A FEEDBACK SYSTEM FOR AUTOMATIC ELECTRICAL STIMULATION OF ABDOMINAL MUSCLES TO ASSIST RESPIRATORY FUNCTION IN TETRAPLEGIA

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Abstract

People with tetraplegia have poor respiratory function leading to limited tidal volume (VT) and reduced cough peak flow (CPF). These problems may cause respiratory failure during both the initial admission and subsequent intercurrent illness. Electrical stimulation of abdominal muscles during expiration can improve respiratory function by increasing VT and CPF. We developed a feedback control system to automatically trigger muscle stimulation, synchronised with the subject's voluntary respiratory activity. The system was tested in a single-subject proof-of-concept study. Significant increases in VT and CPF were observed suggesting that the technique may have potential use in both acute and established tetraplegia to increase minute ventilation and to improve cough clearance of secretions.

Introduction

People with tetraplegia have poor respiratory function due to: 1) intercostal paralysis causing reduced tidal volume (VT) and 2) abdominal paralysis reducing cough peak flow (CPF) [1]. These problems may cause respiratory failure during both the initial admission and subsequent intercurrent illness. It has been shown that electrical stimulation of abdominal muscles during expiration can improve respiratory function by increasing VT and CPF [2, 3]. While the increase in CPF is a direct result of the improved respiratory pressure due to the additional input from the abdominal muscles, the increase in VT can be attributed to the reduction of lung volume below the functional residual capacity and the subsequent passive recoil during inspiration.

Stimulation of the abdominal muscles is normally used in tetraplegic individuals with spontaneous breathing (although studies exist with individuals who are unable to breath spontaneously [4]). Stimulation therefore needs to be synchronised with their voluntary breathing activity. In previous studies manual intervention is typically needed to trigger the stimulation. This can be done either by the therapist [5] or by the individual, using for example a chin-controlled joystick [6].

Stanic et.al [7, 8] used a system to trigger the stimulation automatically depending on the individual's spontaneous breathing pattern. A measurement of the airflow at the mouth was used to determine the onset of expiration. The experiments reported are limited to quiet breathing.

This study is aimed at developing a feedback control system to automatically trigger muscle stimulation, synchronised with the subject's voluntary respiratory activity. The system works over a wide variety of breathing patterns, and it does not interfere with non-regular respiratory patterns (for example during speaking). It is designed to detect different situations such as quiet breathing and coughing which require adjustment of the stimulation pattern. While initially the flow at the mouth was used directly to generate the trigger signal, preliminary experiments suggest the suitability of other sensors such as a plethysmographic belt which are not located at the mouth and therefore do not interfere with other activities.

Methods

One subject (16-year-old, male, C4 complete, one year post injury) participated in the study. He breathes spontaneously but with reduced VT and CPF. The subject is in the normal sitting position in his wheelchair during the experiments.

We stimulated abdominal muscles bilaterally using self-adhesive surface electrodes (PALS, Axelgaard, 33mm$\times$53mm rectangular and 50mm round). Four stimulation channels were used: two channels stimulated the mm. rectus abdominis, while the other two channels stimulated the lateral abdominal muscle group (mm. transversi and mm. obliqui ext. et int.) on both sides, cf. figure 1. The stimulation parameters were controlled from a laptop PC through a RS232 interface. A stimulation frequency of 50Hz was used. Monophasic charge-balanced stimulation pulses were delivered with a constant current of 70mA and variable pulsewidths of 200-500μs. The pulsewidths were adjusted separetly for the mm. rectus abdominis and the lateral abdominal muscle group.

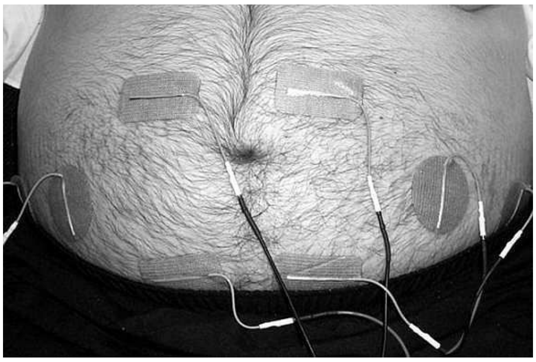

Figure 1: Placement of the electrodes.

A feedback control algorithm was used to automatically generate a trigger signal for stimulation which is synchronised with the voluntary respiratory activity. The respiratory activity was determined by direct measurement of air-flow at the mouth, using a spirometer (Microloop, Micromedical) with an RS232 interface. A plethysmographic belt (Protech) which measures changes in abdominal girth was used as an alternative way to observe the respiratory activity. It was connected to the PC using a custom-built signal amplifier and a data-acquisition card. Whereas measuring the air-flow directly at the mouth requires the subject to wear a face mask, the plethysmographic belt is simply worn around the abdomen.

