Proceedings of the 28th IEEE
EMBS Annual International Conference
New York City, USA, Aug 30-Sept 3, 2006

EMG-Based Detection of Inspiration in the Rat Diaphragm Muscle

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Abstract—**An algorithm to detect the timing of each breath from an electromyogram (EMG) signal was developed. The algorithm has low computation cost and would be suitable for applications of implantable diaphragm pacing devices or as a trigger for each breath generated by a mechanical ventilator. The algorithm was implemented both in a LabView program on a desktop computer and in a C program on a microcontroller chip, and was tested on the EMG signal from the left diaphragm muscle of an anesthetized rat via implanted electrodes. The breath detection by the algorithm was over 99% accurate when the anesthetized rat was lying still, but for periods when the rat was gently wiggled to introduce noise and irregular breathing patterns, 19% of the breaths were missed and false positives occurred 6% of the time.**

I. Introduction

Loss of contractile activity of the diaphragm muscle leads to insufficient ventilation and may lead to the need for mechanical ventilation for sustaining life. Causes for the loss of effective diaphragm activity include loss of function of the innervating phrenic nerve resulting from trauma or surgery [1], or from lack of central nervous system (CNS) excitation of the phrenic nerve resulting from motoneuron loss (e.g., in multiple sclerosis or anterior horn disease), spinal cord injury [2], congenital central hypoventilation syndrome [3, 4] or stroke. Prolonged periods of mechanically-driven ventilation lead to atrophy and weakness of the diaphragm muscle, increased risk of infection, difficulty in relearning to effectively breathe without the ventilator and inspiratory muscle dysfunction [5, 6].

Since currently available mechanical ventilators pace each breath at a constant rate, not based on any input from physiological signals, voluntary activities such as speaking or swallowing are interfered with [7]. Physiological signals that may serve to encode the timing of each breath include compound action potentials (CAP) within the phrenic nerve, hypoglossal nerve [8], posterior cricoarytenoid muscle that opens the glottis [7], and diaphragm muscle. Triggering the timing of the generated breaths based on these physiological signals may attenuate some of the complications [9, 10] and enhance the quality of life [7], for periods when respiratory assistance is required.

Manuscript received April 24, 2006. This work was supported by NIH Grant HL37680.

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We have developed an algorithm to analyze real-time diaphragm CAP activity that correlates with breathing and to detect the timing of inspiration activity for each breath. This detection of timing could be used to trigger engineered devices to generate the inspiratory action of a breath by mechanical ventilation [9, 10] or by electrical stimulation of the phrenic nerve [2, 4, 11]. To test this algorithm, we have recorded the EMG signal of an innervated and functioning diaphragm muscle by an implanted pair of electrodes in a rat. The algorithm was implemented in both a LabView program on a desktop computer and a C program on a microcontroller chip within a circuit that was designed for chronic implantation studies in animals.

II. Methods and Materials

A. EMG Recording Electrodes

For development and testing of the algorithm to detect respiratory inspiration, the EMG signal from an innervated and functioning diaphragm muscle was monitored using a pair of stainless steel electrode wires (AS631, Cooner Wire, Chatsworth, CA, USA) implanted chronically into adult, male, Sprague-Dawley rats. The animal experimental protocol was approved by the Institutional Animal Care and Utilization Committee and was in accordance with the guidelines established in the United States Public Health Service Guide for the Care of Laboratory Animals (NIH Publication 85-23). For surgical electrode implantation and for later EMG recordings, the rat was anesthetized with ketamine (80 mg/kg) and xylazine (20 mg/kg) by intramuscular injection. The electrodes of the pair were implanted parallel to one another, separated by a distance of ~4 mm. Each electrode wire had a 5 mm segment of insulation stripped and inserted into the left diaphragm muscle. The insulated portion of the wire extending from each electrode was brought from the diaphragm through an opening in the abdominal region and tunneled subcutaneously to a region in the back near the shoulder blades, such that the wires could be accessed for EMG recording at a later date when the animal was again under anesthesia.

B. Breath Detection Algorithm

1) Conditioning of Signal

The EMG signal was amplified, filtered and passed into an analog to digital (A/D) converter sampled at 2 kHz, such that x_i was the value of the most recent sample, and x_{i-1} was the value sampled just prior to x_i. The first digital processing step was to correct for any DC bias and to rectify the values.

$$y_i = |x_i - o| \quad \textbf{(1)}$$

where y_i is the offset corrected value, i is the index of the sample, and o is the DC offset value that was determined during baseline calculations described below (eq. 7).

