positive and negative peaks of the servo motor align with those of the piezoelectric belt for all scenarios, showing that the servo motor can detect both inspirations and expirations accurately as compared to a respiratory reference.

During normal breathing, the servo outputs a peak-to-peak voltage of about 1.4V. The short-circuit current was approximated by measuring the voltage across a 1Ω resistor shunted across the servo motor terminals. The peak-to-peak short-circuit current measured is about 84.8mA, resulting in a peak power of 29.7mW.

The servo motor was able to generate significant harvested power from respiration, while sensing respiration accurately as compared to the piezoelectric chest belt [39]. However, the servo motor does need a very high force from respiration for the armature to turn, and a more ergonomic design could be implemented.

3.2 Improved Servo Motor Design

To improve the electromagnetic biosensor design, a smaller off-the-shelf servo motor with metal gears was modified into a chest belt. A smaller motor was chosen to increase comfort and decrease profile and weight as compared to the Futaba servo motor. A metal-gear servo was chosen because the stiction between lubricated cast iron materials
is lower than the stiction between nylon materials [40]. Metal gears also have more rotational inertia, which provides smoother turning of the armature and thus more continuous voltage outputs.

The metal-gear servo is a 9G EXI D213F [41], as shown in Fig. 3.7. This servo weighs only 9 grams, and its dimensions are 22.6mm x 11.4mm x 22.2mm. The servo is fixed and screwed down inside a trench cut into a wood block. The wood block dimensions are 7.5cm x 5.0cm x 1.25cm. The wood block serves to stabilize the servo against the chest and prevent any slipping or movement of the servo motor housing during operation. The backside of the wood block is covered with adhesive Velcro, shown in Fig. 3.8, which can attach to an elastic chest belt.
A ball-point pen spring is used as the restorative force. One end is fixed to the block, and the other end is attached to one end of the servo armature. Fishing wire is used to tie two loops through opposite ends of the armature. An adjustable, non-elastic belt with spring clips is clipped onto the loops. This non-elastic belt turns the armature to generate power. Fig. 3.9 shows the modified servo motor.

![Fig. 3.9: Top view of servo motor block](image)

The servo motor block is attached to an elastic Velcro belt, which is worn around the chest. The servo motor block is positioned right on the sternum. The non-elastic belt is also tied snug around the chest, over the elastic Velcro belt, with the armature positioned as shown in Fig. 3.10a.

During inspiration, the chest circumference expands. The ends of the non-elastic belt are pulled in opposite directions, thus turning the armature as well as stretching the spring, as shown in Fig. 3.10b. During expiration, the spring pulls the armature and non-elastic belt back to its original position, as shown in Fig. 3.10c. Fig. 3.11 shows the full servo motor belt, worn around the chest, with the servo motor positioned on the sternum.
The Pnuemotrace II piezoelectric chest belt was worn simultaneously with this servo motor on a single, stationary, standing subject, and both voltage outputs were recorded during slow, normal, and fast breathing. Figs. 3.12-3.14 show the voltage outputs for both the piezoelectric belt and servo motor for slow, normal, and fast breathing, respectively, for a single, stationary, standing subject. Like the Futaba servo, the 9G EXI servo is capable of detecting respiratory inspirations and expirations accurately as compared to the piezoelectric chest belt reference.

Voltage and current outputs were measured from the servo motor belt during normal breathing. Like the Futaba servo, the voltage output measured was the open-
circuit voltage, and the short-circuit current was approximated from the voltage output across a $1\Omega$ shunt resistor. During normal breathing, the servo outputs a peak-to-peak voltage of about 1.66V. The peak-to-peak short-circuit current measured is about 86mA. This results in a peak power of about 35.7mW, which is about 20% higher than the power output of the Futaba servo.

Fig. 3.12: Piezoelectric belt and servo motor belt voltage outputs, slow breathing.

Fig. 3.13: Piezoelectric belt and servo motor belt voltage outputs, normal breathing.
The 9G EXI metal gear servo is smaller, lighter, and has a slightly higher power output than the Futaba servo. The 9G EXI servo can also detect respiratory effort as accurately as the Futaba servo, as compared to the piezoelectric chest belt reference. However, the wooden block needed to stabilize the servo against the chest added to the weight of the overall device. The ball-point pen spring, which served as the restorative force, also needs to be fixed a couple centimeters away from the servo, in order to provide enough force to turn the armature back to its original position. This limits how small the overall device can be. The force needed to turn the armature of the 9G EXI servo, although less than that of the Futaba servo, is still quite high for normal respiration to overcome.

### 3.3 EM Biosensor Design with a DC Metal Gearmotor

Both the Futaba and 9G EXI servos generated high power outputs and accurate detection of inspirations and expirations during respiration. However, the force needed to turn the armatures for both servos were high, and the force from normal inspiration among a number of subjects would not be able to turn the armature consistently. To
overcome this limitation, DC metal gearmotors with different gear ratios were investigated.

The gear ratio of a motor directly affects the force needed to turn the armature. High gear ratios increase the voltage output of the motor, but require higher force to turn the armature. Low gear ratios require lower force to turn the armature, but the voltage output of the motor decreases. Thus, there is a balance between force and power output. If the gear ratio is too high, respiration alone will not be able to turn the armature, resulting in zero voltage. If the gear ratio is too low, respiration can turn the armature easier, but power output decreases. Unlike servos, DC metal gearmotors usually don’t have accompanying armatures or housing, so both would have to be fabricated.

Three Pololu DC Micro Metal Gearmotors, all the same size, with different gear ratios, were modified into wearable chest belts and tested to find an optimum motor for the EM biosensor. The motors are the Pololu #993 (30:1 gear ratio), Pololu #1098 (50:1 gear ratio), and the Pololu #992 (100:1 gear ratio) [42]. All three motors are 24mm x 10mm x 12mm, and about 9.6 grams. One of the motors is shown in Fig. 3.15a.

To modify the motor into a chest belt, the motor is fit into a plastic housing, fitted with a 2cm long plastic armature, and mounted onto a piece of hard felt to help stabilize the apparatus against the body. Elastic bands are attached across the Velcro strips on the top and bottom of the back piece and tied around the body. One end of adjustable, non-elastic wire is attached to a wing nut (fixed to the plastic housing), wrapped around the chest, and the other end is attached to the armature. A spring is attached between the armature and wing nut, and used to provide a restorative force to the armature during expirations. The wire is wrapped snugly around the chest, with the motor placed right on the sternum. The apparatus is tight enough to capture as much of the chest motion as possible, but not tight enough as to restrict the user’s breathing. Fig. 3.15b shows the modified motor, and Fig. 3.15c shows a subject wearing the modified motor.
The operation of the servo motor during respiration is shown in Fig. 3.16. During inspiration, the chest circumference expands. The fixed ends of the wire are pulled in opposite directions, thus turning the armature as well as stretching the spring. During expiration, the chest circumference retracts, and the spring pulls the armature back to its original position, and the process can repeat. Fig. 3.17 shows a top view of the chest, while wearing the EM biosensor, during respiration.
Fig. 3.16: EM biosensor operation before inspiration (a), during inspiration (b), and during expiration (c).

Fig. 3.17: Top view of EM biosensor on chest during respiration.
3.4 Measuring Static Friction

According to the maximum power transfer theorem, to obtain the maximum external power from a source with a finite internal resistance, the resistance of the load must equal the resistance of the source. For the selected motors, a load resistor should be chosen equal to the motor’s internal resistance. The internal resistances of the two servos and the three DC metal gearmotors are listed in Table 3.1.

Table 3.1: Internal resistances of selected motors.

<table>
<thead>
<tr>
<th>Motor</th>
<th>Internal Resistance (Ω)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Futaba S3003 Servo</td>
<td>6</td>
</tr>
<tr>
<td>9G EXI Servo</td>
<td>6</td>
</tr>
<tr>
<td>Pololu #993 (30:1)</td>
<td>16</td>
</tr>
<tr>
<td>Pololu #1098 (50:1)</td>
<td>16</td>
</tr>
<tr>
<td>Pololu #992 (100:1)</td>
<td>16</td>
</tr>
</tbody>
</table>

Though choosing a load-matching resistance should obtain the maximum power transfer, the force needed to turn the armature from rest (referred to as static friction, or stiction) greatly increases. As load resistance is increased, the stiction decreases. The stiction for all selected motors was measured with a force meter, as an increasing load resistance was applied. When the load resistance is about ten times the internal resistance, the stiction of the motor is at a minimum – equal to the stiction when no load resistance is connected (open-circuit). This applies to all selected motors. Fig. 3.18 shows the relationship between load resistance and stiction for the Pololu #1098 50:1 gearmotor.
Although increasing the load resistance decreases the stiction, the power transferred decreases as well. For testing purposes, all load resistances were selected to be ten times the motor’s internal resistance to provide minimum stiction, and thus minimum force for normal respiration to exert.

3.5 Mechanical Chest Design for EM Biosensor Testing

To get a comparable measurement of power output for the motors, a mechanical chest model was built to simulate respiration and provide a repeatable testbed. The mechanical chest is shown in Fig. 3.19.
The mechanical chest is built around a Griffin Motion MLS-050-BS-A-F-S-0-00 programmable linear stage, which oscillates the front of the chest to simulate respiration at the desired displacement and frequency [43]. The stage can travel 50mm with an accuracy of 5µm. Fig. 3.20 shows the top view of the mechanical chest. This figure also labels the sagittal displacement, the measurement from the back to the front of the chest, and the circumference, the measurement around the entire chest.
The three DC metal gearmotors were tested individually on the mechanical chest. The Futaba servo and the 9G EXI servo were also tested on the mechanical chest for power output comparisons. To simulate normal respiration, the linear stage was programmed to travel at 12 br/min (0.2Hz), at 1cm sagittal displacement. Fig. 3.21 shows the EM biosensor on the mechanical chest. Load resistors were attached between the terminals of all the motors. The load resistance chosen was ten times the internal resistance of the motor. The power output, $P$, for each motor was calculated using the voltage output, $V$, and the load resistance, $R_{load}$, shown in (3.1).

$$P = \frac{V^2}{R_{load}}$$  \hspace{1cm} (3.1)
Table 3.2 shows a comparison between the two servos and the three DC metal gearmotors. Included in the table are the gear ratio, open-circuit stiction, average power output, size, weight, and cost of all the motors.

