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**Automatic electrical stimulation of abdominal wall muscles increases tidal volume and cough peak flow in tetraplegia**

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## Abstract

Paralysis of the respiratory muscles in people with tetraplegia affects their ability to breathe and contributes to respiratory complications. Surface functional electrical stimulation (FES) of abdominal wall muscles can be used to increase tidal volume ( $V_T$ ) and improve cough peak flow (CPF) in tetraplegic subjects who are able to breathe spontaneously.

This study aims to evaluate the feasibility and effectiveness of a novel abdominal FES system which generates stimulation automatically, synchronised with the subjects' voluntary breathing activity. Four subjects with complete tetraplegia (C4-C6), breathing spontaneously, were recruited.

The automatic stimulation system ensured that consistent stimulation was achieved. We compared spirometry during unassisted and FES-assisted quiet breathing and coughing, and measured the effect of stimulation on end-tidal  $\text{CO}_2$  ( $\text{EtCO}_2$ ) during quiet breathing.

The system dependably recognised spontaneous respiratory effort, stimulating appropriately, and was well tolerated by patients. Significant increases in  $V_T$  during quiet breathing (range 0.05-0.23L) and in CPF (range 0.04-0.49L/s) were observed. Respiratory rate during quiet breathing decreased in all subjects when stimulated, whereas minute ventilation increased by 1.05-2.07L/min. The changes in  $\text{EtCO}_2$  were inconclusive.

The automatic stimulation system augmented spontaneous breathing and coughing in tetraplegic patients and may provide a potential means of respiratory support for tetraplegic patients with reduced respiratory capacity.

**Keywords:** tetraplegia, pulmonary function, electrical stimulation, automatic control system

# 1 Introduction

Cervical spinal cord injury (SCI) can result in dysfunction in both the lower and upper limbs (tetraplegia), and may be accompanied by a range of secondary complications such as reduced breathing function due to paralysis of the respiratory muscles. As a result, pulmonary complications remain a significant cause of both acute and long term morbidity in this patient group [16].

High complete tetraplegic patients (with an injury at level C1-C2) require long-term ventilatory support. Individuals with intact phrenic nerve pathways may be suitable for phrenic nerve pacing [9]. Other ventilatory support options for this group include direct diaphragmatic pacing [5], high thoracic spinal cord stimulation in combination with phrenic nerve pacing [6], and low thoracic spinal cord stimulation for cough assistance [4].

Patients with lower levels of tetraplegia may require supportive ventilation within the first few days of injury and, although most will wean from the ventilator, these people remain at life long risk of respiratory compromise [3]. Paralysis of the intercostal muscles and abdominal wall muscles reduces vital capacity and the efficiency of coughing in this group.

Previous authors [15,20,21] have described surface functional electrical stimulation (FES) of abdominal wall muscles to increase tidal volume and improve cough in tetraplegic subjects who are able to breath spontaneously. These studies established abdominal surface FES augmentation of breathing and cough but were limited in their potential application. The systems required a manual trigger to synchronise the stimulation with the voluntary breathing activity. This reduced the accuracy and consistency of stimulation and limited the time over which stimulation could be applied. Šorli et al. [22] and Spivak et al. [19] have demonstrated that stimulation could be automatically synchronised with natural respiratory activity for specific respiratory situations. While the study reported in [22] focused on quiet breathing their system was not applicable for cough induction. On the other side, the study in [19] focused on a system to generate an automatic trigger for cough, but did not consider other breathing situations.

In the present study we used a novel control system [11] for automatic and appropriate triggering of abdominal surface FES in tetraplegia, designed to increase both tidal volume ( $V_T$ ) during normal breathing, and peak flow during cough (CPF) while allowing the subject to speak and breathe spontaneously. This allows, for the first time, automatic abdominal stimulation over periods of time exceeding several minutes, ensuring that the stimulation is delivered consistently in synchrony with the subject's voluntary respiration.

While the functionality of the automatic system has been demonstrated for short periods of time (over several breaths) in [11], the aim of the present study was to evaluate this system in repeated experiments with a number of tetraplegic subjects and to observe the effect of abdominal FES for prolonged stimulation periods. During quiet breathing we were able to detect changes in tidal volume, respiratory rate and minute ventilation, while changes in CPF could be monitored during continuous coughing. In addition we recorded end-tidal partial pressure of  $\text{CO}_2$  in expired air ( $\text{EtCO}_2$ ).