The next step clipped the signal to lie within a boundary rail of expected values. The purpose of the boundary rail was to minimize the influence of any momentary large spikes in the signal upon the windowed mean calculations described below. Such large momentary spikes could derive from a transient noise source or the electrocardiogram signal, depending upon the location of the electrodes on the diaphragm muscle in relation to the heart. All EMG signals related to the contraction of the diaphragm muscle during the inspiratory part of a regular breath were expected to have a magnitude much lower than the value of the boundary rail, r.

$$z_i = \left| \begin{array}{l} y_i \le r : y_i \\ y_i > r : r \end{array} \right. \quad \textbf{(2)}$$

where z_i is the boundary restrained EMG value, r is the rail that defines the maximum positive and negative values allowed. The value for r was determined during baseline calculations described below (eq. 8).

2) Mean over a Window

To rectify and integrate the values over a moving window of time, the mean value for the most recent m samples was calculated.

$$w_i = \sum_{k=i-m+1}^{i} z_k \Big/ m \quad \textbf{(3)}$$

where m is the number of samples in the window, and w_i is the mean value for the window that corresponds to sample point z_i. To minimize the calculation cost of eq. 3, w_i was calculated using a running sum value, a "first in last out" type stack containing the z values in the current window, and a binary right shift operation was utilized to execute the division (eq. 4-6).

$$s_i = s_{i-1} - z_{i-m} + z_i \quad \textbf{(4)}$$

where s_i is the running sum for all the z values within the window that corresponds to z_i. If the window size, m, is restricted to have power of 2 values (e.g., 4, 8, 16, 32, 64, etc.), then p could be defined as

$$m = 2^p \quad \textbf{(5)}$$

To calculate w_i, instead of dividing s_i by m, a binary right shift operation was done.

$$w_i = s_i >> p \quad \textbf{(6)}$$

In the current implementation, the EMG signal was sampled at 2 kHz, m was 16 samples (8 ms window), and p was 4.

3) Baseline Calculations

Recordings of the EMG signal were processed prior to the start of the breath detection algorithm to determine certain baseline values that were then used to process and analyze each new sampled value by the breath detection algorithm. In the current implementation, the baseline calculation was only determined during several initial 66 second periods of passive monitoring of the EMG signal, but for a long term application, such as an implanted device, the EMG signal could be constantly monitored to periodically adjust the baseline values. For instance, periodic baseline adjustments would help account for long-term changes in the sampled signal that may result from increased connective tissue around the electrodes.

For the baseline calculations, the EMG signal was sampled and processed for a total of n samples. Each new EMG value was received and utilized in the next three equations. The DC offset value (o; eq. 1) was calculated as

$$o = \sum_{i=0}^{n} x_i \Big/ n \quad \textbf{(7)}$$

The value for the boundary rail (r; eq. 2) was calculated using the mean window value (w_i; eq. 3).

$$r = q \times Max(w_i) \quad \textbf{(8)}$$

where the maximum value of all the w_i values is multiplied by a factor q. In the current implementation, q was set at 4.

A threshold value, t, was compared against w_i values to help determine changes in the state of the breath detection algorithm. For example, a sequence of w_i values above the threshold may indicate inspiration, and a sequence of w_i values below the threshold may indicate expiration. This threshold was calculated as the average value after offset correction and rectification (y_i; eq. 1).

$$t = \sum_{i=0}^{n} y_i \Big/ n \quad \textbf{(9)}$$

4) States

The breath detection algorithm consisted of a series of states (Fig. 1). At initialization, the *Baseline* state was entered and the EMG signal was passively monitored to determine the DC offset (o; eq. 7), boundary rail (r; eq. 8) and threshold (t; eq. 9) values. Upon completion, the *Look for Exhale* state was entered, one of six states that the algorithm cycled through from this point forward. For each new EMG sample (x_i), the window mean value (w_i; eq. 6) was calculated (Fig. 1).

