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Measuring Chest Circumference Change during Respiration with an Electromagnetic Biosensor

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Abstract—**In this paper, an off-the-shelf DC motor is modified into a chest belt and used to successfully measure circumference change on a mechanical chest model, while simultaneously harvesting significant power. Chest circumference change can provide information on tidal volume, which is vital in assessing lung function. The chest circumference change is calculated from the motor's voltage output. Calculated values are within 0.95mm of measured circumference changes, with a standard deviation of 0.37mm. The wearable motor can also harvest at least 29.4µW during normal breathing.**

I. Introduction

Tidal volume is a key index of the mechanical status of the ventilatory system. Tidal volume can also determine the effectiveness of a patient's gas exchange system [1], as well as forecast lung disease. The spirometer is a common tool for measuring tidal volume. However, spirometers can be uncomfortable for the patient, can change the patient's breathing pattern, and limit their mobility. Portable spirometers and other wireless biosensor systems require regular battery replacement for sensors and/or the radio frequency (RF) wireless link, thus making continuous, remote respiration monitoring infeasible.

Tidal volume has been demonstrated to be directly proportional to chest wall displacement [2] as well as chest circumference change [3], both of which can be measured non-invasively and inexpensively. Chest wall displacement has been measured via piezoelectric sensors [4] and Doppler radar [5]. Several methods have been developed to measure chest circumference dynamically. Pneumatic belts utilizing air [6] or mercury [7] 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 [8]. 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 [9]. While these methods can measure chest circumference change dynamically and accurately, the proposed electromagnetic biosensor has the added benefits of low cost, portability, simplicity, and energy harvesting capability.

Wearable technology, also known as smart clothing, can collect important physiological data without inconveniencing the patient or limiting the patient's mobility. Most of these systems provide physiological data such as heart rate and respiratory rate, using embedded sensors in conjunction with an integrated battery-operated transmitter that collects data and sends it over a wireless link to an accompanying battery-operated wristwatch or smart phone [10]. Wearable biosensors have been investigated for remote health and fitness monitoring, including applications ranging from wound healing to athletic training [11]. Wearable sensors have included ring, ear, and body sensors [11-14]. To power these systems, simple batteries and proximity radio frequency (RF) power scavenging [15] have been used.

However, continuous health monitoring with these smart clothing systems require constant changing of batteries, resulting in a large amount of disposed batteries that present an environmental hazard. In addition, many people tend to stop using such devices, or throw them away once the batteries run out.

Human energy harvesting may be used to power such smart clothing systems. Human energy harvesting for wearable and portable electronics was proposed in the mid 1990's [16]. However, human energy harvesting has mostly been focused on kinetic energy [15, 17, 18], more specifically, the locomotion during walking. It has been demonstrated that electromagnetic scavenging is more efficient than piezoelectric [13, 18].

In [19, 20], the concept of self-powered biosensors through sensing and harvesting of respiratory effort was introduced. In [21], a wearable off-the-shelf servo motor was used to extract respiratory rate accurately as well as harvest respiratory energy.

In this paper, a wearable, off-the-shelf DC motor is used to measure circumference change on a mechanical chest model, as well as harvest respiratory energy. This concept can be used for chest circumference measurements, which can lead to tidal volume measurements. The energy harvested can be used to power a low-power system-on-chip (SoC), which can process the motor output, extract respiratory data, and transmit it wirelessly to a receiver display, resulting in a zero-net energy biosensor.

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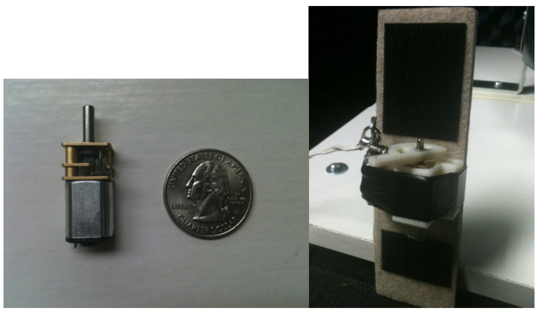
II. WEARABLE DC MOTOR DESIGN

Electromagnetic generation is based on Faraday's Law, shown in (1), which states that a time-varying magnetic field induces an electric field across a conductor, resulting in an electric current.

$$\nabla \times E = -\frac{dB}{dt} \tag{1}$$

A Pololu 50:1 DC brushed micro metal gearmotor was chosen based on size, cost, and easy modification into a chest belt. The motor, shown in Fig. 1a, is modified into a wearable chest belt. The modified device is shown in Fig. 1b.



(a) (b)

Fig. 1. Pololu 50:1 micro metal gearmotor #1098 (a) and modified wearable DC motor (b).

The motor is fit into a plastic housing 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 hard felt 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 snuggly 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. 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.

