Diaphragm electromyography using an oesophageal catheter: current concepts

Yuan Ming LUO*, John MOXHAM† and Michael I. POLKEY‡
*Guangzhou Medical College, National Key Laboratory of Respiratory Disease, Guangzhou 510210, People’s Republic of China, †Department of Respiratory Medicine, King’s College Hospital, London SE5 9PJ, U.K., and ‡Respiratory Muscle Laboratory, Royal Brompton Hospital, London SW3 6NP, U.K.

ABSTRACT

The usefulness of diaphragm electromyography recorded from an oesophageal electrode depends on a reliable signal which is free of artefact. The diaphragm EMG (electromyogram) recorded from chest wall surface electrodes may be unreliable because of signal contamination from muscle activity other than the diaphragm. Initially, the oesophageal electrode catheter for human studies had only one electrode pair, which could be difficult to position accurately and was influenced by a change in lung volume. Recently, a multipair oesophageal electrode has been developed which allows a high-quality EMG to be recorded. In the present review, the progress of oesophageal electrode design is outlined. The effects of signal contamination, electrode movement and particularly the effect of change in lung volume on the diaphragm EMG are discussed. The diaphragm EMG, recorded from a multipair oesophageal electrode, is useful to assess neural respiratory drive and diaphragm function in different groups of patients with respiratory disease, including patients with neuromuscular disease and sleep-disordered breathing, and those in the intensive care unit. When combined with cervical and cranial magnetic stimulation, an oesophageal electrode can be used to partition the central respiratory response time and phrenic nerve conduction time.

INTRODUCTION

The diaphragm is the most important respiratory muscle, accounting for 70% of respiration in normal humans [1]. Diaphragm weakness is associated with breathlessness and respiratory failure [2–6]. Electrical assessment of diaphragm and phrenic nerve function can provide important information on the diagnosis and management of patients with neuromuscular disease [7–13]. By recording the diaphragm CMAP (compound muscle action potential) and MEPs (motor-evoked potentials), respiratory neural pathways can also be assessed [14–19]. The diaphragm EMG (electromyogram) is also useful in detecting diaphragm fatigue [20–25] and assessing neural respiratory drive [26–34], as well as to trigger [33] and adjust [34] mechanical ventilation. Measurement of the electrical activity offers different and complimentary information to that provided by the measurement of oesophageal pressure and $P_d$ (transdiaphragmatic pressure). However, the usefulness of the diaphragm EMG is dependent on the accurate recording of the signal without artefact. Three methods can be used in humans to record a diaphragm EMG: needles, surface electrodes and an oesophageal catheter. Although needle electrodes have been used in humans in small physiological studies [35,36], they are impractical for most clinical studies, particularly if there are factors which increase the risk of and from a pneumothorax.

Chest wall surface electrodes provide a non-invasive technique and have been used to record the diaphragm

Key words: diaphragm, diaphragm function, electromyography, neural respiratory drive, oesophageal electrode, respiratory muscle.

Abbreviations: CMAP, compound muscle action potential; COPD, chronic obstructive pulmonary disease; EMG, electromyogram; Fc, centroid frequencies; H/L ratio, ratio of high-frequency power to low-frequency power; ICU, intensive care unit; MEP, motor-evoked potential; OSA, obstructive sleep apnoea; $P_d$, transdiaphragmatic pressure; RMS, root mean square.

Correspondence: Professor Michael I. Polkey (email m.polkey@rbh.nthames.nhs.uk).
EMG, but this approach has disadvantages. First, the diaphragm EMG recorded from the chest wall surface electrodes can be unreliable because of signal contamination [37–40]. It has been reported that the diaphragm CMAP recorded from chest wall surface electrodes is likely to be contaminated by signals from adjacent muscles when the brachial plexus is co-activated, particularly likely with magnetic stimulation [37–39] (Figure 1). Similarly, the power spectrum and RMS (root mean square) activity derived from the spontaneous diaphragm EMG recorded from chest wall surface electrodes can be unreliable because of cross-talk signals from intercostal and abdominal muscles, which are also active during loaded breathing [40]. Secondly, the diaphragm EMG recorded from chest wall surface electrodes can be affected by subcutaneous fat, which significantly reduces signal strength due to muscle-to-electrode filtering effects [41]. Thirdly, it has been reported that the surface diaphragm CMAP is subject to power line artefacts [42]. Finally, although several possible sites have been reported [13, 17, 18, 37–39, 43, 44], there is no standardized method for placing the chest wall electrodes, which makes data comparison between subjects and studies difficult. In contrast, the diaphragm EMG recorded from an oesophageal electrode is less affected by obesity, power line artefact and cross-talk signals [37–41]. Consequently, oesophageal diaphragm electromyography has become an increasingly important clinical and research technique [7, 10, 11, 18, 20–28, 30–34, 37–41, 44]. Because assessment of diaphragm function, including electromechanical efficiency, is best done by recording both $P_d$ (gastric pressure–oesophageal pressure) and the diaphragm EMG, a single EMG electrode catheter mounted with two balloons for measurement of oesophageal and gastric pressure [45–47] has additional advantages over chest wall surface electrodes.