In [44], the circumferential force from respiration was measured on a number of subjects. The average circumferential force from respiration was determined to be 0.46-0.65N. Thus, the optimum motor should have an open-circuit stiction in this range. This applies to the 30:1 and 50:1 motor. However, from the mechanical chest experiments, the average power output of the 30:1 motor was 0.46µW, which is comparable to the PZT power output mentioned in Chapter One. Thus, the 30:1 motor would not be an efficient respiratory energy harvester. The 50:1 motor has a slightly higher open-circuit stiction, but a much higher power output of 35.7µW. State-of-the-art system-on-chips can be powered with tens to hundreds of µW, so the 50:1 motor may be feasible as both a respiratory effort sensor and harvester. Thus, the 50:1 motor, with a load resistance of 160Ω, was selected for human testing.
Table 3.2: Motor characteristics comparison.

<table>
<thead>
<tr>
<th>Motor</th>
<th>Gear Ratio</th>
<th>Open-Circuit Stiction (N)</th>
<th>Average Power (µW)</th>
<th>Size (L x W x H) (mm)</th>
<th>Weight (g)</th>
<th>Cost (Motor only)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Futaba S3003 Servo</td>
<td>278:1</td>
<td>3.30</td>
<td>208.66</td>
<td>40 x 20 x 36</td>
<td>37</td>
<td>$25</td>
</tr>
<tr>
<td>9G EXI Servo</td>
<td>320:1</td>
<td>0.96</td>
<td>34.41</td>
<td>22.6 x 11.4 x 22.2</td>
<td>9</td>
<td>$10</td>
</tr>
<tr>
<td>Pololu #993</td>
<td>30:1</td>
<td>0.40</td>
<td>0.46</td>
<td>24 x 10 x 12</td>
<td>9.6</td>
<td>$10</td>
</tr>
<tr>
<td>Pololu #1098</td>
<td>50:1</td>
<td>0.50</td>
<td>35.70</td>
<td>24 x 10 x 12</td>
<td>9.6</td>
<td>$10</td>
</tr>
<tr>
<td>Pololu #992</td>
<td>100:1</td>
<td>0.84</td>
<td>55.23</td>
<td>24 x 10 x 12</td>
<td>9.6</td>
<td>$10</td>
</tr>
</tbody>
</table>
Chapter 4

Human Testing Protocol

The EM biosensor experiments were conducted according to CHS protocol number 19176. The protocol for this study was approved by the Committee on Human Studies (CHS), which is the unit designated to function as the federally mandated Institutional Review Board (IRB) for the University of Hawaii (UH) system. The protocol proposal, consent form, health history questionnaire, and IRB approval letter can be found in Appendices A-D.

4.1 Equipment and Instrumentation Set-up

All experiments were conducted in the Human Performance Laboratory with the Department of Kinesiology and Rehabilitation Science. Each subject was fitted with two EM biosensors, a Polar chest belt heart rate monitor, and a spirometer. The spirometer provided a reference for respiratory rate and tidal volume. One EM biosensor was centered on the subject’s sternum, and the belt was wrapped snug around the rib cage. Another EM biosensor was centered on the subject’s navel, and the belt was wrapped snug around the abdomen. During the tightening of the belt, the subjects were asked if they were uncomfortable or if the belt was too tight or loose. The general feedback from all the human subjects was positive, and that the EM biosensor was not uncomfortable.

4.2 Study Population

Twenty subjects volunteered to participate in this study. The subject body consists of thirteen males and seven females. The biological metrics of the subjects are shown in Table 4.1. The subjects did not receive any money reimbursement. The test procedures, along with potential risks, were explained and a signed consent form and completed
health history questionnaire was obtained from each subject prior to the start of every session.

Table 4.1: Descriptive statistics of the study sample

<table>
<thead>
<tr>
<th>Metric</th>
<th>Entire Sample (Mean ± Standard Deviation)</th>
<th>Males (Mean ± Standard Deviation)</th>
<th>Females (Mean ± Standard Deviation)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age (years)</td>
<td>27 ± 3.5</td>
<td>28 ± 3.4</td>
<td>26.1 ± 3.4</td>
</tr>
<tr>
<td>Height (cm)</td>
<td>174.4 ± 8</td>
<td>176.9 ± 6</td>
<td>169.8 ± 9.8</td>
</tr>
<tr>
<td>Body Mass (kg)</td>
<td>70.8 ± 9.9</td>
<td>75.2 ± 6.1</td>
<td>62.5 ± 10.6</td>
</tr>
<tr>
<td>BMI (kg/m²)</td>
<td>23.3 ± 3</td>
<td>24.1 ± 1.8</td>
<td>21.8 ± 4.2</td>
</tr>
</tbody>
</table>

4.3 Testing Procedure

The testing procedure involved five minutes with the subject sitting down, five minutes with the subject standing up, five minutes walking on a treadmill at 3.5mph, 1% grade, and five minutes walking at 2.5mph, 1% grade. Fig. 4.1 shows a single subject during the walking procedure. To determine if the EM biosensors would bestow any significant metabolic burden on users, each subject completed the entire testing procedure twice – once with the EM biosensors and once without the biosensors, with a twenty minute rest period in between. The difference in calculated energy expenditure with and without the EM biosensors confirmed there was no significant change in the metabolic rate of the subjects. Thus, the biosensors did not place any significant burden on the users. Verbal feedback from the subjects also confirmed that the biosensors were not uncomfortable or restrictive.

During the testing procedures, the voltage outputs of the EM biosensors are measured with a National Instruments 14-bit portable data acquisition (DAQ) device, recorded and digitized by LabView with a 1kHz sampling rate. Fig. 4.2 shows the voltage output from the EM biosensor on the chest of a single subject during the entire 20-minute testing period. The voltage outputs during both walking portions of the test are plotted together.
Fig. 4.1: Subject walking on treadmill during testing procedure.
Fig. 4.2: Voltage output of EM biosensor on chest of a single subject sitting (a), standing (b), and walking (c).
In cases where the subject is stationary, the voltage output may have some noticeable gaps between breaths, as shown in Fig. 4.3. This can be attributed to the stiction of motor being too high for respiration to overcome, thus resulting in zero voltage.

![EM biosensor output with undetected breaths.](image)

For cases where the subject is walking, the voltage output may show many more peaks than breaths taken, as shown in Fig. 4.4. This can be attributed to motion artifacts – the subject’s chest twisting, the subject’s arms having incidental contact with the chest belt, vibrations from stepping, etc.
Fig. 4.4: EM biosensor detecting more breaths than spirometer reference.
Chapter 5

Harvested Power

The power output was calculated by squaring the voltage output and dividing by the load resistance (3.1). The power output of an EM biosensor on the chest of a single subject is shown in Fig. 5.1. Power outputs are shown for the sitting, standing, and walking portions of the testing procedure.
Fig. 5.1: Power output from EM biosensor on chest of subject sitting (a), standing (b), and walking (c).

During the stationary portions of the test (sitting, standing), subjects are breathing normally with minimal effort. The average power output among the sample population during these two phases is between 6.87-12.68µW from the chest and between 3.58-5.70 µW from the abdomen.
During the exercise portion of the test (walking), subjects are breathing more frequently and with more effort, thus resulting in a larger power output compared to the stationary portions of the test. The average power output among the sample population during the walking phase is 71.91µW from the chest and 83.16µW from the abdomen.

The average power output per subject is shown in Fig. 5.2, for each of the three phases. Average power outputs from the chest and abdomen are also shown.
A summary of the average power output for the entire sample population is shown in Table 4.2. Low-power SoCs only need a couple μW to be powered on, so the average power being harvested, even during stationary periods, can accomplish that. To perform digitization, signal processing, and data transmission though, a longer harvesting period is needed to charge up a battery or capacitor. Chapter Eight explores the SoC power budget in greater detail, and investigates the feasibility of monitoring, given the power harvested.

**Table 5.2: Average power output for subject population.**

<table>
<thead>
<tr>
<th>Phase</th>
<th>Average Power Output, Chest (μW)</th>
<th>Average Power Output, Abdomen (μW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sitting</td>
<td>6.87</td>
<td>3.58</td>
</tr>
<tr>
<td>Standing</td>
<td>12.68</td>
<td>5.70</td>
</tr>
<tr>
<td>Walking</td>
<td>71.91</td>
<td>83.16</td>
</tr>
</tbody>
</table>
Chapter 6
Respiratory Rate

As shown in (2.5), an electromagnetic generator, when modified into a chest belt, will output a positive or negative voltage depending on if the chest circumference expanded (inspiration) or retracted (expiration). The inspirations and expirations can be identified by the corresponding sign of the voltage output. Respiratory rate can then be determined by the number of inspirations or expirations over a period of time.

Respiratory rate was detected with the EM biosensor on both the mechanical chest as well as during human testing. The linear stage and spirometer served as respiratory rate references, respectively. Respiratory rates measured from the references were compared to rates calculated from the EM biosensor voltage outputs during testing.

6.1 Respiratory Rate Detection on Mechanical Chest

The mechanical chest was programmed at different sagittal displacements as well as different frequencies, and respiratory rate was detected from the EM biosensor outputs. The voltage output from the EM biosensor was recorded at sagittal displacements of 1cm and 3cm, each at frequencies of 6, 12, and 30 breaths/min.

