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„Measurement of respiratory activity during abdominal muscle stimulation“

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Measurement of respiratory activity during abdominal muscle stimulation

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Introduction

Respiratory complications are a leading cause of morbidity and mortality in people with spinal cord injury (SCI) [1]. People with SCI in the cervical region, affecting all four limbs and resulting in tetraplegia, are at increased risk of respiratory complications as their injury also results in partial or complete paralysis of the breathing muscles. While high-level tetraplegia can lead to complete or partial loss of function of the diaphragm, tidal volume is further reduced by intercostal paralysis, and cough peak flow is compromised due to abdominal paralysis [2]. In particular, the reduction in cough peak flow affects the ability to clear the airways which contributes to an increased likelihood of respiratory infections in individuals with chronic tetraplegia.

Respiratory function in this patient group can be improved by electrical stimulation of the abdominal muscles [3, 4] which results in a uniform, well-defined response to support expiration [5]. Cough is enhanced as the contraction of the stimulated abdominal muscles leads directly to an increase in respiratory pressure and consequently to improved expiratory flow. Tidal volume is increased during normal breathing due to expiration below the functional residual volume. Stimulation is typically delivered through surface electrodes that are easy to apply.

Abdominal muscle stimulation for respiratory support is mainly used in tetraplegic individuals who can breathe spontaneously and therefore needs to be synchronised with their voluntary respiration. Limited hand function resulting from tetraplegia usually restricts any interface which is to be operated independently of a caregiver to simple systems such as buttons or chin-controlled joysticks Automatic systems can ensure appropriate timing of the stimulation, using a measurement of the subject's respiratory activity such as a direct observation of the airflow at the mouth and nose [6]. In experiments with tetraplegic subjects (C4-C6, complete) it was demonstrated that this approach can lead to marked increases in tidal volume and cough peak flow [3].

While a direct measurement of the airflow at the mouth is feasible in a laboratory setting where it also allows the collection of spirometry data, this approach is not suitable for use outside a research environment. Spivak et al. [7] suggest an EMG-activated stimulation system for abdominal stimulation independent of a caregiver. Another possible approach to obtain a measurement for the respiratory activity include the direct observation of the abdominal and chest movement using appropriate sensors such as plethysmographic respiratory effort belts. While these sensors are routinely used in sleep studies to collect data for offline analysis [8], their suitability for real-time measurement of respiration in the presence of electrical stimulation has not yet been investigated.

This paper presents a respiratory effort sensor system based on plethysmographic belts to measure breathing activity online in the presence of abdominal muscle stimulation. Measurements from an evaluation in a healthy subject are presented to demonstrate the feasibility of this approach.

Materials and Methods

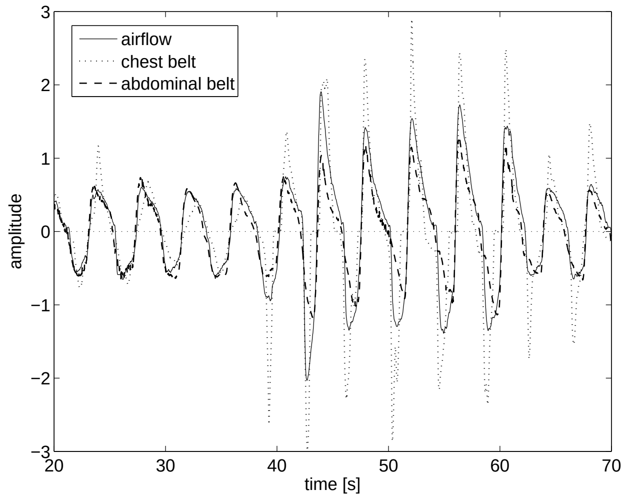
Two piezoelectric respiratory effort sensors (Pro-Tech, Pittsburgh, USA) were worn around the subject's abdomen and chest. They were connected to a PC via a custom-made differential signal amplifier and a data-acquisition card (DAQCard-6024E, National Instruments, Texas, USA). The respiratory effort belts generate a small electric charge when they are stretched. This was used to detect the small movements of the abdomen and of the chest during breathing.