Historically, the diaphragm was considered as one muscle [48]; however, studies have suggested that the diaphragm consists of two separate muscles, the costal and crural diaphragm [49]. This distinction is important because the oesophageal EMG measures signals from the crural component, whereas surface electrodes measure signals from the costal component. Some studies have found a small difference between the crural and costal diaphragm. An animal study [50] showed that there was more post-inspiratory activity in the costal than the crural muscle, but the distinction was most apparent during anaesthesia. It has also been reported, in cats, that the crural EMG is more sensitive to CO$_2$ stimulation and to changes brought about by posture than the costal EMG [51]. However, the close correlation of the electrical activity in the crural and the costal parts has been well documented in humans during respiratory tasks [52, 53]. The EMG recorded from the costal diaphragm was similar to that from the crus in terms of synchronous activation and time course in both quiet and loaded breathing [48, 51, 54]. Lastly, in a series of studies, Sharshar et al. [18] showed that the effect of various respiratory tasks on the diaphragm MEP was the same whether measured from oesophageal or surface electrodes. It should be noted, however, that other tasks relating to specific functions (e.g. postural or gastro-oesophageal sphincter control) may well be associated with differential activation of different portions of the diaphragm. The diaphragm EMG recorded by an oesophageal electrode is mainly from the crural part [55–58].

**OESOPHAGEAL ELECTRODE DESIGN**

Oesophageal electrodes have been used for more than 40 years since they were introduced for recording human diaphragm EMGs in the 1960s [55, 59]. An oesophageal electrode consists of an oesophageal catheter with metal coils at the distal end. Initially, oesophageal electrode catheters contained only one electrode pair [55, 59]. Later, a modified oesophageal electrode with three pairs of electrodes was used [56]. Multipair electrodes with more than three pairs have been widely used [7, 10, 11, 18, 27, 28, 30–34, 37–39, 45, 60, 61]. Different metals have been used to construct oesophageal electrodes, including stainless steel [60, 61], copper [23, 25–28], silver [55, 58, 62] and platinum [46, 63].
Interelectrode distance is important for the accuracy of recording the diaphragm EMG. Because the amplitude of the diaphragm CMAP elicited by phrenic nerve stimulation, and the magnitude of the diaphragm EMG during spontaneous breathing, relates to interelectrode distance [58] (Figure 2), to ensure data are comparable a constant interelectrode distance should be used. However, different interelectrode distances have been used since oesophageal recording was introduced. Initially, a 1 cm interelectrode distance was usually used [46,53,55,59,63], but other electrode configurations have also been employed; for example Gross et al. [64] used a 1.8 cm interelectrode distance. Some investigators have used a <1 cm interelectrode distance [23,25,60,61]. McKenzie and Gandevia [58] found that the amplitude of CMAP increased with increasing interelectrode distance and they therefore used a 5 cm interelectrode distance [58,65]. There is no universal agreement about the optimal interelectrode distance, although an interelectrode distance of 1 cm was usually used to record interference pattern EMGs by Sinderby’s group [32,34]. We have found that a good quality diaphragm EMG or CMAP can also be recorded using an interelectrode distance of 3 cm within a pair [7,10,11,27,28,38].

A power line artefact is one of the most common problems with recording the diaphragm EMG; it originates from the power line and mains power equipment and will typically have a 50 or 60 Hz signal depending on the power source. Proper earth connections and using a differential amplifier with a high common mode rejection ratio (e.g. >100 dB) are useful to reduce the artefact [7,66]. An isolated pre-amplifier with ‘notch filters’ and shielded electrode cable can minimize the artefact further.

Another source of artefact is electrode motion, which usually occurs during voluntary contractions, but it is relatively easy to deal with by using a high-pass-filter setting at 20 Hz [25,66]. The diaphragm CMAP produced by single twitch is usually free of motion artefact, as electrical activity occurs prior to any mechanical response [39]. Oesophageal peristalsis produces an additional low-frequency large amplitude deflection of the baseline, but does not cause major problems if a high-pass-filter setting at 10 or 20 Hz is used [27,28,66].