If w_i was below threshold (t; eq. 9), the algorithm switched to the *Putative Exhale* state, and a counter variable, c, was reset to 0 (Fig. 1). For the algorithm to pass from the *Putative Exhale* state to the *Ready for Inspire* state, the value for w_i had to remain below the threshold t for a certain number of times, R_e. In the current implementation R_e was set at 35. Thus, 35 successive w_i values had to remain below threshold t in the *Putative Exhale* state (Fig. 1), before the algorithm advanced to the next state.

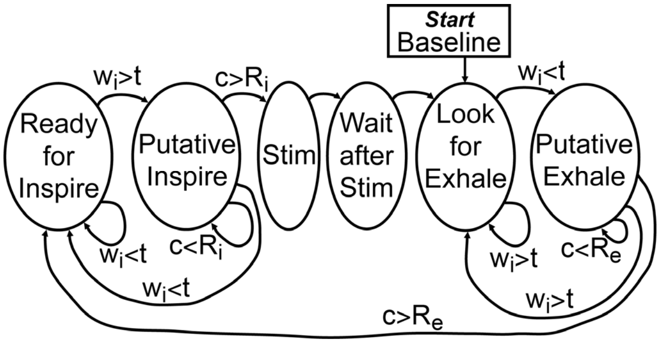

Fig. 1. State Diagraph for the breath detection algorithm, where w_i is the window mean, t is the threshold, c is a count variable, R_i is the number of repeats to indicate an inspiration, R_e is number of repeats to indicate an expiration.

Once in the *Ready for Inspire* state, the algorithm began to look for activity that might indicate inspiration. If a next w_i value was above threshold t, the algorithm was able to advance to the next state, *Putative Inspire*, and the counter c was reset to 0 (Fig. 1).

For the algorithm to pass from the *Putative Inspire* state to the *Stimulate* (*Stim*) state, the value for w_i had to remain above the threshold t for a certain number of times, R_i, also set at 35 in the current implementation (Fig. 1).

During the *Stim* state, the algorithm would initiate the generation of inspiration. Depending on the final application, a mechanical ventilator might be triggered or a train of stimulation pulses might be delivered to a paralyzed phrenic nerve or diaphragm muscle. To simulate the generation of inspiration in the current implementation, a stimulation voltage signal was held high for 200 ms. After this period, the stimulation signal was turned off, the *Wait after Stim* state was entered, and the count c was reset to 0 (Fig. 1).

The *Wait after Stim* state imposed a minimum period after stimulation in which another stimulation signal could not be asserted. In the current implementation, this minimum period was 25 ms. After the wait, the *Look for Exhale* state was entered. The algorithm continued through these series of states as described above (Fig. 1).

C. Setup to Test Algorithm

The algorithm was implemented in a LabView program running on a desktop computer, and in a C program running on a microcontroller chip. The EMG signal coming directly from the electrodes in a rat was amplified, filtered and passed into either the A/D converter sampling at 2 kHz within the desktop computer data acquisition board or within the microcontroller chip in the device. The stimulation signal generated by the algorithm running on the device was captured simultaneously by the data acquisition board and stored in a file for later analysis to determine the number of correct and incorrect detection of breaths by the algorithm.

III. Results

In the current implementation of this algorithm running on a microcontroller chip, 48 instruction clock cycles were required to process each new sample, x_i, to digitally condition the sample (eq. 1-2) and to calculate the mean window value (eq. 4,6). The full computational cost to process each new sample (eq. 1,2,4,6) and handle the state determination (Fig. 1) ranged from 88 to 99 instruction clock cycles. If a future implementation would continually monitor the EMG signals to periodically adjust the baseline parameters (eq. 7-9), some additional instruction clock cycles would be required for each new sample.

Fig. 2 shows the EMG signal and output of the breath detection algorithm implemented in the LabView program. The small size of the window (8 ms) did not mask out the spikes from the electric signals related to the heart beat, but the repeat value R_i of 35 (17.5 ms) ensured that the window mean was above the threshold successively for 17.5 ms, which did mask out the spikes related to the heartbeat. The design tradeoff was that the wider the window and more required repeat values R_i, the greater the accuracy of the breath detection algoritm, but also the greater the delay between the actual start of the inspiration signal and the detection of that signal by the algorithm and assertion of the stimulation signal.