## 2 Materials and Methods

### 2.1 Study design and subjects

The study was designed as a prospective, proof-of-concept, single cohort trial. Four tetraplegic subjects attending the Queen Elizabeth National Spinal Injuries Unit were recruited. The subjects were in good general health and were non-ventilator dependent with no major co-morbidities. All had a functionally complete injury with no voluntary power or sensation below the level of injury. Three subjects, S1, S2 and S3, required ventilation at the time of their initial injury. Subject details

are summarised in Table 1.

Subject	age	sex	injury level	time post injury
<b>S1</b>	37 years	male	C4	9 months
<b>S2</b>	16 years	male	C4	12 months
<b>S3</b>	49 years	female	C5	60 months
<b>S4</b>	45 years	male	C6	3 months

Table 1: Subject details

All applicable institutional and governmental regulations concerning the ethical use of human volunteers were followed in the course of this research.

## 2.2 Stimulation System and Measurements

A system of automatic abdominal wall stimulation was designed to provide two main functions; augmentation of spontaneous tidal volume during quiet breathing, and increased cough impulse.

During quiet breathing, stimulation was triggered by the start of expiration, causing the abdominal wall muscles to contract and forcing the subject to exhale below the functional residual capacity. At the end of expiration the stimulation stopped and the thorax elastically recoiled, resulting in the next breath being augmented by the extra volume.

For cough support, stimulation was applied at the end of inspiration with the glottis closed, resulting in increased pressure as a result of the abdominal muscle contraction and augmenting the subject’s own active cough impulse.

A four channel stimulator (Motive 8, Stanmore, UK) was used to stimulate the major muscle groups bilaterally. At each site two pairs of self-adhesive surface electrodes (PALS, Axelgaard, Fallbrook, CA, USA) were placed over rectus abdominis muscles medially, and two pads over transversalis abdominis and external and internal oblique muscles laterally. The stimulation currents (range 30-100mA) and pulsewidths (up to 400 $\mu$ s) were set individually for each subject to obtain a good movement of the abdominal wall, which resulted in an additional expiratory airflow during quiet breathing. Stimulation intensity was increased until either the current and pulsewidth limits were reached, or undesired co-stimulation of other muscle groups and/or strong spasticity was observed. A stimulation frequency of 50Hz was used, resulting in rapid and smooth muscle contractions. The respiratory data was recorded at the same sample frequency.

The respiratory activity was detected by measuring the airflow at the mouth using a spirometer (Microloop, Micromedical, Chatham, UK) with a low dead-space full face mask (Hans Rudolph Inc., Kansas City, Missouri, USA). A PC acquired the measurements from the spirometer in real-time and controlled the stimulator via separate RS232 interfaces.

Software control algorithms were designed to allow the system to automatically trigger the stimulation consistently at an appropriate point in the respiratory cycle. A simplified block diagram of the setup is shown in figure 1.

Based on the respiratory activity (in this case the airflow at the mouth as measured by the spirometer), the system was able to detect an attempt to cough or a quiet breathing situation: (i) If the subject took a deep breath which stopped abruptly before expiration an attempt to cough was detected and a strong stimulation burst was delivered at the end of inspiration to support the built-up of intra-abdominal pressure required for coughing. (ii) If the respiratory activity was sufficiently similar to a previously recorded reference pattern of quiet breathing then stimulation was applied at the beginning of expiration to increase tidal volume. Stimulation was muted when none of these two situations was detected.