Fast Fourier Transform (FFT) was applied to determine the frequency of the EM biosensor voltage outputs. No filtering was done to the biosensor outputs. The maximum points were detected on the resulting periodograms to extract the frequency of the voltage outputs. The frequencies of the voltage outputs are compared to the programmed frequency of the mechanical chest model. The voltage outputs and respective periodograms are shown in Figs. 6.1-6.6, and a summary of the periodograms and detected frequencies are shown in Table 6.1.
Fig. 6.1: Voltage output (a) and periodogram (b) at sagittal displacement of 1 cm and frequency of 6 breaths/min.
Fig. 6.2: Voltage output (a) and periodogram (b) at sagittal displacement of 1cm and frequency of 12 breaths/min.
Fig. 6.3: Voltage output (a) and periodogram (b) at sagittal displacement of 1 cm and frequency of 30 breaths/min.
Fig. 6.4: Voltage output (a) and periodogram (b) at sagittal displacement of 3cm and frequency of 6 breaths/min.
Fig. 6.5: Voltage output (a) and periodogram (b) at sagittal displacement of 3cm and frequency of 12 breaths/min.
Fig. 6.6: Voltage output (a) and periodogram (b) at sagittal displacement of 3cm and frequency of 30 breaths/min.
Table 6.1: Periodogram comparison with mechanical chest programmed frequencies

<table>
<thead>
<tr>
<th>Displacement (cm)</th>
<th>Programmed Frequency (br/min)</th>
<th>Periodogram Detected Frequency (br/min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>6</td>
<td>6.0001</td>
</tr>
<tr>
<td>1</td>
<td>12</td>
<td>12.0002</td>
</tr>
<tr>
<td>1</td>
<td>30</td>
<td>30.0005</td>
</tr>
<tr>
<td>3</td>
<td>6</td>
<td>6.0001</td>
</tr>
<tr>
<td>3</td>
<td>12</td>
<td>12.0002</td>
</tr>
<tr>
<td>3</td>
<td>30</td>
<td>30.0005</td>
</tr>
</tbody>
</table>

On the mechanical chest, all simulated breaths are detected. The series of peaks in each inspiration and expiration are caused by the stiction of the motor’s gears, as well as the cogging effect. At a displacement of 3cm and a frequency of 30 breaths/min, the voltage output is continuous, rather than a series of peaks. At this displacement and frequency, the motor armature is being pulled fast enough and forcefully enough to greatly overcome the gear stiction, resulting in a smoother armature rotation and a smoother, continuous voltage output. The mean absolute error is 0.00027 breaths/min with a standard deviation of 0.00019 breaths/min, which shows that the EM biosensor can detect respiratory rate very accurately as compared to a mechanical model.

6.2 Respiratory Rate Detection for Human Testing

A spirometer provided a respiratory rate reference during human testing. The average respiratory rate was measured from the spirometer every thirty seconds, and was compared to the average respiratory rate detected from the EM biosensor.

As mentioned in Chapter 4, motion artifacts were present in the voltage output of the EM biosensor, particularly during the walking phases of testing. To remove motion artifacts, all voltage outputs were filtered with a low-pass, finite impulse response (FIR) filter. The corner frequency is 0.5Hz (30 breaths/min) and the cutoff frequency is 0.6667Hz (40 breaths/min). The group delay (7.5s) was also compensated for in the filtered outputs. Fig. 6.7 shows a comparison of the periodograms for filtered and
unfiltered EM biosensor outputs, from the chest of a single subject walking. FFT was applied to a 30s period during the 3.5mph walking phase. The comparison shows that before filtering, a respiratory rate of 122 br/min is detected from the EM biosensor. Clearly, there is a large error, which can be attributed to motion artifacts. After filtering, the detected respiratory rate is 14 br/min, which is the same rate as measured by the spirometer for that time period.
An FFT sliding window (spectrogram) was applied to the filtered voltage outputs, and the resulting spectrogram frequencies are compared with the spirometer-measured average respiratory rate. The FFT window size is 30 seconds, and the window is slid at 0.5 second increments. Fig. 6.8 shows the resulting spectrograms and spirometer averages for the sitting, standing, and walking procedures for a single subject.
EM Biosensor vs. Spirometer, 30s Resp. Rate Comparisons - Subject Sitting

EM Biosensor vs. Spirometer, 30s Resp. Rate Comparisons - Subject Standing
6.3 Results and Error Analysis

The absolute error was calculated as the difference between the spectrogram frequency and spirometer’s average respiratory rate at the 30-second marks. The mean absolute errors and standard deviations are calculated for all 20 subjects and recorded in Tables 6.2-6.4, and plotted in Fig. 6.9.
Table 6.2: EM biosensor vs. spirometer respiratory rate mean absolute error and standard deviation, subjects sitting.

<table>
<thead>
<tr>
<th>Subject #</th>
<th>Mean (br/min)</th>
<th>Standard Deviation (br/min)</th>
<th>Subject #</th>
<th>Mean (br/min)</th>
<th>Standard Deviation (br/min)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td>0.93</td>
<td>1.75</td>
<td>16</td>
<td>2.73</td>
<td>2.98</td>
</tr>
<tr>
<td>4</td>
<td>1.86</td>
<td>1.32</td>
<td>17</td>
<td>13.92</td>
<td>4.04</td>
</tr>
<tr>
<td>6</td>
<td>2.40</td>
<td>4.18</td>
<td>18</td>
<td>1.29</td>
<td>1.41</td>
</tr>
<tr>
<td>7</td>
<td>3.61</td>
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Table 6.3: EM biosensor vs. spirometer respiratory rate mean absolute error and standard deviation, subjects standing.

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<th>Standard Deviation (br/min)</th>
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Table 6.4: EM biosensor vs. spirometer respiratory rate mean absolute error and standard deviation, subjects walking.

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<td>2.26</td>
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</table>

Fig. 6.9: Respiratory Rate Mean Absolute Error.

The highest respiratory rate errors were due to missed breaths for some seated or standing subjects. Subjects with an average absolute error greater than 5 breaths/min
were excluded from the Bland-Altman analysis. Individual Bland-Altman plots are shown for the sitting, standing, and walking cases in Fig. 6.10.
Fig. 6.10: Bland-Altman plots for sitting (a), standing (b), and walking (c).

For the sitting procedure during human testing, respiratory rate was detected for human subjects with a mean difference of 0.36 breaths/min and a standard deviation of 2.83 breaths/min. The respiratory rates detected by the EM biosensor are between 5.18 breaths/min below and 5.90 breaths/min above the rate measured by the spirometer with 95% confidence. For the standing procedure, the mean difference was 0.23 breaths/min with a standard deviation of 2.64 breaths/min. The respiratory rates detected by the EM biosensor are between 4.93 breaths/min below and 5.40 breaths/min above the rate measured by the spirometer with 95% confidence. For the walking procedure, the mean difference was 0.48 breaths/min with a standard deviation of 3.06 breaths/min. The respiratory rates detected by the EM biosensor are between 5.52 breaths/min below and 6.48 breaths/min above the rate measured by the spirometer with 95% confidence. This indicated a high level of agreement between the respiratory rate measurements done by the EM biosensor and the spirometer.

6.4 Discussion

The EM biosensor has been shown to detect respiratory rate accurately as compared to a mechanical chest and spirometer reference. The EM biosensor was tested
on a mechanical chest with programmable displacement and frequency, as well as on 20 human subjects during sitting, standing, and walking procedures.

On the mechanical chest simulator, respiratory rate was detected with a mean absolute error of 0.00027 breaths/min and a standard deviation of 0.00019 breaths/min, as compared to the mechanical chest’s programmed frequency. All simulated breaths were detected with no motion artifacts, and no filtering was done on the biosensor outputs.

For human testing, the average absolute error was 1.69 ± 2.29 br/min for subjects sitting, 1.66 ± 2.06 br/min for subjects standing, and 2.08 ± 2.29 br/min for subjects walking. The mean difference between the detected respiratory rates for the sitting, standing, and walking cases is 0.36, 0.23, and 0.48 breaths/min, respectively. The 95% confidence interval ranges from -5.18 to 5.90 breaths/min, -4.93 to 5.40 breaths/min, and -5.52 to 6.48 breaths/min for the sitting, standing, and walking cases, respectively. This indicates a high level of agreement between the respiratory rate measurements done by the EM biosensor and the spirometer.

Higher absolute error was apparent during sitting and standing phases, due to the EM biosensor missing some breaths. In these cases, the motor’s stiction was greater than the force of inspiration. During the walking phase, most, if not all, breaths are detected, thus improving the accuracy of the biosensor.

The EM biosensor, therefore, can be used during physical activity if motion artifacts are filtered. To improve the accuracy of the biosensor on stationary subjects, armature length and gear ratio can be modified to increase sensitivity, while optimizing size and output power. The armature length should at minimum be equal to the expected circumferential change. Increasing the armature length would decrease the required force by the same factor, but would also increase the size of the biosensor and decrease angular displacement. For the same circumference change, angular displacement would decrease by the same factor as the armature length increase. Lower angular displacement results in lower angular velocity, and thus a lower voltage output from the biosensor.
Capnography (carbon dioxide concentration sensors) and acoustic sensors [16] have been demonstrated to be very accurate in sensing respiratory rate (+/- 1 br/min). Sensors that don’t measure airflow directly, such as thoracic impedance sensors, have accuracies of about +/- 3 br/min. The EM biosensor is not as accurate as the airflow sensors, but the accuracy can still be improved. Variable gear trains could possibly be implemented to adjust the sensitivity of the EM biosensor per non-physical or physical activity.
Chapter 7
Tidal Volume

Tidal volume has been demonstrated to be directly proportional to chest (rib cage) circumference change [24, 25]. Thus tidal volume can be detected by calculating the quantitative chest circumference change during respiration. According to (2.8), chest circumference change is a linear function of the integral of the electromagnetic generator output. The coefficients $R$ and $C_0$ can be calculated from the voltage integrals of two known circumference changes.