The microcontroller device also processed the EMG signal and asserted a stimulation signal (Fig. 3). This test was made for two cases, one when the rat was lying on its side and undisturbed (Fig. 3A), and the other when the rat was gently wiggled to simulate some aspects of the noise that would be added to the EMG signal from an awake and actively moving animal (Fig. 3B). To determine whether the algorithm correctly asserted the stimulation signal during a breath, the EMG signal was visually compared with the

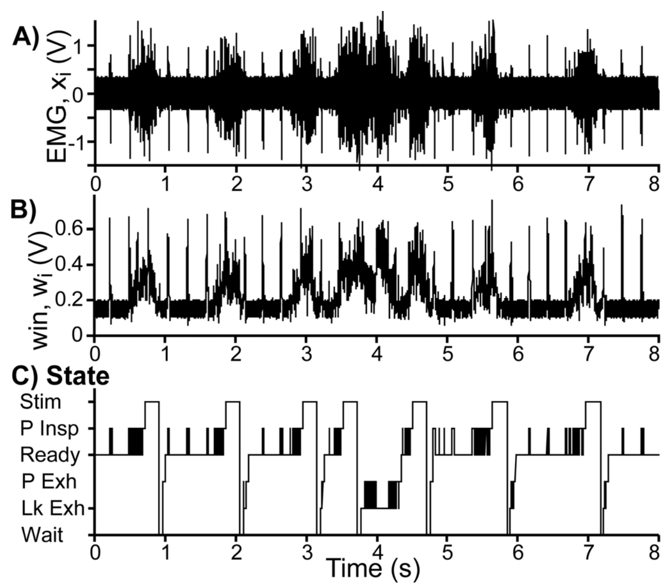

Fig. 2. Labview implementation of the breath detection algorithm. The EMG signal (A) was processed to calculate a windowed mean, w_i, over each 8 ms (B). The algorithm output was plotted in (C), with Stim indicating a period when an inspiration could be generated, Wait indicating a minimum period of no activity after Stim, Lk Exh indicating the Look for Exhale state, P Exh indicating the Putative Exhale state, and P Insp indicating the Putative Inspire state.

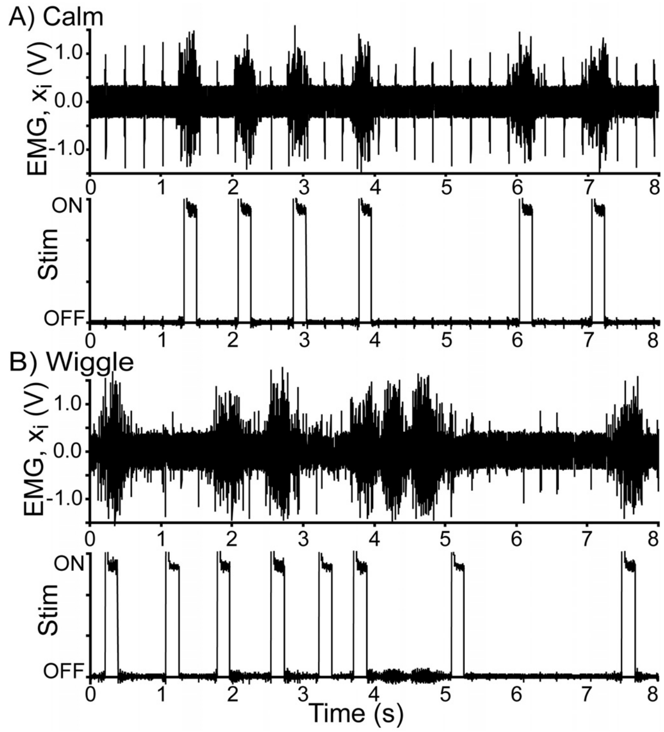

Fig. 3. Microcontroller device implementation of the algorithm showing the EMG signal and the device output of a stimulation assert signal (Stim) for two test cases. Case when anesthetized rat was calm and undisturbed is plotted in A), and case when rat was gently wiggled is plotted in B). The stimulation signal at the ON level would indicate the trigger for artificial inspiration of a breath.

stimulation assertion signal as in Fig. 3. For the case when the rat was calm and undisturbed, the EMG was analyzed for a 410 second period, within which 361 breaths occurred. Under these conditions, the device correctly detected over 99% of the breaths. Out of the 361 breaths, 1 missed breath occurred where the device did not initiate a stimulation signal, and 1 false positive occurred, where a stimulation signal was initiated, but the EMG signal indicated no breath (Table 1).

