34th Annual International Conference of the IEEE EMBS
San Diego, California USA, 28 August - 1 September, 2012

Respiratory Effort Energy Estimation Using Doppler Radar*

Ehsaneh Shahhaidar, *Student Member, IEEE*, Ehsan Yavari, *Student Member, IEEE,* Jared Young,
Olga Boric-Lubecke, *Senior Member, IEEE,* Cris Stickley

***Abstract*— Human respiratory effort can be harvested to power wearable biosensors and mobile electronic devices. The very first step toward designing a harvester is to estimate available energy and power. This paper describes an estimation of the available power and energy due to the movements of the torso during breathing, using Doppler radar by detecting breathing rate, torso displacement, torso movement velocity and acceleration along the sagittal movement of the torso. The accuracy of the detected variables is verified by two reference methods. The experimental result obtained from a healthy female human subject shows that the available power from circumferential movement can be higher than the power from the sagittal movement.**

I. Introduction

Human body is an incredible storehouse of energy. If a small fraction of this stored energy could be scavenged, a mobile device may potentially have a renewable resource to draw upon, reducing the dependency on the battery power and cost of maintenance. It would be convenient if this energy can be harvested by the user's everyday actions [1]. However, the very first step toward designing an energy harvesting system is the estimation of the available power and energy. The next step then, will be designing a system with the highest efficiency based on the estimated available power.

Perhaps one of the most energy abundant and readily available forms of human power is respiration. This form of energy depends on various personal factors such as breathing rate, depth, body mass index, etc. Energy from respiratory effort can be very small empirically [2-3] although few studies [1] have analytically analyzed available power and published data which seems overly optimistic. In [1] the force exerted from chest during normal breathing on a chest belt was estimated to be 100N which is higher than the force used when physiotherapists apply vibration to a human subject [4], and on the same order of magnitude as the force used during CPR.

Pulmonary ventilation, commonly referred to as breathing, involves movements of air into and out of the lungs increasing the volume inside the lungs during inspiration and decreasing the volume during expiration.

The force and work exerted on the thoracic area is initially from the change of air pressure and volume during pulmonary ventilation, resulting in the movement of the torso which can be detected conveniently.

The movements originating from respiratory effort can be detected via different methods. The most common one is using a piezoelectric chest belt, however this method typically does not provide absolute displacement [5]. Accelerometers have shown to be a promising way of breathing rate detection, as well as detection of some other variables like breathing tilt and acceleration [6-9]. Short range Doppler radar also can be used to measure any moving target like torso's displacement in a non-contact way [10-11].

In this paper, first, applied force and energy resulting from air pressure and volume change in the thorax will be calculated for a healthy subject. Then sagittal (frontal radial) displacement of the whole thoracic area is detected via a Doppler radar. Thereby, maximum circumferential displacement (in transverse plane, (Fig. 1)) is calculated by modeling the torso, including thorax and abdomen, as a simple half-cylinder with varying radius length.

Two methods will be used for determining acceleration and displacement: body mounted accelerometers, and Doppler radar based physiological movement sensing. Then the dynamic force, available power and energy due to sagittal and circumferential movement (along the transverse plane)

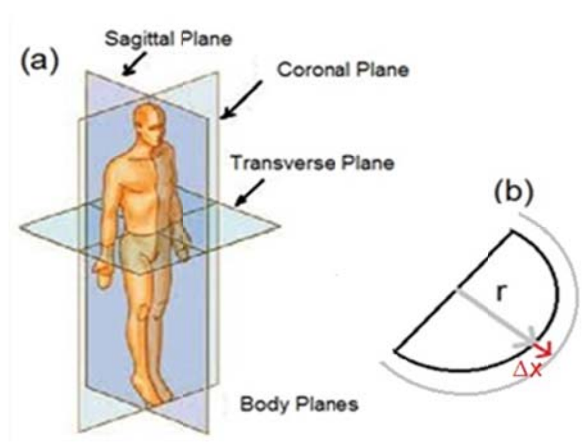

Figure1. a) Anatomical planes, b) Cross section of human torso as a half cylinder with varying radius along sagittal plane.

* This work was supported in part by the National Science Foundation under grant ECCS-0926076, the University of Hawaii Renewable Energy and Island Sustainability (REIS) Program, and by Department of Energy grant, DE-OE0000394.

Ehsaneh Shahhaidar is a PhD student at the Electrical Engineering Department, University of Hawaii at Manoa, Honolulu 96822 USA (e-mail: ehsaneh@hawaii.edu).