7.1 Calculating Chest Circumference

(2.8) was verified using the mechanical chest. Accurate circumference measurements can be made as the sagittal displacement of the mechanical chest is changed. To determine the coefficients $R$ and $C_0$ in (2.8), the voltage integral was calculated at sagittal displacements of 0.4cm and 3.0cm, at 12 br/min. 0.4cm is the minimum displacement at which the EM biosensor outputs visible voltage peaks. 3.0cm is an estimate of the maximum sagittal displacement during respiration. Circumference measurements were made with tape measure. Fig. 7.1 shows the mechanical chest circumference as the sagittal displacement increases from the minimum position of the linear stage. Each inspiration, denoted by the positive voltage peaks, is integrated, and the average value for all inspirations over one minute is calculated. An illustration of this is shown in Fig. 7.2.
Fig. 7.1: Mechanical chest circumference vs. sagittal displacement.

Fig. 7.2: Identified inspirations from EM biosensor output.
At 0.4 cm displacement, the measured circumference change of the mechanical chest is 0.55 cm, and the average integral of the voltage during inspiration is 0.0077 V·s. At 3 cm displacement, the measured circumference change of the mechanical chest is 3.6 cm, and the average integral of the voltage during inspiration is 0.240 V·s. Using these values, the coefficients $R$ and $C_0$ were calculated and shown in (7.1) and (7.2).

$$R = 13.1239 \frac{cm}{V \cdot s} \quad (7.1)$$

$$C_0 = 0.034208 V \cdot s \quad (7.2)$$

Voltage outputs were recorded at different sagittal displacements ranging from 0.4 cm to 3 cm, at a frequency of 12 br/min. The integrals are calculated for each case, and the circumference changes are calculated and compared with the measured circumferences. The sagittal displacements, voltage integral values, measured circumference changes, calculated circumference changes, and absolute errors are recorded in Table 7.1. The measured and calculated circumference changes are plotted in Fig. 7.3. The absolute error for all cases is less than 1 mm.
Table 7.1: Calculated circumference changes vs. measured circumference changes, 12 br/min.

<table>
<thead>
<tr>
<th>Disp. (cm)</th>
<th>[\int V \cdot dt \ (V \cdot s)]</th>
<th>Measured Circumf. Change (cm)</th>
<th>Calculated Circumf. Change (cm)</th>
<th>Absolute Error (cm)</th>
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<td>0.19487</td>
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</table>

Fig. 7.3: Measured vs. calculated chest circumference changes.

According to (2.8), the integral of the voltage, with respect to time, should be constant regardless of the frequency. The voltage outputs are recorded at 1, 2, and 3cm
sagittal displacements, each at 6, 12, 18, and 30 br/min. Voltage integrals are calculated for each case. At different frequencies, the voltage integrals for different circumference changes are fairly consistent. Although the voltage integrals vary as frequency changes, the absolute error for all cases all remain under 1mm.

Table 7.2: Calculated circumference changes vs. measured circumference changes at different frequencies.

<table>
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<tr>
<th>Disp. (cm)</th>
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<th>Calculated Circumf. Change (cm)</th>
<th>Absolute Error (cm)</th>
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The results of the tests on the mechanical chest show that the chest circumference change is linearly related to the integral of the EM biosensor voltage output. The EM biosensor was tested on the mechanical chest at different sagittal displacements and different frequencies, and the voltage integrals were calculated for all inspirations and averaged over one minute for each case. Circumference change was calculated with these integral values, and compared with the measured circumference of the mechanical chest. All tested cases have an absolute error below 1mm. The average absolute error between calculated circumference changes and measured circumference changes for all tests is
0.54mm ± 0.36mm. This shows that the EM biosensor can detect chest circumference change accurately as compared to a mechanical circumference model.

### 7.2 Detecting Tidal Volume

Tidal volume can be calculated as a linear function of chest circumference change, as shown in (2.9). Section 7.1 shows that chest circumference change can be calculated accurately from the integral of the voltage output of the EM biosensor. (2.9) shows that the chest circumference change can be converted to tidal volume by a conversion coefficient $K$, which is in units of mL/cm. This coefficient $K$ can be found by correlating the tidal volume values, measured by the spirometer, to the corresponding calculated chest circumference changes, measured by the EM biosensor. Fig. 7.4 shows a portion of the EM biosensor output on a single subject during the sitting phase. The voltage peak groups synchronized with the spirometer-detected breaths are integrated to calculate the chest circumference change during these breaths. The calculated chest circumference changes are then compared to the corresponding tidal volumes measured by the spirometer.

![Fig. 7.4: Correlating EM biosensor voltage with spirometer-measured tidal volumes.](image)

Fig. 7.4: Correlating EM biosensor voltage with spirometer-measured tidal volumes.
Fig. 7.5 shows a plot relating chest circumference change to tidal volume for a single subject walking. For each subject in each testing phase (sitting, standing, walking), an average conversion coefficient $K$ was determined. The coefficients for every subject are recorded in Tables 7.3-7.5. For some subjects, less than 10 breaths were detected by the EM biosensor for an entire phase, including sitting, standing, or both. Conversion coefficients were not calculated for these phases. Conversion coefficients were calculated for all subjects during the walking phase, though.

![Tidal Volume Vs. Chest Circumference Change, Subject Walking](image)

Fig. 7.5: Tidal volume vs. chest circumference change, subject walking.
Table 7.3: Coefficients converting chest circumference change to tidal volume, subjects sitting.

<table>
<thead>
<tr>
<th>Subject #</th>
<th>Coefficient K (mL/cm)</th>
<th>Subject #</th>
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Table 7.4: Coefficients converting chest circumference change to tidal volume, subjects standing.

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<th>Subject #</th>
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</table>
Table 7.5: Coefficients converting chest circumference change to tidal volume, subjects walking.

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<th>Coefficient K (mL/cm)</th>
<th>Subject #</th>
<th>Coefficient K (mL/cm)</th>
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</table>

7.3 Results and Error Analysis

Using the calculated chest circumference changes and conversion coefficients, tidal volume was calculated from the EM biosensor output and compared to the spirometer-measured tidal volumes. The mean absolute errors and standard deviations for all 20 subjects are recorded in Tables 7.6-7.8 for sitting, standing, and walking phases, respectively, and are also plotted in Fig. 7.5.
Table 7.6: EM biosensor vs. spirometer tidal volume mean absolute error and standard deviation, subjects sitting.

<table>
<thead>
<tr>
<th>Subject #</th>
<th>Mean (%)</th>
<th>Standard Deviation (%)</th>
<th>Subject #</th>
<th>Mean (%)</th>
<th>Standard Deviation (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td>41.4</td>
<td>21.8</td>
<td>16</td>
<td>38.2</td>
<td>40.1</td>
</tr>
<tr>
<td>4</td>
<td>88.4</td>
<td>191.2</td>
<td>17</td>
<td></td>
<td></td>
</tr>
<tr>
<td>6</td>
<td>30.9</td>
<td>17.8</td>
<td>18</td>
<td>65.1</td>
<td>103.2</td>
</tr>
<tr>
<td>7</td>
<td>27.7</td>
<td>21.0</td>
<td>19</td>
<td></td>
<td></td>
</tr>
<tr>
<td>9</td>
<td>32.0</td>
<td>45.6</td>
<td>20</td>
<td>35.3</td>
<td>47.5</td>
</tr>
<tr>
<td>10</td>
<td></td>
<td></td>
<td>21</td>
<td>20.3</td>
<td>23.8</td>
</tr>
<tr>
<td>12</td>
<td>63.0</td>
<td>66.8</td>
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<td>41.9</td>
<td>38.5</td>
</tr>
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<td>13</td>
<td>38.0</td>
<td>29.1</td>
<td>23</td>
<td></td>
<td></td>
</tr>
<tr>
<td>14</td>
<td>43.3</td>
<td>68.0</td>
<td>24</td>
<td>24.3</td>
<td>26.1</td>
</tr>
<tr>
<td>15</td>
<td></td>
<td></td>
<td>25</td>
<td>45.7</td>
<td>32.0</td>
</tr>
</tbody>
</table>

Table 7.7: EM biosensor vs. spirometer tidal volume mean absolute error and standard deviation, subjects standing.

<table>
<thead>
<tr>
<th>Subject #</th>
<th>Mean (%)</th>
<th>Standard Deviation (%)</th>
<th>Subject #</th>
<th>Mean (%)</th>
<th>Standard Deviation (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3</td>
<td></td>
<td></td>
<td>16</td>
<td>28.5</td>
<td>16.6</td>
</tr>
<tr>
<td>4</td>
<td>35.5</td>
<td>29.3</td>
<td>17</td>
<td></td>
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</tr>
<tr>
<td>6</td>
<td>53.2</td>
<td>31.7</td>
<td>18</td>
<td>21.5</td>
<td>17.1</td>
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<td>38.7</td>
<td>24.9</td>
<td>19</td>
<td>37.3</td>
<td>28.2</td>
</tr>
<tr>
<td>9</td>
<td>23.4</td>
<td>16.9</td>
<td>20</td>
<td>24.4</td>
<td>16.7</td>
</tr>
<tr>
<td>10</td>
<td>76.5</td>
<td>53.1</td>
<td>21</td>
<td>35.9</td>
<td>29.8</td>
</tr>
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<td>12</td>
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<tr>
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<td>38.0</td>
<td>38.1</td>
<td>23</td>
<td>38.3</td>
<td>40.5</td>
</tr>
<tr>
<td>14</td>
<td>38.4</td>
<td>21.1</td>
<td>24</td>
<td>19.3</td>
<td>28.3</td>
</tr>
<tr>
<td>15</td>
<td></td>
<td></td>
<td>25</td>
<td>34.1</td>
<td>31.1</td>
</tr>
</tbody>
</table>

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Table 7.8: EM biosensor vs. spirometer tidal volume mean absolute error and standard deviation, subjects walking.

