

Breathing pattern detection and classification for automatic abdominal functional electrical stimulation

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Introduction

Abdominal Functional Electrical Stimulation (AFES) has been shown to improve the respiratory function of the tetraplegic population [1]. For AFES to be successful, a cough and a quiet breath require different stimulation parameters. Therefore a standalone automatic AFES system should be capable of classifying a breath type in real time so that the correct stimulation parameters can be applied. As AFES is applied during exhalation, classification should be made by the end of inhalation to allow the correct stimulation to be applied.

The air flow rate measured using a spirometer has previously been used for breathing pattern detection [2,3]. However a spirometer is usually coupled with a face mask which is intrusive, leaving a user unable to speak, eat or drink. Replacing the spirometer with a non-intrusive sensor would make the system considerably more practical. One potential option is piezoelectric belt sensors which have provisionally been shown to be capable of breathing pattern detection [4].

A statistical classifier was previously developed for breathing pattern classification using a spirometer [5]. In this new study the classifier was modified to use the signal from a piezoelectric belt. The belt signals, along with the signal from a spirometer for reference, were analysed to establish whether piezoelectric belts could be used as a non intrusive method of real time breathing pattern detection and classification.

Aims

The aims of this study were twofold:

- 1) To identify whether piezoelectric belts are suitable for real time breathing pattern detection.
- 2) To develop a statistical classifier capable of breathing pattern classification using the signal from piezoelectric belts.

Methods

This study was approved by the Faculty of Biomedical and Life Sciences ethics committee, University of Glasgow. Ten able bodied subjects were recruited and asked to attend 2 sessions. In each session the subject was asked to breathe in different ways, with and without AFES. Breathing was recorded using a spirometer (Microloop,

Micromedical, Chatham, UK), connected to a full face mask, and two piezoelectric belts (ProTech, WA, USA) placed around the subject's abdomen and chest.

As the spirometer signal provides a direct measurement of air flow, the positive zero crossing of the spirometer signal gives the time point at which exhalation starts and where stimulation should be applied. Therefore to establish whether the belts were suitable sensors to trigger stimulation, the timing of the positive gradient zero crossings (Marked on Figure 1) were compared to the spirometer zero crossings.

From the inhalation of each breath 25 features were identified from both the frequency and time domain. Example features are shown in Figure 1.

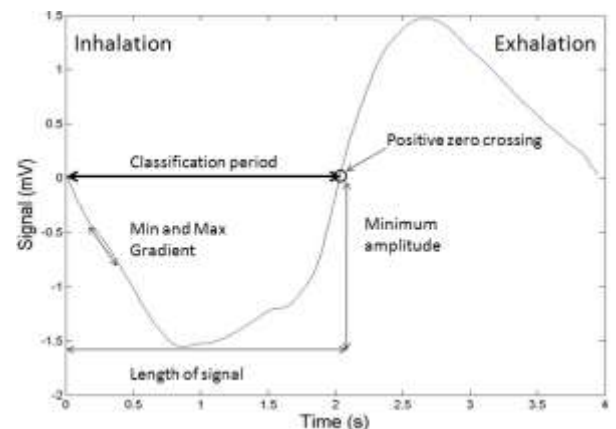


Figure 1. Example of features extracted during the inhalation from the piezoelectric belt signal.

A student t-test was used to determine which of the 25 features were statistically significantly different ($p < 0.05$) between a quiet breath and cough. The statistically different features were used to construct a statistical classifier based on a maximum likelihood classification method. The classification structure is illustrated in Figure 2.

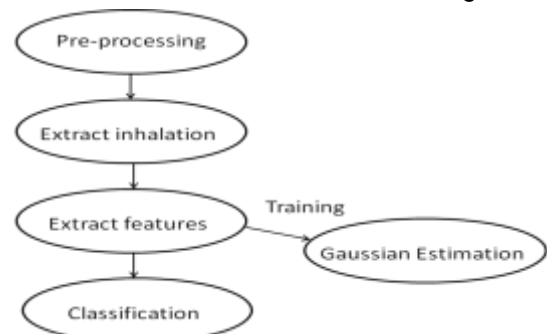


Figure 2. Classification structure.

The statistical classifier was trained for each subject using approximately 40 coughs and 40 quiet breaths. Following this, the trained classifier was tested using the remaining breathing data (approximately 80 coughs and 180 quiet breaths). This procedure was repeated for all three sensors. Each breath could be classified as either a quiet breath or cough and classification was performed before the end of inhalation as shown in Figure 1.

For evaluation, the classification of each breath made by the statistical classifier was compared to the classification automatically determined by the peak expiratory air flow rate of each breath measured by the spirometer. Note the exhalation data would not be available in a real time system.

For each breath type (cough or quiet breathe), the classifier was evaluated using two measurements. Classification accuracy was defined as the percentage of breaths which were correctly classified out of the total number of breaths of that type and error rate was defined as the percentage of breaths which were incorrectly classified as that type out of the total number of breaths of that type. To determine whether a piezoelectric belt could be used instead of a spirometer for breathing pattern classification the accuracies and error rates achieved using the belt signals were compared to the accuracies and error rates achieved using the spirometer signal. To determine the most suitable position of the belt, the accuracies and error rates achieved using the signal from the two belts were also compared.

Results

The time when the end of inhalation was detected by the belts compared to the spirometer is shown in Table 1. A positive delay means that the signal occurred after the spirometer and vice versa.

Table 1. Comparison of time delay between piezoelectric belts and spirometer.

Sensor	Mean delay +/- std (s)
Abdominal belt	-0.01 +/- 0.12
Chest Belt	0.04 +/- 0.12

Table 2. Mean percentage classification accuracy (A) and error rate (ER) (+/- standard deviation) of statistical classifier for quiet breaths (QB) and coughs (C) for different sensors.

Sensor	A _{QB} (%)	A _C (%)	ER _{QB} (%)	ER _C (%)
Spiro'	97.4 +/- 2.2	96.0 +/- 5.3	1.7 +/- 2.3	6.1 +/- 5.0
Chest belt	94.4 +/- 5.2	97.0 +/- 4.8	1.3 +/- 1.9	10.7 +/- 8.5
Abdo belt	89.7 +/- 8.4	83.9 +/- 12.2	6.5 +/- 5.0	21.5 +/- 15.2

The classification accuracies and error rates of the classifier are shown in Table 2, with the results which could be achieved when using the spirometer signal shown for comparison.

Discussion

The signal from the piezoelectric belt positioned around the abdomen detected the end of inhalation 10ms before the signal from the spirometer. The signal from the belt placed around the chest detected the end of inhalation 40ms after the spirometer. However as the differences are small the signals from both belts are suitable for triggering stimulation at the start of exhalation.

Previously the signal from a spirometer had been used for breathing pattern detection in an AFES system as this signal is a direct measure of airflow [1,3]. As can be seen from the results the signal from a piezoelectric belt placed around the abdomen or chest is also suitable for real time breathing pattern detection and classification. Of the two belts the signal from the belt placed around the chest gave the best combination of high accuracy and low error rate, with values which were comparable to those achieved with the spirometer signal. The signal from the abdominal belt did not give as high accuracy.

The accuracy and error rates shown may be improved by using a larger amount of training data or by combining the signal from the two belts. In the future the classifier will also be developed to be capable of not applying stimulation in unusual situations such as speaking.

The results of this study suggest that a piezoelectric belt placed around the chest is the most suitable non intrusive sensor for use with a statistical classifier for real time breathing pattern detection and classification. The belt and classifier combination will be used to develop a standalone AFES system to help improve the respiratory function of tetraplegics.

References

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