Ehsan Yavari is a PhD student at the Electrical Engineering Department, University of Hawaii at Manoa, Honolulu 96822 USA (e-mail: ehsan@hawaii.edu).

Jared Young is a BS student at the Electrical Engineering Department, University of Hawaii at Manoa, Honolulu 96822 USA (email:Jared9@hawaii.edu).

Olga Boric-Lubecke is an Associate Professor at Electrical Engineering Department, University of Hawaii at Manoa, Honolulu 96822 USA (e-mail: olgabl@hawaii.edu).

Cris Stickley is an Assistant Professor at the Kinesiology and Rehabilitation Science Department, University of Hawaii at Manoa, Honolulu 96822 USA (email: cstickle@hawaii.edu).

978-1-4577-1787-1/12/$26.00 ©2012 IEEE

during normal respiratory effort are calculated. The extracted respiratory rate, as one of the variables, is verified by a piezoelectric chest belt and a 3-axis accelerometer.

II. Mathematical Model

A dynamics model of whole torso area is composed of the lung volume and the chest wall represented as a cylindrical piston [12]. The pressure of the gas inside the piston/lungs (p) is in fact the force exerted by the gas molecules (F_g) on the surface of the enclosure of thoracic area (A):

$$p = \frac{F_g}{A}, [Pa = \frac{N}{m^2}] \quad (1)$$

The pressure changes required for adequate ventilation at rest are really quite small. At standard atmospheric pressure (760mmHg), inspiration may decrease the pressure in the lungs by only about 2 to 3mmHg [13]. According to [14] the effective area of the thorax is mentioned to be about $0.082m^2$. Applying numbers above, this pressure will exert 32.8N force on the thoracic area. At the same time, tidal volume gives the volume change during normal respiration. Knowing pressure and tidal volume, work/mechanical energy during normal respiration resulting from F_g is calculated from:

$$W_g = p \,.\Delta V_g \quad (2)$$

where ΔV_g is the gas/air volume change which is the same as tidal volume here (normally between 450ml for female and 500ml for male [15]). The amount of W_g (the work/ mechanical energy resulted from this pressure and volume change in pulmonary ventilation process) varies between 0.119J to 0.179J for a healthy female subject and between 0.133J to 0.2J for a healthy male subject.

A complete mathematical model of the thorax is a two-chamber spring-mass system (two lungs) and in simplified model is a single chamber one-lung model [12] (Fig.2) assuming that the pressure from the air ,p, is acting simultaneously on both lungs. The model is a single degree of freedom system in which chest wall response (displacement, velocity and acceleration) mathematical model will be represented by:

$$M.\frac{d^2x}{dt^2} + C.\frac{dx}{dt} + Kx = A.[p(t) + P_0 - (\frac{V}{V - Ax})^{\gamma}.P_0] \quad (3)$$

where A is the effective area, M is the effective mass of the torso (thorax and abdominal area) which is moving during respiration., V is the initial gaseous volume of the lungs, x is the displacement, C is the damping coefficient and K is the spring constant representing the viscoelastic damping properties of the tissues and elastic effect of the rib cage respectively. P0 is the ambient pressure, p(t) is the intra-thoracic (lung) pressure change over the time; and γ is the polytropic exponent for gas in lungs.

The displacement is detectable by use of Doppler Effect. A CW Doppler radar transceiver transmits continuous wave and receives phase-demodulated signal reflected from a target. In case of the periodically moving chest, the demodulated phase is proportional to the time-varying chest displacement related

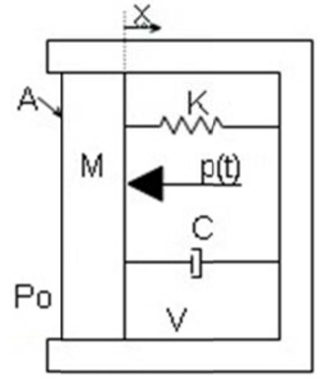

Figure 2: Single-Chamber One-Lung Model

to the breathing.

Quadrature receiver architecture is used to overcome radar sensitivity to target position [10-11]. Assuming that target's motion variation is given by Δx (t), the quadrature baseband output assuming balanced channels can be expressed as [10]:

$$B(t) = A_r.\exp[\theta + 4\pi\Delta x(t)/\lambda] \quad (5)$$

where θ is the constant phase shift related to the surface of a target and the phase delay between the mixer and antenna and λ is the wavelength. The ratio of the quadrature outputs, phase information linearly proportional to target's torso motion during breathing is extracted after arctangent demodulation.