<table>
<thead>
<tr>
<th>Subject #</th>
<th>Mean (%)</th>
<th>Standard Deviation (%)</th>
<th>Subject #</th>
<th>Mean (%)</th>
<th>Standard Deviation (%)</th>
</tr>
</thead>
<tbody>
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<td>36.6</td>
<td>33.8</td>
<td>18</td>
<td>29.3</td>
<td>22.4</td>
</tr>
<tr>
<td>7</td>
<td>24.8</td>
<td>34.1</td>
<td>19</td>
<td>26.3</td>
<td>21.5</td>
</tr>
<tr>
<td>9</td>
<td>45.7</td>
<td>40.3</td>
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<td>10</td>
<td>44.3</td>
<td>48.6</td>
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<td>54.7</td>
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<td>30.9</td>
<td>30.0</td>
<td>23</td>
<td>45.5</td>
<td>29.7</td>
</tr>
<tr>
<td>14</td>
<td>35.8</td>
<td>87.3</td>
<td>24</td>
<td>19.1</td>
<td>19.6</td>
</tr>
<tr>
<td>15</td>
<td>32.9</td>
<td>23.4</td>
<td>25</td>
<td>35.0</td>
<td>21.3</td>
</tr>
</tbody>
</table>

Fig. 7.5: Tidal Volume Mean Absolute Error.

Errors could be attributed to the EM biosensor slipping or loosening during testing, especially during the walking phase. This would reduce or change the amount of
motion translated to the EM biosensor, resulting in chest circumference errors and consequent tidal volume errors. Implementing a shoulder strap(s) to the EM biosensor could keep the chest belt from slipping down. High friction, durable elastic bands (to keep the EM biosensor stabilized against the body) could also be used to prolong the bands from stretching out, as well as keeping the device from slipping.

Subjects with mean absolute errors above 300mL were excluded from the Bland-Altman analysis. Individual Bland-Altman plots are shown for the sitting, standing, and walking cases in Fig. 7.5.
For the sitting phase during human testing, tidal volume was detected for human subjects with a mean difference of 1.43mL and a standard deviation of 193.7mL. The tidal volumes detected by the EM biosensor are between 378.30mL below and 381.17mL above the volume measured by the spirometer with 95% confidence. For the standing phase, the mean difference was -8.43mL with a standard deviation of 211.6mL. The tidal volumes detected by the EM biosensor are between 423.22mL below and 406.35mL above the volume measured by the spirometer with 95% confidence. For the walking phase, the mean difference was -21.16mL with a standard deviation of 225.0mL. The tidal volumes detected by the EM biosensor are between 462.25mL below and 419.93mL above the volume measured by the spirometer with 95% confidence. This indicates a high level of agreement between the tidal volume measurements done by the EM biosensor and the spirometer.

7.4 Discussion

The EM biosensor has been shown to detect tidal volume accurately as compared to a spirometer reference. The EM biosensor was tested on 20 human subjects, and the voltage output was used to calculate chest circumference change and consequently tidal
volume. The calculated tidal volumes were compared to the tidal volumes measured by the spirometer.

For human testing, the mean difference between the detected tidal volumes for the sitting, standing, and walking cases is 1.4, -8.4, and -21.2mL, respectively. The 95% confidence interval ranges from -378.30mL to 381.17mL, -423.22mL to 406.35mL, and -462.25mL to 419.93mL for the sitting, standing, and walking cases, respectively. This indicates strong agreement between the tidal volume measurements done by the EM biosensor and the spirometer.

Higher absolute error was apparent during sitting and standing phases, due to the EM biosensor missing some breaths. In these cases, the motor’s stiction was greater than the force of inspiration. During the walking phase, most, if not all, breaths are detected, thus improving the accuracy of the biosensor.

Current standards of spirometry require errors to be no greater than +/-3.5% [45]. For the human testing population, the mean absolute errors were 31.8%, 31.6%, and 31.2% for sitting, standing, and walking, respectively. Thus, there is significant room for accuracy improvement. As mentioned previously, errors could be corrected by securing the biosensor placement on the torso. This should create more consistency between the circumference changes and resulting voltage outputs.

From subject to subject, however, chest circumference change and tidal volume will be related by different conversion coefficients. To determine the coefficient for a particular subject, the electromagnetic generator will have to be calibrated with a tidal volume reference, such as a spirometer. For example, a subject would be fitted with both the electromagnetic generator and spirometer for thirty seconds. The conversion coefficient, which is the ratio between tidal volume and chest circumference change, could be calculated as the ratio between the standard deviations of the spirometer and integral of the EM biosensor voltage output during the thirty second calibration period [46].
Chapter 8
Continuous and Remote Monitoring
Feasibility

The SoC chosen for the overall study was Texas Instrument’s (TI) CC430 chip, which combines the company’s low-power MSP430 microcontroller with its low-power CC1101 sub-1-GHz RF transceiver [47]. The evaluation kit came with two EM430F6137RF900 evaluation boards and two 868/915 MHz antennas. Informational materials, such as the CC430’s datasheet and user’s guide, were found on the CC430’s product page [48].

To realize the zero-net energy biosensor concept, the system-on-chip (SoC) used for monitoring should operate under minimum power consumption, and the harvested power from respiration should be sufficient enough for the SoC to sense and transmit information at reasonable intervals.

8.1 System-on-Chip Power Consumption
Specifications

To sense respiratory rate and tidal volume, the clock, analog-to-digital converter (ADC), central processing unit (CPU), and wireless link must be implemented. For the Texas Instruments CC430, the minimum power requirements for the aforementioned components are summarized in Table 8.1. The power requirements are calculated using the minimum power supply voltage, 2.2V. In [49], the time to transmit respiratory rate data is about 3.6ms. Respiratory rate can be represented by 7 bits (rates up to 127 br/min can be represented), and tidal volume, in milliliters, can be represented by 13 bits (volumes up to 8095mL can be represented). Thus, the time to transmit tidal volume data
was estimated to be about twice that of respiratory rate, or 7.2ms. The total time to transmit both respiratory rate and tidal volume data would then be about 10.8ms.

Table 8.1: CC430 SoC power requirements.

<table>
<thead>
<tr>
<th>Component</th>
<th>Operation Time</th>
<th>Power Requirement (µW)</th>
<th>Energy Requirement (mJ)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Low-Power Mode</td>
<td>Continuous during harvesting periods</td>
<td>2.2</td>
<td>0.132/min</td>
</tr>
<tr>
<td>(RAM retention)</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Clock (32.768kHz)</td>
<td>Continuous during sensing periods</td>
<td>6.6</td>
<td>0.396/min</td>
</tr>
<tr>
<td>ADC</td>
<td>Continuous during sensing periods</td>
<td>275</td>
<td>16.5/min</td>
</tr>
<tr>
<td>CPU</td>
<td>Continuous during sensing periods</td>
<td>352</td>
<td>21.12/min</td>
</tr>
<tr>
<td>TX</td>
<td>Once per sensing interval</td>
<td>31,240</td>
<td>0.3374/TX</td>
</tr>
</tbody>
</table>

8.2 Harvested Power Vs. SoC Power Requirements

The average power harvested from the EM biosensor determines how long it takes to store enough energy to carry out the SoC functions. The average power harvested from the mechanical chest and from human testing is shown in Table 8.2.

Continuous monitoring (24/7) is not necessary for most applications. Sensing and transmitting data at periodic intervals is more feasible regarding power constraints and necessary data. Thus, the power harvested does not have to be equal to or greater than the total SoC power requirement on a second-to-second basis. However, the power harvested does need to be greater than 2.2µW, the power required for the SoC to remain in low-power mode.
The EM biosensor can harvest and store energy when the SoC is in low-power mode. The SoC can be powered on, sense and transmit data when needed, then revert back to low-power mode.

Table 8.2: Average power harvested on mechanical chest and during human testing.

<table>
<thead>
<tr>
<th>Testing Scenario</th>
<th>Average Power Harvested (µW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mechanical Chest</td>
<td>35.695</td>
</tr>
<tr>
<td>1.0cm sagittal displacement, 12 br/min</td>
<td></td>
</tr>
<tr>
<td>Sitting</td>
<td>10.45</td>
</tr>
<tr>
<td>Standing</td>
<td>18.38</td>
</tr>
<tr>
<td>Walking</td>
<td>155.07</td>
</tr>
</tbody>
</table>

For each of the testing scenarios in Table 8.2, the ratio between required harvesting time and sensing time was calculated. For one minute of sensing and one data transmission, the SoC requires 38.016mJ. The SoC consumes 0.132mJ during low-power mode. Table 8.3 shows the amount of time the EM biosensor needs to harvest energy in order for the SoC to collect respiratory data for one minute and perform one data transmission. Calculations are based on assumptions that all energy harvested can be used for the SoC, and that the physical activity of the subject is constant during harvesting periods.
Table 8.3: Harvesting Time Required per 1 Minute of Monitoring and 1 Data Transmission

<table>
<thead>
<tr>
<th>Testing Scenario</th>
<th>Energy Collected Over 1 Minute (mJ)</th>
<th>Harvesting Time Required (minutes)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mechanical Chest</td>
<td>2.1417</td>
<td>18</td>
</tr>
<tr>
<td>Sitting</td>
<td>0.627</td>
<td>76</td>
</tr>
<tr>
<td>Standing</td>
<td>1.1028</td>
<td>38</td>
</tr>
<tr>
<td>Walking</td>
<td>9.3042</td>
<td>3</td>
</tr>
</tbody>
</table>

Even during periods without major physical activity (sitting, standing), enough respiratory energy can be harvested to sense and transmit information once or twice an hour, which is a reasonable interval for most applications. Data transmission may need to occur more than once or twice an hour, in which harvesting efficiency would need to be increased and SoC power consumption would need to be decreased. Considerations for power optimization are mentioned in Chapter Nine under future work.