To obtain a direct measurement of the respiratory activity of the subject, a handheld spirometer (Microloop, Micromedical, Chatham, UK) with a low dead-space full face mask (Hans Rudolph Inc., Missouri, USA) was used to record airflow at the mouth and nose of the subject. The spirometer data was connected to the PC via a RS232 interface, the allowing acquisition of respiratory flow in real-time. The data from the spirometer were used as a reference measurement to evaluate how accurately the signals from the respiratory effort belts could represent the respiratory activity in various breathing situation.

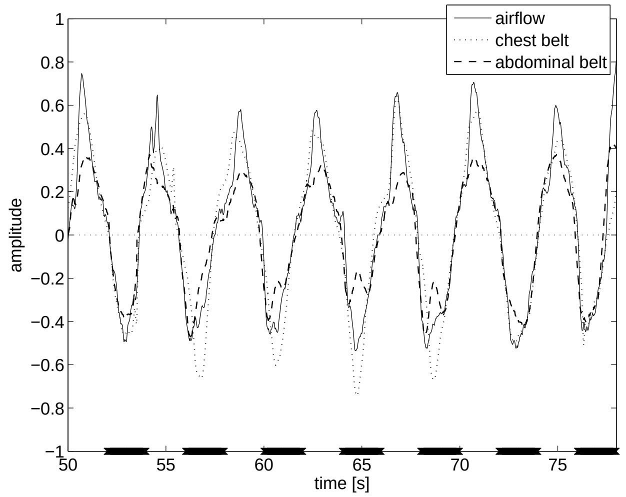
To investigate to what extend the measurements from the respiratory effort belts was influenced by electrical stimulation, the abdominal muscles were stimulated bilaterally, using two stimulation channels. Self-adhesive surface electrodes (PALS, Axelgaard) were placed over the mm. rectus abdominis on both sides of the umbilicus and connected to a neuromuscular stimulator (RehaStim, Hasomed GmbH, Magdeburg, Germany). The timing and parameters of the stimulation pulses were controlled from the PC via an isolated USB interface. The stimulator delivered biphasic, charge-balanced, current controlled pulses at a frequency of 50Hz with a constant current of 20mA and a variable pulse-duration in the range of 100-400μs. The stimulation intensity was selected in such a way that a strong muscle contraction could be observed, but was limited by sensation of the subject. Stimulation was delivered for 2 s every 4 s. The subject was asked to adjust their breathing patterns in such a way that air was expeled during muscle stimulation.

Results

Figure 1a shows the airflow at the mouth and nose together with the signals from the respiratory effort belts at the chest and the abdomen. Positive signals correspond to inspiration while a negative signal indicates expiration. The timing and the shape of the belt signals correspond well with the airflow data throughout the measurement. During heavier breathing (from approximately 42 sec onwards) the amplitudes of the belt signals increase relative to the airflow measurement.



(a) Without stimulation.



(b) With stimulation.

Figure 1: Respiratory effort belts and airflow. Airflow amplitude is in L/s while the belt signals are presented in normalised units.

Figure 1b shows results obtained with stimulation. Periods when stimulation was applied are indicated by the bold black lines at the bottom of the figure. It can be clearly seen that the contraction of the abdomen due to stimulation is represented by an additional peak in the abdominal belt signal during expiration (e.g. at 65 and 69 sec), and that no additional electrical artifacts are present.

Discussion and Conclusions

The aim this study was to develop a system which can measure respiratory activity online with appropriate accuracy to represent the main characteristics of various breathing situations without obstructing the subject's mouth or nose. The system should also provide accurate measurements in the presence of abdominal muscle stimulation.

The results demonstrate that respiratory effort belts can accurately represent breathing activity in a variety of situations. The custom-built signal amplifier sufficiently attenuates any electrical artifacts from the stimulation which might influence the piezoelectric belt sensors.

This preliminary investigation indicates that the signal from one belt alone may not be sufficient to characterise the respiratory pattern for different breathing situations. A suitable combination of the belt signals might be required to provide an accurate measurement of respiration which can be used to automatically synchronise abdominal stimulation with the subject's voluntary breathing. When used with tetraplegic subject, their modified breathing patterns needs to be taken into consideration for this.

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