Feedback control algorithms

The control algorithm uses a feedback signal of the respiratory activity (air-flow or abdominal girth) to generate a suitable stimulation signal which does not interfere with the voluntary breathing. The main task is to detect the end of inspiration or the beginning of expiration, and to synchronise the artificial stimulation signal accordingly.

Two approaches to generate the stimulation trigger signal from the measured respiratory activity were used: During quiet breathing, the onset of expiration was determined by observing the derivative of flow with respect to time. If this value exceeded a pre-set threshold, a zero-flow crossing from inspiration to expiration was detected. The stimulation was triggered at a specified time (usually 0.2s) after the onset of expiration. The maximal duration of one stimulation burst was limited to 1.5s. The stimulation burst would be terminated when inspiration was detected.

A modified algorithm was used for coughing, where stimulation was applied at the end of voluntary inspiration which was detected by observing the derivative of flow with respect to time, taking into account that the maximal inspired flow must exceed 1.0 l/s to be classified as an attempt to cough. The subject was instructed to hold his breath for a short time after inspiration to allow the abdominal muscles to contract and the pressure to build up before beginning of expiration. The length of the stimulation burst was set to 1.5s.

Results

The subject tolerated several minutes of stimulation without discomfort or other adverse effects. He could speak normally without inappropriate triggering of the stimulation.

Typical experimental results for quiet breathing, forced expiration and coughing are shown in figures 2-4, using the air-flow signal from the spirometer to generate the stimulation trigger.

Note that positive flow corresponds to expiration (volume increase), while a negative flow relates to inspiration (volume decrease). The bold horizontal lines indicate when stimulation was applied.

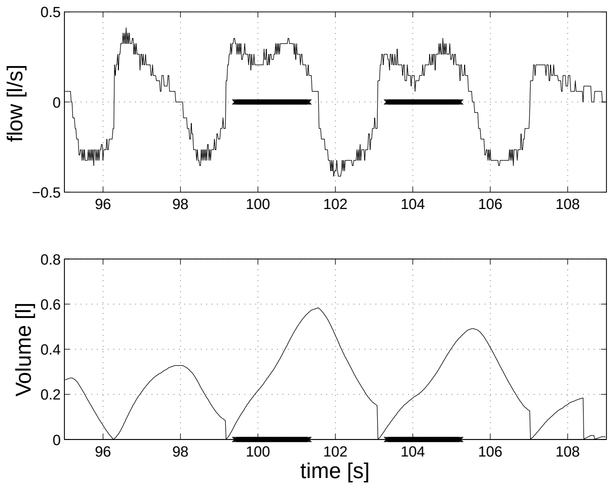

Figure 2: Quiet breathing. The bold lines indicate when stimulation is applied.

Figure 2 shows four quiet breaths, extracted from an experiment of approximately 3min duration. Stimulation is applied during the second and third breaths. The flow diagram shows a large additional expiratory flow as a result of the stimulation of the abdominal muscles. The tidal volume increases from approximately 0.3 l without stimulation to over 0.55 l with stimulation, cf. the bottom graph in figure 2.

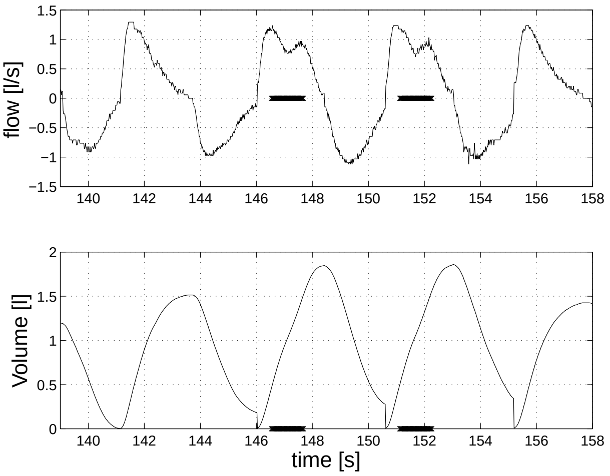

Figure 3: Forced expiration. The bold lines indicate when stimulation is applied.

The results shown in figure 3 were obtain when the subject was instructed to breath out forcefully. The four breaths shown here are extracted from an experiment of 3min duration. Stimulation is applied during the second and third breath. A larger expiratory flow can again be observed as a result of the stimulation. The maximal tidal volume increases from 1.5 l without stimulation to 1.85l when stimulation is applied.