Dynamic force related to every moving part along Δx is calculated by Newton's second law which relates the net force F and the resulting translational motion as:

$$F_s(t) = M.\frac{d^2x_s(t)}{dt^2} \quad (6)$$

During our measurements subject is seated facing the radar, therefore, displacement detected by radar would be the sagittal displacement ($x_s(t)$, indicated as Δx in Fig.1) and the force resulting from moving thorax mass along the sagittal displacement would be the sagittal force. The torso effective mass, M which is the same parameter in (3), can be estimated from a known body weight. Based on segmental properties [16], the torso moving mass, including thorax and abdomen, can be estimated as 27.4 % of the person's body weight.

For a simple half-cylinder-shaped torso with varying radius length, work/kinetic energy in radial (sagittal) direction is defined as:

$$W_s(t) = \frac{1}{2}M.(\frac{dx(t)}{dt})^2 \quad (7)$$

Derivative of the instantaneous energy with respect to time gives the instantaneous power:

$$P_s(t) = \frac{dW_s(t)}{dt} \quad (8)$$

(6) to (8) are indicating the amount of force, energy, and power available during normal breathing without using any harvesting device. The amount of harvested power will depend on the efficiency of the particular harvesting method and reversely to the degree of comfort of the harvester.

The harvesting device could be embedded in a chest/abdomen belt wrapped around the chest [3, 17]. In this case, the applied force from body to the belt depends on the elasticity (Young's modulus or in a simple format, spring constant) of the linkage (belt) transferring the sagittal displacement to a circumferential one. The force exerted by torso on a stretched chest/abdominal belt during inspiration would be calculated as:

$$F_{circum} = K_{belt}.\Delta x_{circum} \tag{9}$$

For a half-cylinder-shaped torso, the maximum circumferential displacement of the thoracic area (Δx_{circum}) will be π times more than the sagittal displacement.

Without using any belt, the spring constant and the amount of this force depends on the Young's modulus of the human body skin, cross sectional area of the skin using as a belt or string and the chest or abdomen circumference.

The elastic potential energy stored is the integral of the above equation with respect to Δx_{circum}:

$$\begin{aligned} U_e &= U_{circum} = \int F_{circum}.dx_{circum} \\ U_{circum} &= \frac{1}{2} K_{belt}.(\Delta x_{circum})^2 \end{aligned} \tag{10}$$

The harvestable circumferential power will be calculated from (8) by using U_{circum} instead of W_s.

III. Experimental Model

The radar used for these measurements has a single antenna mono-static configuration of received RF signal. An HP83640B signal generator operating at 2.45GHz was used as the signal source. ZFSC-2-2500 couplers from mini-circuits were also included in the system to split the signal source output into the transmitter antenna and local oscillator paths with 90 degrees phase difference. The transmitted CW microwave at the antenna input was measured to be 10dBm. The Antenna Specialist (ASPPT2988) antenna was used with 8dBi gain and 60 degree E-plane beam width for transmitting and receiving. A Minicircuit FM4212 mixer was used to mix down the received signal to the baseband. The subject facing the antenna is seated at a distance of 1m.

The accuracy of the radar was tested by a linear moving target before performing the human testing measurements; the channel imbalance factors were measured and applied to get the most accurate result.

A chest band (Pneumotrance 1132 piezoelectric respiration transducer) and an analog 3-axis accelerometer (ADXL327) were used on the subject as references for breathing rate and the derived acceleration. The reference systems were adjusted to be on the solar plexus point right below sternum which gives the best signal and the most reliable output. The movement of the solar plexus is an indication of both abdomen and diaphragm movements [7]. Four low noise amplifiers, SR560, were used for amplification of the output signals, and the recorded I and Q outputs of the Doppler, piezoelectric chest belt and the accelerometer were all digitized by a 12-bit National Instruments data acquisition device, to the PC with the sampling rate of 1000Hz by Labview.

IV. Measurement Results

The respiration of the seated subject facing radar was recorded for 60 seconds via the DC coupled radar, piezoelectric chest belt, and the 3-axis digital accelerometer.