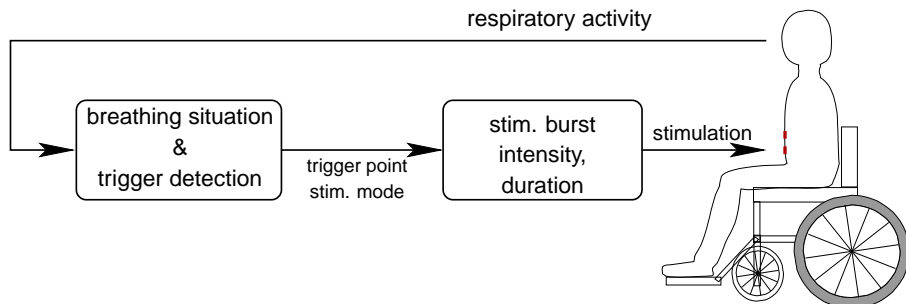


Figure 1: Simplified control structure for automatic abdominal stimulation: From the measured respiratory activity signal the stimulation mode (cough or quiet breathing) and the corresponding trigger point are detected which determine the parameters (intensity, duration) of the stimulation burst.

The implemented algorithm can be tuned towards an individual subject by adjusting a number of key parameters: For the detection of cough, an inspiratory cough threshold,  $\tau_{flow}$ , had to be exceeded, together with a flow derivative threshold,  $\tau_{dflow}$ , marking the end of inspiration. A cough stimulation with a duration of  $T_{p,c}$  was then applied. If no cough was detected, the breath was compared to a reference pattern and stimulation was applied if the cross-correlation coefficient exceeded a threshold,  $\tau_{xcorr}$ . The duration of the stimulation burst was set by  $T_{p,q}$ . The algorithm parameters used in the present study are summarised in table 2, while full details of the implementation can be found in [11]. .

Subject	cough			quiet breathing	
	$\tau_{flow}$ [L/s]	$\tau_{dflow}$ [L/s <sup>2</sup> ]	$T_{p,c}$ [s]	$\tau_{xcorr}$	$T_{p,q}$ [s]
<b>S1,S2,S4</b>	1.2	0.3	1.0	0.7	1.5
<b>S3</b>	0.4	0.1	1.0	0.5	1.0

Table 2: Stimulation algorithm parameters. The same parameters could be used for subject S1, S2 and S4, while different values were applied for subject S3.

### 2.3 Experimental protocol and outcome measures

Each subject participated in three experimental sessions on separate days, each lasting approximately one hour. During the first session, the subject was familiarised with the experimental procedures and individual experimental parameters were adjusted. Data were recorded during the subsequent two sessions.

Two tests were performed to evaluate the efficacy of abdominal stimulation: (i) During quiet breathing the subject was asked to breathe normally, and (ii) during coughing, the subject was instructed to make maximal repeated coughing efforts.

The stimulation was triggered automatically based on measurements of the respiratory activity, using the control algorithm described above. During each test, the experimenter used a button switch to alternate periods of unassisted breathing with periods of breathing assisted by stimulation. Each experimental test therefore resulted in alternating measurements for unassisted and assisted respiration.

Although the subjects had no abdominal cutaneous sensation they were aware of abdominal contractions resulting from the stimulation. It was therefore not possible to blind them to the

stimulation.

Airflow and respiratory volumes at the mouth and nose were measured directly using the spirometer and mask. Expiratory tidal volume was recorded for quiet breathing, and cough peak flow was recorded during the cough tests. From the airflow data the beginning of expiration and inspiration were determined. This information was used to derive the number of breaths for each experimental situation and to derive respiratory rate (RR) and minute ventilation  $\dot{V}$  during quiet breathing.

### 2.3.1 End-tidal partial pressure of CO<sub>2</sub> in expired air (EtCO<sub>2</sub>)

EtCO<sub>2</sub> was measured before and after a test of continuous quiet breathing assisted by stimulation. The sample line of the end-tidal CO<sub>2</sub> monitor (Datex Normocap, Datex Instrumentarium, Helsinki, Finland) was connected between the face mask and the spirometer.

## 2.4 Data and Statistical Analysis

For each subject stimulated breaths are compared with unassisted breaths. The mean and standard deviations (SD) for assisted and unassisted breaths over all tests, together with the changes are calculated. A paired T-test was performed to compare assisted and unassisted breaths for the same subject and the resulting p-values are reported. No inter-subject comparison was carried out due to the high variation in the respiratory capacity between subjects.

## 3 Results

During the experiments all subjects were in the normal sitting position in their wheelchairs, except subject S3 who was unable to sit upright and underwent stimulation while reclined. All subjects tolerated the stimulation without discomfort.