The time needed for harvesting, \( t_{\text{harvest}} \) (min), can be generalized based on the desired time of sensing, \( t_{\text{sense}} \) (min), the desired number of transmissions per sensing interval, \( X \), and the average power harvested, \( P_{\text{avg}} \) (µW). The equation for this generalization is shown in (8.1). Fig 8.1 shows a plot of the harvesting time needed per one minute of sensing and one transmission, depending on the average power harvested.

\[
t_{\text{harvest}} = \frac{0.6336(t_{\text{sense}}) + 0.005623(X) - (P_{\text{avg}})(t_{\text{sense}})}{(P_{\text{avg}}) - 0.0022} \tag{8.1}
\]
Fig. 8.1: Harvesting time needed per one minute of sensing and one transmission, for average power harvested.
Chapter 9
Conclusion

Respiratory monitoring has traditionally been done with devices that are uncomfortable and/or immobilize patients. Wearable technology can accurately measure important physiological signals while offering patient comfort and mobility, but their reliance on batteries can make continuous, remote monitoring inconvenient, difficult, or impossible. Harvested respiratory energy can provide power for sensing and data transmission, thus creating a zero-net energy biosensor.

The work in this dissertation has proven the ability to develop an electromagnetic biosensor capable of monitoring respiratory rate and tidal volume, while harvesting significant respiratory energy for a low-power system-on-chip. Testing was performed on both a constructed mechanical chest as well as on 20 human subjects.

The respiratory rates detected from the EM biosensor had a mean difference of $0.36 \pm 2.83$ br/min, $0.23 \pm 2.64$ br/min, and $0.48 \pm 3.06$ br/min for subjects sitting, standing, and walking, respectively.

To detect tidal volume, chest circumference change was extracted from the EM biosensor output. On the mechanical chest, the circumference change detected by the EM biosensor had an absolute error of $0.54 \pm 0.36$mm as compared to measured values of the circumference change using tape measure.

The tidal volumes detected by the EM biosensor had a mean difference of $1.4 \pm 193.76$mL, $-8.4 \pm 211.61$mL, and $-21.2 \pm 225.07$mL for subjects sitting, standing, and walking, respectively.

Across the 20-subject population, an average of 10, 18, and $155\mu W$ was harvested from both the chest and abdomen during sitting, standing, and walking phases respectively. The amount of power to operate the system-on-chip (CC430) was estimated, and harvesting times were determined per minute of respiratory monitoring and data transmission. During stationary phases (sitting and standing), 38-76 minutes of harvesting are needed to do one minute of monitoring and one data transmission. During the walking
phase, only 3 minutes of harvesting is needed for one minute of monitoring and one data transmission.

9.1 Future Work

Respiratory rates and tidal volumes were detected solely from the EM biosensor on the chest (rib cage) of the human subjects. During human testing, an extra EM biosensor was placed on the abdomen for increased harvested power, as well as an abdominal respiratory signal. Further signal processing could determine if the abdominal EM biosensor would increase the accuracy of respiratory rate and tidal volume detection.

To obtain the maximum transferrable power, a load-matching resistor should be connected across the motor terminals; however, this significantly increases the stiction of the motor. The load resistance chosen was ten times the motor’s internal resistance. This minimized the stiction, but resulted in 33% of the maximum transferrable power. A motor with inherently lower stiction could be selected or designed, so that connecting a load-matching resistor wouldn’t increase the motor stiction beyond that of the force of normal respiration. This would result in the maximum power transferrable, while still maintaining stiction low enough for normal respiration to overcome.

The series of peaks in the voltage output of the EM biosensor are caused by the stiction of the motor’s gears, as well as the cogging effect. To alleviate cogging, a motor could be designed with skewed rotors, so that the rotor teeth cross the stator pole at a slant instead of a straight line, making the effect more gradual [36]. Using a coreless motor would eliminate cogging. Non-cogging designs would result in a more continuous voltage output, rather than series of voltage peaks. This may increase the power output efficiency, but may also decrease the voltage amplitude. If voltage amplitudes are too low, rectification of the voltage output may be difficult or impossible. The theory of respiratory information detection developed in this dissertation may still be used on non-cogging motors.

The comparison between harvested power and the SoC’s power requirements show that during physical activity, sensing and data transmissions can be just a few minutes apart. During nonphysical activity, the harvesting time greatly increases. By
arraying more generators down the torso, more movement can be translated and thus more power can be harvested. More generators on the body would mean more burden on the subject, so metabolic cost testing could be carried out to verify if the addition of generators will have any significant impact.

The SoC power requirements could also be minimized. All specifications listed are based on the chip’s built-in features. Lower frequency clocks could be utilized to reduce the power consumption of both the clock operation and CPU. Accurate power measurements would need to be done with real EM biosensor signals.

Tidal volume was found to be a linear function of the integral of the voltage output of the EM biosensor. Integrating hardware (resistor and capacitor with an operational amplifier) could be used to integrate the voltage output before signal digitization. This could reduce the computational power needed to digitally integrate the signal. However, the op-amp integrator is not passive, and voltage supplies need to be applied for the rails of the op-amp. This may still consume less power than digitally integrating the signal, but the power comparison between the two options would need to be explored in detail.

Linear EM generators are more parallel to the circumferential motion of the chest during respiration, and would therefore have a lower profile and be more ergonomic than rotational generators. Custom linear EM generators could be fabricated and integrated into a chest belt or shirt. The theory for detecting respiratory rate and tidal volume could still be applied to these fabricated generators.

### 9.2 Summary

Contributions of this dissertation work to electrical engineering include the design of an electromagnetic respiratory effort harvester, the first human study data of respiratory rate and tidal volume detection using electromagnetic biosensors, and an investigation of simultaneous harvesting and sensing feasibility with a low-power system-on-chip. The study achieved the following objectives:
1. Designing an inexpensive, ergonomic electromagnetic biosensor capable of harvesting and sensing respiratory effort.

2. Detecting respiratory rate from the electromagnetic biosensor voltage output.

3. Detecting tidal volume from the electromagnetic biosensor voltage output.

4. Investigating the feasibility of continuous and remote respiratory monitoring with the electromagnetic biosensor and a low-power SoC.

Zero-net energy respiratory biosensing can be possible with efficient harvesting materials and techniques, as well as minimum power utilization on the sensing, processing, and data transmission side. The development of zero-net energy biosensing can be very beneficial for continuous and remote respiratory activity monitoring. Respiratory monitoring can be useful for patients with breathing problems, aiding in the diagnosis of disease, and evaluating the response to medication that affects the respiratory system. Patients with chronic diseases, elderly citizens living at home alone, and those working under heavy loads, including firefighters and military personnel, could benefit from continuous and remote respiratory monitoring. Further physical minimization and ergonomic optimization could allow for easier and unobtrusive respiratory monitoring on infants, children, and animals as well. Integration into clothing could lead to self-powered wearable technology for patients in the near future.
Appendix A

CHS Protocol No. 19176

RESEARCH PROTOCOL PROPOSAL

Title of Project: A Pilot Study for Respiratory Effort Sensing and Harvesting

Principal Investigator: Olga Boric-Lubecke; PhD
Electrical Engineering Department

A.1 Abstract

Unobtrusive sensing of respiratory effort and heartbeat can be valuable for continuous medical monitoring. Our research group is developing low-cost zero-net energy wearable biosensors for unobtrusive, continuous, respiratory effort and heart beat sensing and harvesting. The goal is to produce self-powered unobtrusive sensors suitable for continues (24/7) health monitoring. The advantages of this technique is that first; the patient is not wired to monitoring equipments, and second; there is no battery to be replaced or disposed. A large sample of human subjects is needed to better understand how variations in the population affect the performance of our biosensors. The objectives of our project and the proposed experiment methods on human subjects are elaborated below.

A.2 Specific Aims

The objectives of this project are to develop and implement zero-net energy wearable biosensors that can sense and harvest energy from respiratory effort. The goal is to produce self-powered unobtrusive sensors suitable for continuous (24/7), unobtrusive health monitoring. Our system consists of three components, 1) physiological sensor/harvester, 2) local module that will condition and store harvested potentials, and
extract physiological data, and communicate this information to the third part of the system which is 3) a remote module, via a short range, low data rate wireless link.

However, a good understanding of how variations in the population affect the performance of our systems is needed, thus we propose to conduct human testing for better understanding and improvement of our works. The experiment requires a human subject wearing a shirt with our embedded system. The system detects his/her respiratory rhythm and heart beat data and harvests the energy from his/her breathing and heart beat. Then the harvested energy will be conditioned and stored to be distributed to other parts of the system when needed and sends the physiological data to the wireless link. Wireless link then collects the data and sends it via a low data rate wireless link to a remote computer to be further processed and analyzed. Each experiment will be done in a laboratory and is expected to take less than ninety minutes.

### A.3 Background and Significance

Human energy harvesting for wearable and portable electronics was proposed in the mid 1990’s [31]. While a number of potential human energy sources were identified in [31], human energy harvesting has mostly been focused on kinetic energy [32, 33, 50], and more recently on thermal energy [51]. Primary self-powered electronic devices have included self-winding wrist watches and more recently laptops with hand cranks and foot pedals [31, 32]. The recent efforts to scavenge human kinetic energy using piezoelectric sensors and electromagnetic generators placed in shoes, and backpacks with spring-loaded straps have shown promise [32, 33]. Also it has been demonstrated that electromagnetic scavenging is more efficient than piezoelectric [33, 52].

Perhaps the most readily available form of human power is respiration, yet no significant work has been done on energy harvesting from movement of the chest walls due to respiratory effort. Similarly, human electrical signals have been identified as a potential energy source [31, 53], with no published efforts to date. On the other hand, wearable biosensors have been investigated for remote health and fitness monitoring, including applications ranging from wound healing to athletic training [54]. Wearable sensors have included ring, ear, and body sensors [54-57]. To power these systems,
simple batteries and proximity RF power scavenging [50] have been used. Our approach is to use self-powered biosensors, through sensing and harvesting methods for respiratory effort, and electro-cardiogram (ECG) potentials.

Movements of the chest due to quiet unforced respiration are mostly determined by movements of the rib cage and the abdominal wall. Studies have found that average chest wall displacement due to respiratory effort is on the order of cm [21]. Since respiratory effort itself is a valuable physiological parameter, the method that combines sensing and energy harvesting would enable efficient biosensing.