Breathing rate was detected and verified by all three methods by FFT in Matlab (Fig. 3). Also, the accuracy of the chest acceleration detected by radar was verified by a body mounted accelerometer. The z-axis of the accelerometer was selected for analysis representing the sagittal displacement of the torso, used as the ground truth reference for Doppler measurements. Detected acceleration in this axis matches with radar results after accelerometer calibration and converting the units to cm/s^2. After simulations in Matlab, the detected variables by Doppler radar for the whole moving area of the torso found to be as follows: the displacement varies between -1.08 to 0.75 cm, the velocity changes between -1.35 to 1.31 cm/s and acceleration is between -2.06 to 2.12 cm/s^2. This is while the acceleration detected by ADXL327 varies between –2.18 and 3 cm/s^2 (Fig. 4). The agreement is satisfactory considering the fact that the radar is detecting the movement of the whole torso not just one part of the body. Since human breathing rate lies between 0.1 to less than 0.5Hz (10 to 30 breaths per minute), the comparison completed in Matlab after DC cancellation of all signals and processed by passing through a FIR band pass filter (0.1-0.5Hz).

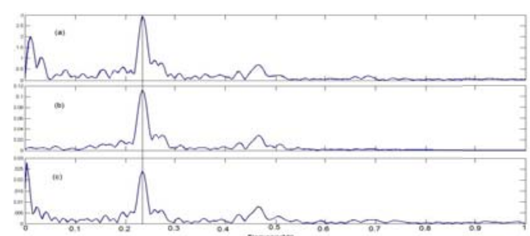

Figure 3. Breathing rate detection via a) Doppler, b) piezoelectric chest belt, and c) accelerometer (0.23Hz =14 breaths per minute).

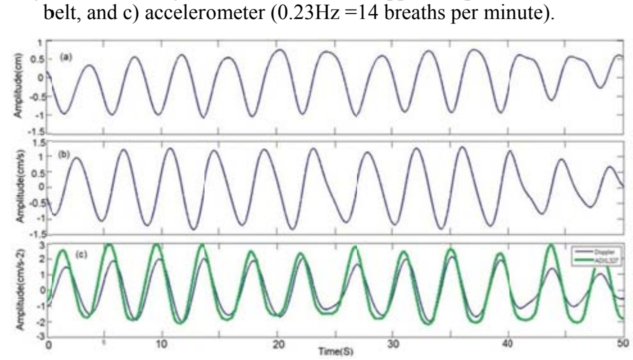

Figure 4. Sagittal a) displacement, b) velocity, c) acceleration detected by Doppler radar (blue) and detected by accelerometer (green)

Afterwards, the instantaneous dynamic force, available power and harvestable energy in a chest belt for a healthy female 60kg subject was determined (Fig. 5) based on Doppler data for the whole torso moving parts both in sagittal and circumferential movement. Simulations in Matlab showed that during sagittal movement of the torso the maximum dynamic force is 0.45N, maximum absolute amount of available power is 3.23mW, and maximum available energy is 1.3mJ. The counterpart amount of force for circumferential movement around the chest for a female subject with 2.65mm skin thickness [18] (assuming A_{skin}= 2.65*2.65mm^2, like a string) on chest area and18.8 MPa Young's Modulus [19], will be maximum 3.53N. Also the absolute amount of available potential energy and the available power in the same direction will be maximum 36.09mJ and 44.74mW respectively. By an efficient harvesting method, energy could be harvested in both directions [3, 17]. Phase difference between the waveforms is because of the different elements that have been used, e.g. F_s and F_{circum} have 180 degrees phase change resulting from acceleration and displacement phase difference in (6) and (9), also making sense considering the delay of transmission of the sagittal force from one linkage to another linkage including circumferential force.

The results show the available force, energy and power in circumferential movement is higher than the sagittal counterparts and also it is more confortable to harvest.

V. Conclusion

A new method for estimation of available power and energy from respiratory effort was investigated by Doppler radar, while the accuracy of breathing rate and acceleration detection was verified by two other reference methods, piezoelectric chest belt and a body-mounted 3-axis accelerometer. The measurements were performed on a healthy female subject. The results demonstrate that the harvestable power in a chest belt with circumferential displacement is higher than for sagittal movement.

Acknowledgment

The authors would like to thank Noah Hafner, Adytia Sing, and Xiaomeng Gao for their assistance during measurements.