The automatic system was able to correctly differentiate attempts to cough and quiet breathing situations in all subjects throughout the experiments, without inappropriate stimulation which would interfere with their voluntary breathing activity.

The results of measurements for expiratory tidal volume ( $V_T$ ), respiratory rate (RR) and minute ventilation ( $\dot{V}$ ) for quiet breathing with stimulation (*assisted*) and without (*unassisted*) are shown in figure 2 (means and SDs). The number of breaths together with the changes of these measurements for each subject are reported in table 3.

Subject	S1		S2		S3		S4	
	unassisted	assisted	unassisted	assisted	unassisted	assisted	unassisted	assisted
No. of breaths	79	120	188	190	228	352	89	132
$\Delta V_T$ [L]		+0.14		+0.21		+0.05		+0.23
$\Delta RR$ [min <sup>-1</sup> ]		-1.0		-8.0		-1.0		-0.9
$\Delta \dot{V}$ [L/min]		+1.05		+1.38		+0.68		+2.07

Table 3: Results of quiet breathing. The number of breaths and the changes in tidal volume ( $\Delta V_T$ ), respiratory rate ( $\Delta RR$ ) and minute volume ( $\Delta \dot{V}$ ) are shown.  $P < 0.05$  for all results.

For coughing, the measurements of the cough peak flow (CPF) with stimulation (*assisted*) and without (*unassisted*) are shown in figure 3. The corresponding number of coughs, the changes of CPF and the p-values are summarised in table 4.

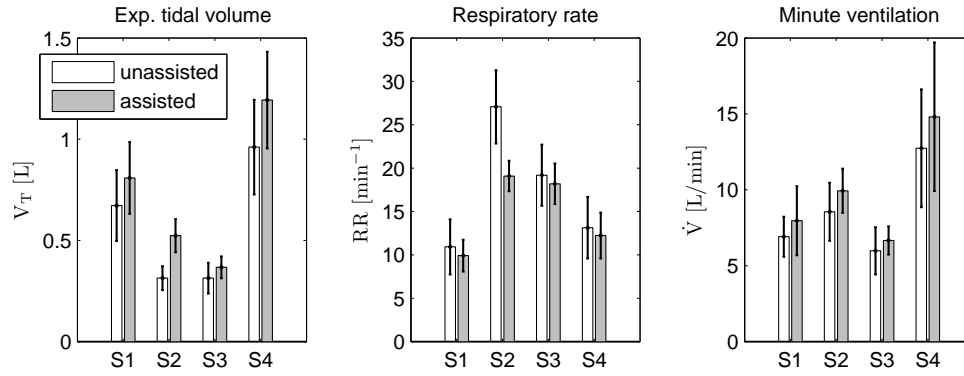


Figure 2: Results of quiet breathing. The bars indicate the mean values for unassisted (white) and assisted (grey) breathing for each subject. The error bars show the corresponding standard deviations.