III. CALCULATION OF CHEST CIRCUMFERENCE CHANGE

Chest circumference change can be calculated from Faraday's Law. According to Faraday's Law for a coil of wire, shown in (2), the induced voltage is proportional to the number of coils, N, and the rate of change of the magnetic flux.

$$V = -N\frac{d\phi_B}{dt} \tag{2}$$

For a rotational electromagnetic generator, the magnetic flux is changed when the armature is turned. Thus, the rate of change of magnetic flux is proportional to the angular frequency, ω, of the armature.

$$\frac{d\phi_B}{dt} = K_1 \cdot \omega \tag{3}$$

Angular frequency is equal to the derivative of the angular displacement, θ.

$$\omega = \frac{d\theta}{dt} \tag{4}$$

$$V = -N \cdot K_1 \cdot \frac{d\theta}{dt} \tag{5}$$

$$\int -\frac{1}{N \cdot K_1} \cdot V \cdot dt = \int d\theta \tag{6}$$

$$-\frac{1}{N \cdot K_1}\left(\int V \cdot dt + C\right) = \theta \tag{7}$$

The arc length, s, can be found by multiplying the angular displacement by the armature radius, r. The arc length of the armature is proportional to the chest circumference change, ΔCC.

$$s = r\theta \tag{8}$$

$$\Delta CC = K_2 \cdot s \tag{9}$$

Thus, the change in chest circumference can be found using the integral of the motor's voltage output over time.

$$\Delta CC = \left(-\frac{K_2 \cdot r}{N \cdot K_1}\right)\left(\int V \cdot dt + C\right) \tag{10}$$

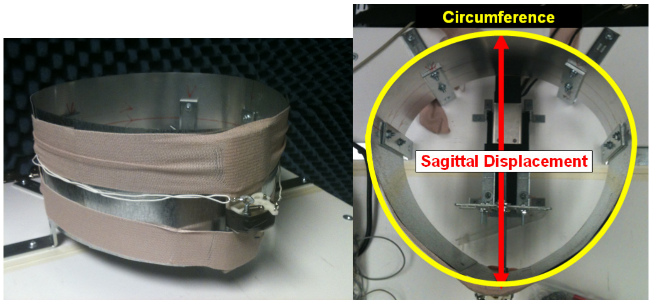
$$Let -\frac{K_2 \cdot r}{N \cdot K_1} = K$$

$$\Delta CC = K \cdot \left(\int V \cdot dt + C\right) \tag{11}$$

The coefficient K and constant C can be found from the voltage outputs of known, measured chest circumference changes.

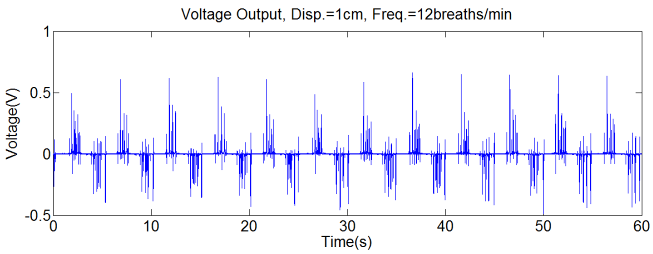
IV. EXPERIMENTAL RESULTS

A mechanical chest model was built to simulate respiration and test the electromagnetic biosensor. Side and top views of the chest model are shown in Fig. 2. The mechanical chest is built around a Griffin Motion programmable linear stage, which oscillates the front of the chest to simulate respiration at the desired sagittal displacement and frequency. Two samples of the voltage output are shown in Fig. 3.

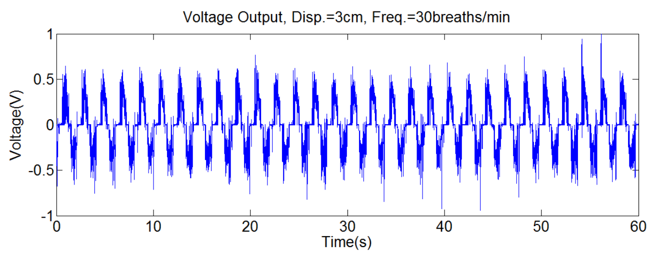


(a) (b)

Fig. 2. Mechanical chest model with EM biosensor (a), and top view showing sagittal displacement and circumference (b).



(a)



(b)

Fig. 3. Voltage output at 1cm displacement, 12 breaths/min (a), and at 3 cm displacement, 30 breaths/min.

Circumference measurements were made with tape measure. According to (11), the integral of the voltage, with respect to time, should be consistent regardless of the frequency. Each inspiration, denoted by the positive voltage peaks, is integrated, and the average value over one minute is calculated and recorded in Table I. The displacements, circumference measurements, frequencies, and integrations are also shown in Table I. At different frequencies, the voltage integrals for different circumference changes are fairly constant. Although the voltage integrals can range up to about 7V·s, the resulting error in calculating circumference change should be minimal.