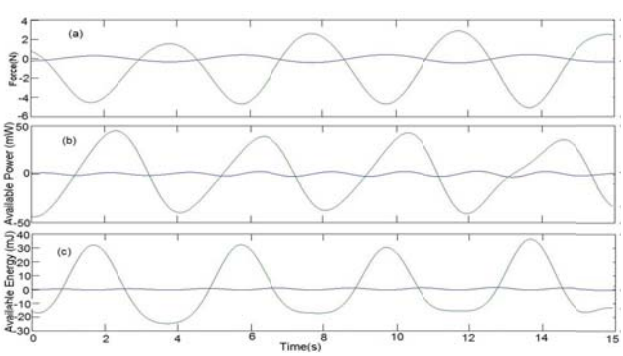

Figure 5. Comparison between sagittal (blue) and circumferential (green) a) dynamic force, b)available power and c) energy

References

[1] T. Starner, and J. Paradiso, "Human generated power for mobile electronics", Low-Power Electronics Design, C. Piguet, ed., Chapter 45, pp.1-35, CRC Press, Boca Raton, FL, 2004.

[2] E. Hausler, and E. Stein, "Implantable Physiological Power Supply With PVDF Film", Ferroelectronics, 1984, Vol. 60, pp. 277-282.

[3] E. Shahhaidar, O. Boric-Lubecke, M. Wolfe, and R. Ghorbani, "Electromagnetic generators as respiratory effort energy harvester", IEEE Power and Energy Conference (PECI), Urbana-Champaign, Illinois, Feb.2011.

[4] B. McCarren, JA. Alison, RD. Herbert ,"Vibration and its effect on the respiratory system", Australian Journal of Physiotherapy. 2006; 52(1):39-43.

[5] O. Boric-Lubecke, V. M. Lubecke, I. Mostafanezhad, and E. Shahhaidar,"Zero-net energy wearable biosensors", ISA 25th International Conf. on Computers and Their Applications (CATA2010), Honolulu, HI, Apr2010.

[6] T. Reinvu, M. Hannula, and H. Sorvoja, "Measurement of respiratory rate with high resolution accelerometer and EMFit pressure sensor", Proc. IEEE Sensor Appl. Sym., pp192-195, Jan2006.

[7] H. Alamäki, R. Sliz, M.I. Ashraf, H. Sorvoja, E. Alasaarela, M. Hämäläinen, "Measurement of ECG, respiratory rate, tilt and temperature of a patient and wireless zigbee data transmission", 2nd Intern. Symp. on Medical Information and Communication Technology, Oulu, Finland, Dec. 2007.

[8] P.D. Hung, S. Bonnet, R. Guillemaud, E. Castelli, P. T. N. Yen, "Estimation of respiratory waveform using an accelerometer" , 5th IEEE International Symposium on Biomedical Imaging From Nano to Macro (2008)

[9] A. Bates, M.J. Ling, J. Mann, D.K. Arvind, "Respiratory rate and flow waveform estimation from tri-axial accelerometer data", Int. Conf. on Body Sensor Networks (BSN, Singapore, June 2010), pp144–150.

[10] B.-K. Park , O. Boric-Lubecke, and V. M. Lubecke, "Arctangent demodulation with dc offset compensation in quadrature Doppler radar receiver systems," IEEE Transactions on Microwave Theory Technique, Vol. 55, 1073-1079, May 2007.

[11] A. Droitcour, O. Boric-Lubecke, V. M. Lubecke, J. Lin, and G. T. A. Kovac, "Range correlation and I/Q performance benefits in single-chip silicon doppler radars for non-contact cardiopulmonary monitoring ," IEEE Transactions on Microwave Theory Technique, Vol. 52, 838-848, March 2004.

[12] A. D. Greer, "Numerical modeling for the prediction of primary blast injury to the lung", University of Waterloo, thesis for Master of Applied Science in Mechanical Engineering, Waterloo, Ontario, Canada, 2006.

[13] H. Wilmore, D. L. Costill, W.L. Kenny, "Physiology of sport and exercise", Fourth Edition, Human Kinetics, 2008.

[14] "Test methodology for protection of vehicle Occupants against Anti-Vehicular Landmine Effects", RTO technical Report, Apr.2007, Chap. 3, pp: 25-26.

[15] A. R. Frisancho, "Human adaptation and accommodation", Ann Arbor: University of Michigan Press, 1993, p180.

[16] J. Enderle, J. D. Bronzino, S. M. Blanchard, "Introduction to biomedical engineering", Elsevier Academic Press (2005), pp-149.

[17] B. Padasdao, and O. Boric-Lubecke, "Respiratory rate sensing using servo motors", IEEE EMBC2011, Boston, MA, Aug.2011.

[18] T. Reilly, "Musculoskeletal disorders in health-related Occupations" IOS Press, 2002, pp.144-145.

[19] F. H. Silver, J.W. Freeman, D. DeVore, "Viscoelastic properties of human skin and processed dermis paper", Skin Res. Technol. Feb. 2001, pp.18-23.

[20]