Table 1: Breath Detection Success Rate

	Missed	False Positive
Calm	<1%	<1%
Wiggled	19%	6%

For the case when the rat was gently wiggled, the EMG signal was monitored for 160 seconds, during which 162 breaths occurred. The success rate of the breath detection algorithm was much lower for this more challenging condition: 30 breaths were missed and 10 false positives occurred (Table 1).

IV. Discussion

The algorithm was developed and tested on a series of EMG signals recorded from one rat. To determine whether the algorithm is adaptable and robust, EMG signals from a number of rats over a longer duration will have to be tested.

This algorithm was developed to chronically reactivate the paralyzed half of the diaphragm muscle in an experimental animal model [12]. The algorithm may have wider applicability for triggering each breath generated by a mechanical ventilator [9, 10] or implanted diaphragm pacing system [2, 4, 11]. The closer correlation between the artificially generated breaths and the brain directed drive to breath may alleviate some of the problems related to mechanical ventilation [9, 10] and may improve quality of life [7].

In the implemented system, the diaphragm muscle EMG signal was monitored. For cases of diaphragm paralysis, other potential CAP sources to monitor may include the hypoglossal nerve [8] or the posterior cricoarytenoid muscle [7].

References

[1] H. F. B. Tripp, J.W.R., "Phrenic nerve injury following cardiac surgery: a review," *Journal of Cardiac Surgery*, vol. 13, pp. 218-23, 1998.

[2] A. F. O. DiMarco, R.P.;Ignagni, A.; Kowalski, K.E.; Mortimer, J.T., "Phrenic nerve pacing via intramuscular diaphragm electrodes in tetraplegic subjects," *Chest*, vol. 127, pp. 671-8, 2005.

[3] J. A. H. McCaul, S., "Transient hemi-diaphragmatic paralysis following neck surgery: report of a case and review of the literature," *Journal of the Royal College of Surgeons of Edinburgh*, vol. 46, pp. 186-8, 2001.

[4] M. L. T. Chen, M.A.; Kun, S.; Keens, T.G., "Diaphragm pacers as a treatment for congenital central hypoventilation syndrome," *Expert Review of Medical Devices*, vol. 2, pp. 577-85, 2005.

[5] T. P. Vassilakopoulos, B.J., "Ventilator-induced diaphragmatic dysfunction," *American Journal of Critical Care Medicine*, vol. 169, pp. 336-41, 2004.

[6] G. D. Gayan-Ramirez, M., "Effects of mechanical ventilation on diaphragm function and biology," *European Respiratory Journal*, vol. 20, pp. 1579-86, 2002.

[7] H. G. Lanmuller, W.;Sauermann, S.;Schwendenwein, I.;Unger, E.; Bijak, M.;Rafolt, D.; Mayr, W., "Long-term electromyogram recording from the posterior cricoarytenoid muscle as a potential biological trigger for phrenic pacing: results of an animal study," *Artificial Organs*, vol. 23, pp. 860-8, 1999.

[8] W. M. P. St-John, J.F.R.;Leiter, J.C., "Uncoupling of rhythmic hypoglossal from phrenic activity in the rat," *Experimental Physiology*, vol. 89, pp. 727-37, 2004.

[9] J. E. G. Remmers, H., "Servo respirator constructed from a positive-pressure ventilator," *Journal of Applied Physiology*, vol. 41, pp. 252-5, 1976.

[10] C. N. Sinderby, P.;Beck, J.;Skrobik, Y.;Comtois, N.;Friberg, S.;Gottfried, S.B.; Lindstrom L., "Neural control of mechanical ventilation in respiratory failure," *Nature Medicine*, vol. 5, pp. 1433-6, 1999.

[11] W. W. L. P. Glenn, M.L., "Diaphragm pacing by electrical stimulation of the phrenic nerve," *Neurosurgery*, vol. 17, pp. 974-84, 1985.

[12] H. Z. Miyata, W.Z.;Prakash, Y.S.;Sieck, G.C., "Myoneural interactions affect diaphragm muscle adaptations to inactivity," *Journal of Applied Physiology*, vol. 79, pp. 1640-9, 1995.