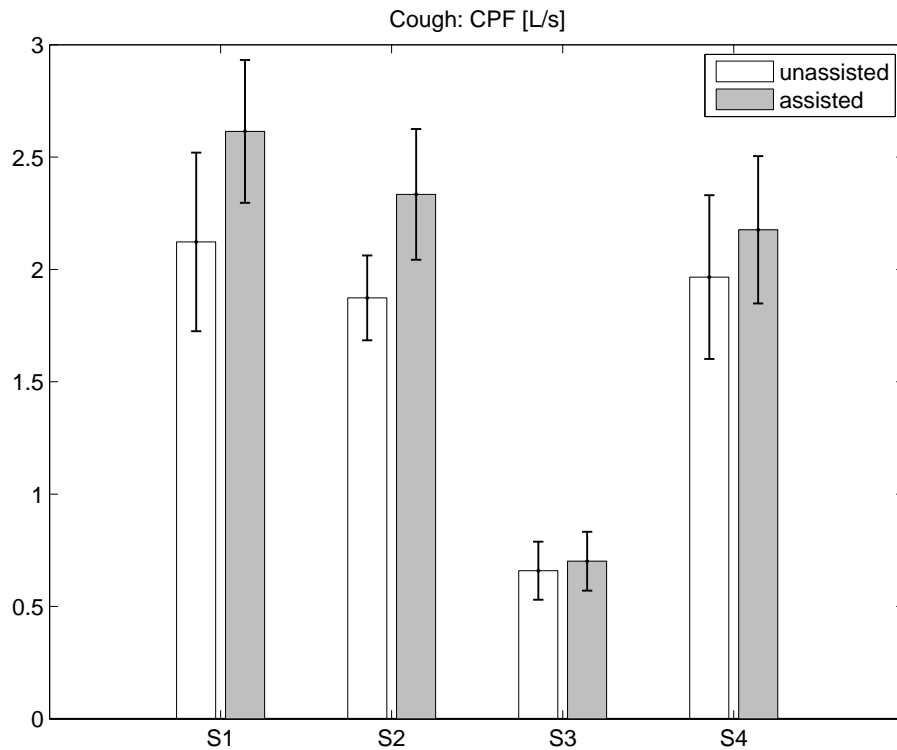


Figure 3: Results for cough. The bars indicate the mean values for unassisted (white) and assisted (grey) breathing for each subject. The error bars show the corresponding standard deviations.



Subject	S1		S2		S3		S4	
	unassisted	assisted	unassisted	assisted	unassisted	assisted	unassisted	assisted
No. of coughs	49	54	43	69	41	80	54	78
$\Delta$ CPF[L/s]	+0.49		+0.46		+0.04		+0.21	
p-value	< 0.05		< 0.05		0.09		< 0.05	

Table 4: Results of coughing. The number of breaths and the changes in peak flow ( $\Delta$ CPF) are shown together with the corresponding p-values.

Results for EtCO<sub>2</sub> are summarised in table 5, reporting values obtained before (baseline) and after (post stim) assisted quiet breathing for five minutes. For S2 we could only measure EtCO<sub>2</sub> over a period of two minutes.

Subject	S1		S2		S3		S4	
	baseline	post stim	baseline	post stim	baseline	post stim	baseline	post stim
EtCO <sub>2</sub> [kPa]	5.9	6.0	5.3	4.5	6.1	6.4	5.1	4.4
$\Delta$ EtCO <sub>2</sub>	+0.1kPa (+1.7%)		-0.8kPa (-15%)		+0.3kPa (+4.9%)		-0.7kPa (-14%)	

Table 5: EtCO<sub>2</sub> before (*baseline*) and after (*post stim*) five minutes of assisted quiet breathing (two minutes for S2).

## 4 Discussion

Previous authors [15,20,21] describe surface FES systems which were successful in improving either cough or tidal volume. Other FES techniques to improve cough in tetraplegia include magnetic stimulation over the lower thoracic spine [14] and direct spinal cord stimulation [4]. In these systems the timing of the stimulation was controlled manually: stimulation was triggered by the experimenter or subject and required activation whenever a cough or breath was needed. While it is possible to use such a system for the occasional cough, it is very difficult to trigger the stimulation at a normal respiratory rate for longer than a few minutes. Manual triggering at this frequency would be impractical for prolonged clinical applications.

In this study an automatic system was used: an input signal (airflow) and customised software, with the stimulator hardware, stimulated continuously without requiring further operator input. This enabled the user to apply electrical stimulation continuously over longer periods of time (up to five minutes). The automatic triggering ensured that the stimulation was delivered consistently while synchronised with the subject’s voluntary breathing. The subject could speak without interference from inappropriate stimulation, and benefited from augmentation of ordinary quiet breathing when mute. The subject could also trigger a cough by taking a large breath; the system sensed the need for a cough and automatically provided the necessary stimulation. The study shows that the system operates well in tetraplegic individuals with different respiratory capabilities.

### 4.1 Quiet breathing

The system produced a significant increase in tidal volume in all subjects during quiet breathing. For one subject (S3) the increase was very small (50mls per breath) and the clinical significance of this is doubtful. Other subjects demonstrated a mean gain of 140-230ml per breath. The difference

between the subjects' responses to stimulation is likely to be multifactorial. S3 could not be examined sitting up and this may have diminished the effects of the stimulation. Previous authors have demonstrated abdominal muscle atrophy in tetraplegic patients [7] and we suggest that different degrees of atrophy may account, at least partly, for differences in response. It may be that S3, as the longest injured subject, had the most abdominal muscle atrophy. In addition, S3 is the only female subject which may contribute to a lower baseline respiratory performance.