The circumference is measured for displacement ranging from 0.4cm to 3cm, at a frequency of 12 breaths/min. 0.4cm is the minimum displacement that the EM biosensor can sense at 12 breaths/min. 3cm is an estimate of the maximum sagittal displacement during breathing.

At 0.4 cm displacement, the measured circumference change is 0.55cm, and the average integral of the voltage during inspiration is 7.7V·s. At 3cm displacement, the measured circumference change is 3.6cm, and the average integral of the voltage during inspiration is 240V·s. K and C from (11) are calculated from the voltage outputs at 0.4cm and 3cm displacement.

$$K = 0.013122 \frac{cm}{V \cdot s} \tag{12}$$

$$C = 34.2494 V \cdot s \tag{13}$$

Voltage outputs were recorded at displacements ranging from 0.4cm to 3cm. The circumference changes are calculated and compared with the measured circumferences and recorded in Table II.

TABLE I. CIRCUMFERENCE CHANGES AND FREQUENCIES OF CHEST MODEL, WITH VOLTAGE INTEGRATIONS

Disp. (cm)	Measured Circumf. Change (cm)	Freq. (breaths/min)	∫V·dt (V·s)
1	1.25	6	67.76
1	1.25	12	66.96
1	1.25	18	68.35
1	1.25	30	66.22
2	2.4	6	155.6
2	2.4	12	155.7
2	2.4	18	150.5
2	2.4	30	152.8
3	3.6	6	239.5
3	3.6	12	240.1
3	3.6	18	245.97
3	3.6	30	238.7

TABLE II. CALCULATED CIRCUMFERENCE CHANGES VS. MEASURED CIRCUMFERENCE CHANGES

Disp. (cm)	∫V·dt (V·s)	Measured Circumf. Change (cm)	Calculated Circumf. Change (cm)	Absolute Error (cm)
0.5	14.910	0.65	0.645	0.005
0.6	16.074	0.75	0.660	0.09
0.7	33.047	0.85	0.883	0.033
0.8	42.670	1.0	1.009	0.009
0.9	52.013	1.15	1.132	0.018
1.0	58.970	1.25	1.223	0.027
1.5	112.62	1.85	1.927	0.077
2.0	154.84	2.4	2.481	0.081
2.5	194.87	3.1	3.006	0.094

All absolute errors are less than 1mm, which shows high agreement between the measured and calculated chest circumference changes. The average absolute error between measured and calculated circumference changes is 0.048cm, with a standard deviation of 0.0037cm.

V. OUTPUT POWER

At normal breathing, which is approximately 1cm sagittal displacement and 12 breaths/min, the EM biosensor can harvest an average of 29.4µW. The power output, from the data shown in Fig. 4a, is shown below in Fig. 4. The load resistance on the motor is 160 ohms. This is the minimum resistance at which the torque needed to turn the armature is the lowest. The lower the torque, the easier it is for respiration to turn the armature, but the power collected drops as well.

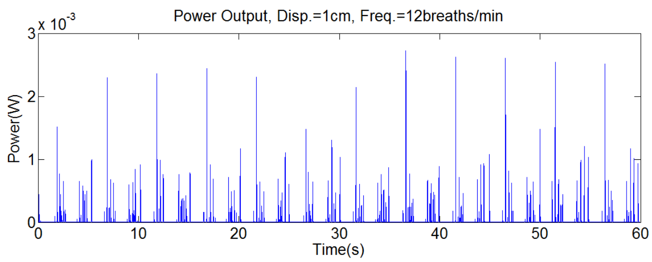


Fig. 4. Voltage output at 1cm displacement, 12 breaths/min.

A low-power SoC, such as the Texas Instruments CC430, can consume just tens to hundreds of microwatts [21]. By using more than one chest belt on the torso, or by minimizing the time of continuous SoC processes, enough power could be generated to create a zero-net energy biosensor.

VI. CONCLUSION

The electromagnetic biosensor can be used to measure chest circumference as well as harvest respiratory energy simultaneously. The circumference changes are measured on a mechanical chest model, simulating different sagittal displacements and respiration rates. The change in chest circumference is found to be linearly related to the integral of the voltage output of the motor. The EM biosensor was able to calculate chest circumference change with a mean absolute error of 0.048cm with a standard deviation of 0.0037 cm.

Further experimentation will be done on human subjects, using a respiratory strain gauge as a reference for chest circumference change, and a spirometer for a tidal volume reference. This can lead to established relationships between the biosensor voltage output, chest circumference change, and tidal volume. Because a chest belt is not ideal for comfort, improvement will be made on the wearable servo motor design, incorporating comfort and size considerations, and integration into clothing.

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