Our break-through approach is to concurrently harvest and sense physiological signals by reusing the hardware components to perform both functions, managed through a highly efficient control and communication protocol.

A.4 Preliminary Studies

Our recent work [34, 35, 39] to implement self-powered biosensors includes applying a method to detect and harvest respiration signals. However, we do not have a good understanding of how variations in the population affect the performance of our systems, thus we propose the experiments on a large number of subjects for better understanding and improvement of our works.

A.5 Research Design and Experimental Methods

The research experiments consist of measuring and categorizing data from our self-powered wearable biosensor system design. Each experiment will be done in the human performance laboratory located in the Kinesiology Department and is expected to be no longer than ninety minutes. The list of experimental instruments can be found in Table A.1.

A.5.1 Human Subjects Involvement

Participants for the pilot will be 100 healthy, physically active adults 19 to 34 years of age recruited from the University of Hawaii (UH) student population and the
Oahu community. Experimental protocol will include one on-site session. Participants will be instructed to report to the University of Hawaii Human Performance Laboratory in a well-hydrated state. Participants will complete health history and exercise questionnaires and sign the approved informed consent form. A healthcare professional (BOC Certified Athletic Trainer) will review each questionnaire and identify exclusion criteria. The subject's weight, height, and thorax dimensions will be measured. Thorax dimensions will be measured both at full inhale and full exhale – including waist circumference, chest circumference, chest breadth, and chest depth.

Inclusion criteria for all participants includes classification as low risk (ACSM Risk Stratification Categories) for exercise testing and free from any cardiovascular, coronary artery, pulmonary, or metabolic diseases (Mahler DA, Froelicher VF, Miller NH, York TD. *ACSM's Guidelines for Exercise Testing and Prescription.* 7th ed. Baltimore: Lippincott Williams and Wilkins; 2009.). Also being pregnant will exclude the participants from experiment.

Changes to the current protocol do not significantly increase participant risk and qualify as eligible for expedited review since the procedures being added would qualify for expedited review within any new IRB applications.

Participants will be informed of potential risks and that involvement is voluntary. Participants will also be informed they may refuse to participate at any time prior to or during the study without penalty.

**A.5.2 Testing Protocol**

All subjects will complete 4 stages of testing, each lasting 5 minutes: 2 resting stages and 2 exercise stages. The only breaks between stages will be those necessary to change positions, treadmill speed, and sensor adjustment.

Prior to test initiation, participants will be fitted with a heart rate monitor, 2 self-powered wearable respiratory effort sensors (one worn around the chest, and one worn around the stomach), a piezoelectric belt as a chest motion reference, headgear, and breathing mask to assess respiratory gas exchange throughout the test. Participants will breathe through the mask throughout the duration of the test which is connected to a
metabolic cart through ventilation tubes. The metabolic cart will be used to determine oxygen consumption (VO_2) and respiratory exchange ratio (RER).

The resting stages will consist of 5 minutes each in the seated and standing positions. The exercise stages will consist of two stages of exercise during which speed and treadmill grade will be held constant within each stage. Each stage will last 5 minutes. Stage 1 will be started and participants will walk at the pace immediately set at _3.5_ mph at 1% grade (approximately equivalent to a 17 minutes per mile walking pace which would be considered a “fast walk”). Stage Two will require participants to walk at _2.5_ mph at 1% grade (approximately equivalent to a 24 minutes per mile walking pace which would be considered a normal walking speed). A cool down will take place upon completion of the test during which the treadmill speed and grade will be immediately decreased allowing an easy walking pace for 2 min. The headgear and breathing mask will be removed at this time.

A.5.3 Outcome Measures

Oxygen consumption, HR, RER, O_2 sat, and RPE will be measured over the 5 minutes of each stage.

- Cardiovascular responses will be collected via standard open circuit spirometry. Inspired ventilation was measured with a previously calibrated dry gas meter (Rayfield RAM-9200) fitted with a potentiometer.
- Expired ventilation will be channeled through Hans Rudolf high velocity valve through low resistance plastic tubing into a 5-liter mixing chamber.
- The concentrations of oxygen and carbon dioxide will be continuously sampled with an Applied Electrochemistry Oxygen analyzer S-3A/1, Oxygen sensor N-22M, carbon dioxide analyzer CD-3A, and a carbon dioxide sensor P-61B which will be calibrated with commercially available primary standard grade gases.
- Heart rate will be measured via Model Q710 electrocardiogram (Quinton Instrument Co., Bothell, Washington).
- Respiratory rate and tidal volume will be measured by both spirometry and the self-powered respiratory sensor during all stages.
- Participants will use the 6-20 point Borg Scale (attached) at the end of each stage by pointing to the appropriate RPE value while continuing to exercise.

**Table A.1: Equipment List**

<table>
<thead>
<tr>
<th>Type of Equipment</th>
<th>Equipment Name</th>
<th>Amount</th>
</tr>
</thead>
<tbody>
<tr>
<td>Electromagnetic Generator Chest Belt</td>
<td>Low-power electromagnetic generator</td>
<td>2</td>
</tr>
<tr>
<td></td>
<td>Customized low-voltage rectifier</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>TI CC430 Wireless link Texas Instruments</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>6-20 point Borg Scale</td>
<td>1</td>
</tr>
<tr>
<td>Velocity Valve</td>
<td>Hans Rudolf high velocity valve</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Low resistance plastic tubing</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>5-liter mixing chamber</td>
<td>1</td>
</tr>
<tr>
<td>Electrochemistry Oxygen analyzer</td>
<td>Electrochemistry Oxygen analyzer S-3A/1</td>
<td>1</td>
</tr>
<tr>
<td>Oxygen sensor</td>
<td>Oxygen sensor N-22M</td>
<td>1</td>
</tr>
<tr>
<td>Carbon dioxide analyzer</td>
<td>Carbon dioxide analyzer CD-3A</td>
<td>1</td>
</tr>
<tr>
<td>Carbon dioxide sensor</td>
<td>Carbon dioxide sensor P-61B</td>
<td>1</td>
</tr>
<tr>
<td>Electrocardiogram</td>
<td>Electrocardiogram Quinton Instrument Co</td>
<td>1</td>
</tr>
<tr>
<td>Reference</td>
<td>Spirometer</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Piezoelectric chest belt</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Polar wear link chest belt</td>
<td>1</td>
</tr>
</tbody>
</table>
A.5.4 Risk and Hazard Evaluations

Small magnets pose very little hazard. Pacemakers and defibrillators can be sensitive to any magnetic fields. If you have a pacemaker or defibrillator, or have health issues which requires that you wear electronics of any sort, avoid using magnet generators. Although there are no known health risks involving the handling of permanent magnets of the sizes we use, just to be on the safe side we would advise pregnant women to avoid any exposure to magnetic fields. So if you are sexually active you should be in a birth control program in order to be in this study.

A.6 Data and Safety Monitoring Plan

1. All experiments will be logged.
2. All complaints from the participants will be logged and reported to the PI and CHS.
3. All unanticipated adverse events will be logged and reported to the PI and CHS.
Appendix B

CHS Protocol No. 19176 – Consent Form
UNIVERSITY OF HAWAI`I

INFORMED CONSENT AND PRIVACY AUTHORIZATION
TO TAKE PART IN A RESEARCH STUDY

Study Title: Respiratory Effort Sensing and Harvesting

Principal Investigator (PI):
Name: Dr. Olga Boric-Lubecke
Institutional Affiliation: Electrical Engineering Department, University of Hawaii at Manoa
Email: olgabl@hawaii.edu
Address: POST 205K, 1680 East-West Rd Honolulu, HI 96822
Phone Number: (808) 956-9648

Co-PI:
Name: Dr. Cris Stickley
Institutional Affiliation: Kinesiology Department, University of Hawaii at Manoa
Email: cstickle@hawaii.edu
Phone Number: (808) 956-3798

Sponsor:
Sponsor Name: REIS
Sponsor Address: 2540 Dole Street, Honolulu, Hawaii, 96822

Abbreviations Used:

PMG: Permanent Magnet Generator
EM: Electromagnetic
UH: University of Hawaii
RA: Research Assistant

Before you decide whether or not you would like to take part in this study, you should understand its purpose, how it may help, any risks, and what you will be asked to do. This process is called informed consent. If you agree to take part in the study, you will be asked to sign this consent form.

Before you learn about the study, it is important that you know the following:

- Taking part in this study is completely voluntary.
- If you decide to take part in the study, you can change your mind at any time and withdraw from the study.

What is the purpose of this study?

The objective of this project is to develop and verify self-powered unobtrusive wearable
biosensors, in the form of a shirt, that can sense and harvest energy from breathing and/or heartbeat.

**Why are you being asked to participate in this study?**

You are being asked to participate in this study because you are above 18 years old and do not have any significant medical problems. Researchers plan to enroll a total of 100 participants from Hawaii.

**How long will the study take and what procedures will be performed on you?**

Each experiment will be done in the Human Performance Laboratory of the Kinesiology Department at UH, and is expected to be no longer than ninety minutes. Some participants may be able to complete the procedures in one visit. Others may need two or three study visits to complete the process.

First, participants will undergo the consent process which provides participants time to read the consent form and ask questions. Then, if they agree to participate, participants will sign the consent form.

Next, participants will fill out a health history questionnaire and an exercise questionnaire. The questionnaire will be reviewed by a healthcare professional (BOC certified athletic trainer) to make sure you are healthy enough to participate.

Then the participant's weight, height, and thorax (the part of the human body between the neck and the diaphragm, partially encased by the ribs and containing the heart and lungs; the chest) dimensions will be measured. The thorax dimensions will be measured both at full inhale and full exhale including waist circumference, chest circumference, chest breadth, and chest depth.