Respiratory rates fell in all four subjects during stimulation but the minute ventilation rose due to the increase in  $V_T$  discussed above. These changes were significant in all subjects. The subjects could not be blinded to stimulation and, although they were encouraged to breathe normally and quietly, there may have been an inadvertent voluntary control of rate and volume.

## 4.2 Cough

The system significantly improved CPF in three subjects. S3 again had a poorer response, perhaps for reasons discussed above. Differences in cough between subjects may be partly explained by the mechanism of active expiration in tetraplegia using the clavicular portion of the pectoralis muscles [8]. Forced expiration and cough are therefore not solely dependent on abdominal muscle action and it is possible that differences in shoulder girdle musculature and level of paralysis may account for the variable responses to stimulation. These activities are also effort dependent which could have influenced their repeatability.

## 4.3 End-tidal partial pressure of CO<sub>2</sub> in expired air

Given the significant increases in minute ventilation in all the subjects we might have expected to see reductions of EtCO<sub>2</sub> in all four. However, only two subjects demonstrated slight reductions in EtCO<sub>2</sub> after stimulation. The abnormal muscle fibre recruitment pattern elicited during FES is recognised to lead to increased muscle fatigue [12] and we suggest that the higher metabolic cost of FES may have offset any small potential improvements in gas exchange.

## 4.4 Further applications

In the current setup, a spirometer attached to a facemask was used to measure airflow at the mouth as the trigger signal for stimulation. The mask was well tolerated in testing sessions of up to one hour and we suggest that, in an acute setting, it could be used for longer with the same level of patient inconvenience as a non-invasive mask ventilation system. While the mask allowed the subjects to speak during the experiments, the system could be improved by less intrusive measurement of the respiratory cycle. We are therefore exploring alternative sensors to detect movement of the abdominal wall and the chest and provide a suitable signal which represents respiratory activity [10]. A further aspect which needs to be considered if the system is to be used for longer periods of time is muscle fatigue. While fatigue can to some extent be compensated for by increasing the stimulation intensity, it might become a limiting factor for the duration for which a sufficient respiratory improvement can be achieved by stimulation. On the other side, the use of abdominal FES can reduce muscle atrophy and might have a training effect. We did not look for evidence of a training effect in the stimulated muscle groups, but FES training of other skeletal muscle groups in both the able-bodied [13] and paralysed [1, 17, 18] is known to improve muscle bulk and power and we suggest that the abdominal wall muscles may have the same potential for training to increase the efficiency of the FES.

The current system is based on surface stimulation which has the advantage of being non-invasive, but requires applying electrodes before the system can be used. In a chronic setting, the

automatic stimulation system could be combined with implanted electrodes which either stimulate the corresponding motor neurons directly [2], or the relevant nerve roots at the spinal cord [4].

The subjects tested had established tetraplegia but there may also be applications to augment cough and tidal volume in newly injured spinal patients for whom non-invasive ventilation is being considered. The study shows that substantial variations in the effectiveness of abdominal FES could be observed between tetraplegic subject with different neurological, anatomical and respiratory capabilities. While it is known that posture affects respiration, it would be worth further investigation in how far augmentation achieved by abdominal stimulation varies between the sitting and supine position. This would be in particularly interesting, since subject S3, who benefited to a limited extend from the abdominal support, was the only subject who was tested in a supine position.

## 5 Conclusions

It was shown that continuous automatic abdominal surface FES is feasible with the system presented here and was well tolerated by patients, leading to improved tidal volume and cough in tetraplegic subjects over several minutes without external interference. Results vary greatly between subjects, with injury level and time post injury potential factors which might influence the efficiency of this method. Further studies are therefore needed to identify patient groups within the tetraplegic population with the greatest potential to benefit from this form of intervention, together with the development of optimal intervention protocols. There may be scope for improving response by FES training of the abdominal muscles which could supplement conventional respiratory therapy.

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