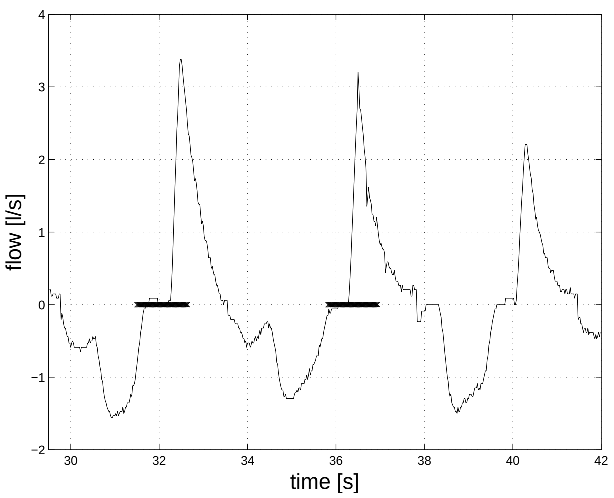

Figure 4: Coughing. The bold lines indicate when stimulation is applied.

A typical experimental result for coughing is shown in figure 4. The aim for assisted cough is to maximise the flow to allow better clearing of the airways. Tidal volume has therefore been omitted from the graph. Three coughs are shown: the first two coughs are with assistance from abdominal stimulation, while the final cough is unassisted. A clear increase in CPF can be observed when stimulation is applied. Note that the increase was consistently observed, independently of whether assisted cough preceded or followed unassisted coughs. We recorded an increase of CPF from 2.6 l/s to 3.4 l/s.

Plethysmographic Belt

Results of initial experiments with a plethysmographic belt to measure breathing activity are shown in figure 5. Stimulation is applied during the first two

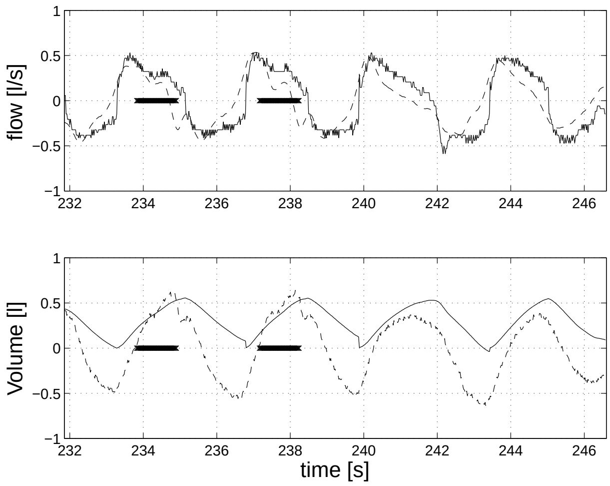

Figure 5: Quiet breathing, with plethysmographic belt measurements (dashed lines, no units). The bold lines indicate when stimulation is applied.

breaths shown, while the last two breaths are unassisted. In addition to the flow and volume traces, the figure also shows the measurements obtained from the belt. The dashed line in the bottom plot corresponds to the raw belt signal which is proportional to abdominal girth and therefore directly related to tidal volume. The dashed line in the top plot shows the differentiated belt signal which is a measurement of the rate of change of the abdominal girth and therefore related to the air-flow. In both plots it can be observed that the relevant belt signal relates well to the volume and flow signal. The feedback signal from the belt was used here to generate the stimulation signal.

Discussion

The results show that the closed loop system presented here can automatically ensure that abdominal stimulation is synchronised with the subject's own respiratory activity. Control algorithms for quiet breathing, forced expiration and for coughing have been implemented and experimentally evaluated. The results show a clear increase in VT (during quiet breathing and forced expiration) and CPF (during coughs). Direct observation of the air-flow at the mouth provides an accurate measure for the breathing activity and can be used as a feedback signal. Any measurement at the mouth interferes, however, with other activities and is therefore not suitable for a system which is to be used in everyday situations. We suggest that a plethysmographic belt which measures abdominal girth can be used to provide an accurate measure of the respiratory activity and can be used as a feedback signal to the stimulation control algorithm. Our initial results show the suitability of this approach.

This proof of concept technique may have potential use in both acute and established tetraplegia to increase minute ventilation and to improve cough clearance of secretions. Besides improving minute ventilation, continuous stimulation during quiet breathing can be used to strengthen the abdominal muscle, which could lead to further improvements in CPF.

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