Prior to test initiation, participants will be fitted with a heart rate monitor, two self-powered wearable respiratory effort sensors (one worn around the chest and one worn around the stomach), a piezoelectric belt (a piezoelectric sensor is a device measures pressure, acceleration, strain or force by converting them to an electrical charge), headgear, and breathing masks to assess respiratory gas exchange throughout the test. Participants will breathe through the mask throughout the duration of the test which is connected to a metabolic cart through ventilation tubes. The metabolic cart will be used to determine oxygen consumption and respiratory exchange ratio.

These sensors are suitable for continuous (24/7) health monitoring. There are four stages of testing, each lasting five minutes: two resting stages and two exercise stages.

The system detects the participant's respiratory rhythm and heart beat data and concurrently harvests the energy from his/her breathing and/or heart beat. Then the harvested energy will be distributed to other parts of the system when needed and the physiological data will be sent to the wireless link. The wireless link then sends the data
to a remote computer to be further processed and analyzed.

**What are the risks and discomforts that you may experience?**

This experimental system runs on small magnets which have been shown to pose very little risk. However, pacemakers and defibrillators can be sensitive to any magnetic fields. If you have a pacemaker or defibrillator, or have health issues which require that you wear electronics of any sort, you should not participate in this study.

Although there are no known health risks involving the handling of permanent magnets of the sizes we use, just to be on the safe side, we would advise pregnant women to not participate in this study. If you are sexually active, you should be in a birth control program if you want to participate in this study.

People who are uncomfortable in confined spaces, have a tendency to be claustrophobic, or who might have a stress reaction to wearing the head gear and mask should not participate in this study.

**How will your respiratory effort information be used?**

There may not be direct benefit to the participants. However, the results from this project will help better identify and address the issues with respiratory effort sensing and harvesting. Mainly the researchers will look into the differences of the attainable energy from different people of various ages, gender, and physical characteristics.

**How will my study data be kept confidential?**

Research data will be confidential to the extent allowed by law. Agencies with research oversight, such as the UH Human Studies Program, have the authority to review research data. All research records will be stored on a computer in a locked room in the primary investigator’s lab for the duration of the research project, and will be destroyed upon completion of the project.

The results of this research may be presented at meetings or in publications; however, you will not be identified.

**Will you be given the results of the study?**

The study PI or research assistant will not provide any individual study results to you or any member of your family, other doctors involved in your care, your insurance company, or your employer.

**How will this study benefit you?**

It is unlikely that you will benefit directly from participating in this study.
Are there costs or payments involved in this study?

There will be no costs or payments for your examination and tests in this study.

Can you revoke your consent for your participation in this study?

If you enter the study and you later change your mind, you can revoke (take away) your consent at any time, and there will be no penalty for you. This means that you can leave the study at any point as you are a voluntary research participant.

The study RA will decide if it is not possible or appropriate for you to continue to participate in this study due to unexpected health concerns or reactions to the experiment. (Add here what the circumstances are which would result in the RA taking a participant out of the study.)

Will you learn about new findings about risks of this study?

You will be told of any new information learned during the study that may change your willingness to continue in this study. At that time, you will be able to decide whether to continue your participation in this research study.

If you have any questions about this study, whom do you contact?

If you feel that you have been injured as a result of taking part in this study, or if you have any questions about the study, you should call the study PI, [Dr. Olga Boric-Lubecke], at [(808) 956-9648].

If you have questions about your rights as a research participant in this study, you should contact the University of Hawaii Human Studies Program at 808.956.5007 or by email at uhirb@hawaii.edu

Authorization to Use and Disclose (Release) Personal Health Information

The federal government has created a “Privacy Rule” under the Health Insurance Portability and Accountability Act (HIPAA). This Rule gives you the right to decide who can use and release your personal health information (also called “protected health information” or “PHI”) for the purposes of research. PHI is health information about study participants that could be linked to their identity. But we will not use your PHI in this study.

What happens if you do not sign this authorization?

Signing this authorization form is voluntary. If you do not sign this form, you will not take part in this research study.
Consent for You to Take Part in this Research Study

My signature indicates that I have read and understand this research consent/authorization form and that my questions have been satisfactorily answered. I understand that if, at any time, I have other questions, I can contact the study PI listed on page 1 of this form. I further understand that I will be given a copy of this signed consent/authorization form for my records.

_________________________________________
Name of the Participant

_________________________________________
Signature of the Participant    Date

***
Appendix C

CHS Protocol No. 19176 – Health History Questionnaire
Pre-Participation Medical History Form/ Physical Activity Readiness Questionnaire

Participant Information
ID number ____________________
Date of Birth:________________  Age (years) _________ Sex:   M   /   F
Home Address:___________________________________
City/State/Zip:______________________________
Email:______________________
Home/Cell Phone (___) _______________
Emergency Contact Person/Relationship/Phone Number:
____________________________________/_______/(___)_______________

Physical Activity Readiness Questionnaire (American College of Sports Medicine, 1997)
Please read the questions carefully and answer each one honestly.
YES   NO
☐   1. If you are female, are you at risk of pregnancy?
☐   2. If you are at risk of pregnancy are you using birth control pills?
☐   3. Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?
☐   4. Do you feel pain in your chest when you do physical activity?
☐   5. In the past month, have you had chest pain when you were not doing physical activity?
☐   6. Do you lose your balance because of dizziness or do you ever lose consciousness?
☐   7. Do you have a bone or joint problem (ex. back, knee or hip) that could be made worse by a change in your physical activity?
☐   8. Is your doctor currently prescribing drugs (ex. water pills) for your blood pressure or heart condition?
☐   9. Do you know of any other reason why you should not do physical activity?

Medical History: the subsequent sections were obtained following guidelines for exercise testing (American College of Sports Medicine, 2005).

A. History: please check the box any condition you currently have or had in the past.
☐ Heart Attack
☐ Heart Surgery
☐ Cardiac Catheterization
☐ Coronary Angioplasty (PTCA)
☐ Pacemaker/implantable cardiac
☐ Defibrillator/rhythm disturbance
☐ Heart valve disease
☐ Heart failure
☐ Heart transplantation
Congenital heart disease
Diabetes
Asthma
Lung Disease
Heart murmur
Seizures
Head injury or concussion
Loss of consciousness or memory

B. Symptoms: please check the box for any symptoms you have or had experienced at rest, during or following exercise.

Chest discomfort
Cough or wheezing
Dizziness, fainting, or blackouts
Difficulty breathing
Abnormal heart beats

Musculoskeletal Symptoms: please check the box for any symptoms you have or had experienced, locate and label the occurrence of each symptom on the figure below.

Numbness
Tingling
Pain
Swelling
Burning
Cramping
C. **Cardiovascular Health:** please check the box for any conditions applicable to you.

- [ ] Male over age 45 years
- [ ] Female over age 55 years
- [ ] Smoke or smoking cessation within the previous 6 months
- [ ] High blood pressure (greater than 140/90 mm Hg)
- [ ] Currently taking blood pressure medication
- [ ] High cholesterol (greater than 200 mg/dL)
- [ ] Family history of heart attack or heart surgery before age 55 (father or brother) or age 65 (mother or sister)
- [ ] Physically inactive (less than 30 minutes of physical activity at least 3 days per week).
- [ ] Overweight

D. **Additional Questions:** please identify any additional health issues by checking the corresponding boxes to answer “yes” or “no.”

<table>
<thead>
<tr>
<th>Yes</th>
<th>No</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Have you had a medical illness or injury since your check up or last physical?</td>
<td></td>
</tr>
<tr>
<td>2. Do you have an ongoing chronic illness?</td>
<td></td>
</tr>
<tr>
<td>3. Are you currently taking any prescription or nonprescription (over the counter) medications or pills or using an inhaler?</td>
<td></td>
</tr>
<tr>
<td>4. Has a physician ever denied or restricted your participation in sports or exercise for any heart problems?</td>
<td></td>
</tr>
</tbody>
</table>

Explain all “Yes” answers here and any checked boxes:

_______________________________________________________________________
_______________________________________________________________________

Signature of Participant: ___________________________ Date: _______________

Date_____________ Participant #___________
Appendix D
CHS Protocol No. 19176 – Approval Letter
MEMORANDUM

November 16, 2012

TO: Olga Boric-Lubecke, Ph.D.
Principal Investigator
Electrical Engineering

FROM: Denise A. Lin-DeShetler, MPH, MA
Director

SUBJECT: CHS #19176- "A Pilot Study for Respiratory Effort Sensing and Harvesting"

Your research project identified above, including the informed consent/privacy authorization form, was approved for one year by the University of Hawaii (UH) Human Studies Program at its IRB meeting on November 14, 2012.

This memorandum is your record of the Human Studies Program approval of this study. Please maintain it with your study records.

The Human Studies Program approval for this project will expire on November 13, 2013. If you expect your project to continue beyond this date, you must submit an application for renewal of this Human Studies Program approval. Human Studies Program approval must be maintained for the entire term of your project.

If, during the course of your project, you intend to make changes to this study, you must obtain approval from the Human Studies Program prior to implementing any changes. If an Unanticipated Problem occurs during the course of the study, you must notify the Human Studies Program within 24 hours of knowledge of the problem. A formal report must be submitted to the Human Studies Program within 10 days. The definition of "Unanticipated Problem" may be found at: http://hawaii.edu/irb/download/documents/SOPP_161_UP_Reporting.pdf, and the report form may be downloaded here: http://hawaii.edu/irb/download/forms/App_UP_Report.doc.

You are required to maintain complete records pertaining to the use of humans as participants in your research. This includes all information or materials conveyed to and received from participants as well as signed consent forms, data, analyses, and results. These records must be maintained for at least three years following project completion or termination, and they are subject to inspection and review by the Human Studies Program and other authorized agencies.

Please notify this office when your project is completed. Upon notification, we will close our files pertaining to your project. Reactivation of the Human Studies Program approval will require a new Human Studies Program application.

Please contact this office if you have any questions or require assistance. We appreciate your cooperation, and wish you success with your research.
References