**ECG ARTIFACT**

The diaphragm CMAP recorded from the oesophageal electrode can be contaminated by the superimposed ECG. The diaphragm EMG frequency is from 20–250 Hz, whereas most of the ECG frequency is 0–100 Hz [25]. It is therefore difficult to eliminate the ECG effectively using filters because of the overlap in the range of the two signals, although the P and T waves are relatively easy to delete by setting a high-pass filter at 20 Hz [20,25]. Wiener filtering could be useful to reduce the ECG artefact, but it is usually unable to completely separate the ECG from the EMG [67]. Many methods have been used to reduce or eliminate the ECG artefact. Gating techniques are often used online or offline, which delete both the EMG and QRS complex simultaneously for approx. 0.4 s around the QRS complex [25,68]. Some investigators have attempted to zero the EMG signal for the duration of QRS complex and to replace it with adjacent EMG activity [67] (Figure 3). A similar method, which we have favoured, is to analyse the EMG between QRS complexes [27,28,64], for example selected data between 50–75 % of the QRS complex interval [67], or to trigger sampling of the EMG with the ECG after a time delay [20]. Because the diaphragm EMG amplitude can be larger than the ECG during maximal breathing, it is not always reliable to trigger sampling with the ECG when using an oesophageal electrode, so surface signals are preferred for this purpose [20]. An additional channel for recording the ECG is sometimes necessary to act as a trigger source [20]. A more complex method is subtraction of the ECG template from the EMG signal in the
section contaminated by the ECG waveform [68]. Several procedures are employed, including generating an ECG template by averaging the number of QRS complexes during expiration, and alignment of an ECG template to the ECG complexes overlapping the EMG and subtracting the ECG template from the EMG contaminated by ECG. These methods are considered to be more complicated than gating but also more accurate [68]. Fortunately, the effect of the ECG on the diaphragm CMAP elicited by phrenic nerve stimulation can be eliminated easily by simply discarding those CMAPs contaminated by the ECG, because the baseline and shape of the CMAP overlapped by the ECG will change [7,37–39,69].

**EFFECT OF CHANGES IN LUNG VOLUME ON THE DIAPHRAGM EMG**

Concerns about changes in lung volume on the diaphragm EMG have hampered the application of the oesophageal diaphragm EMG in clinical practice. Many studies [53,60,62,65,70,71] have tried to address the effect of lung volume on the amplitude of the diaphragm EMG by using different electrode design. It was hypothesized that a change in the amplitude of the diaphragm CMAP was due to alternations in the distance between the electrode and the diaphragm secondary to the change in lung volume during respiration [65]. On this basis, a balloon-anchored oesophageal electrode catheter was considered, at this time, to be useful to more reliably record the diaphragm CMAP. However, Smith and Bellemare [62] found that diaphragm CMAP changed with the movement of the electrode during respiration, despite using a balloon-anchored oesophageal catheter. To retain the anatomical relationship between the electrode and diaphragm during respiration, Gandevia and McKenzie [65] added an additional weight at the proximal part of the catheter, which itself had a ‘stabilized’ balloon at the distal end, to let the catheter be slightly pulled out by the weight during expiration and pulled in by the pushing of the balloon by the diaphragm during inspiration. However, they found the diaphragm CMAP amplitude (peak to peak) at TLC (total lung capacity) was 3.4 times larger than at RV (residual volume), indicating either that their technique could not effectively maintain electrode position or that the distance between the electrode and the diaphragm was not the cause of the variance. They also observed that during deep breathing the electrode catheter could move up to 8 cm, which is beyond the maximal movement of the crus [61,70]. The large movement of the catheter during respiration suggested that the catheter with both a balloon and the weight was actually unable to reliably record the diaphragm EMG because the electrodes failed to track the movement of the crus of the diaphragm [70,71]. Previously, we have used a multipair oesophageal electrode with an interelectrode distance of 3 cm within a pair and found that changes in lung volume of up to 2 litres had little effect on the amplitude of the diaphragm CMAP, providing the oesophageal electrode catheter was optimally positioned and then fixed on the nose [71]. However, Beck et al. [70] reported that the amplitude of the diaphragm CMAP changes with lung volume when using an electrode with an interelectrode distance of 1.0 cm within a pair. Nevertheless, it now seems clear that the oesophageal electrode with a stabilized balloon does not record the diaphragm EMG reliably [27,28,56,61,70]. To overcome the influence of lung volume on the diaphragm EMG, multipair oesophageal electrodes have been used to record the diaphragm EMG. Daubenspeck et al. [60] found that the diaphragm EMG could be reliably recorded by summation of the EMG recorded from seven consecutive pairs of electrode in spite of a change in lung volume. It has been suggested that the orientation of the crural diaphragm fibres with respect to the axis of the oesophagus is perpendicular [41]. Therefore the electrode pairs caudal and cephalad to the central area of the crus would have a similar amplitude and opposite polarity if both electrode pairs were the same distance from the centre [7,38,39,71]. Sinderby and co-workers [67] have developed a cross-correlation method to determine the electrode pair closest to the diaphragm. A catheter with seven [72] or eight [67] consecutive pairs of electrodes with a 1 cm interelectrode distance within a pair was used to track the position of the diaphragm’s electrically active centre. A correlation coefficient was calculated for every other pair and the difference between the signals that had the best correlation was obtained (the double-subtracted signal) [67,72] to enhance the signal-to-noise ratio and to reduce the artefact originating from diaphragm movement. With this technique, they also found that the diaphragm EMG was independent of changes in lung volume; however, it should be pointed out...
that the double-subtracted signal was from the electrode pairs slightly away from the diaphragm, rather than the pair at the electrically active centre, although the signal obtained from the pair exactly at the electrical activity centre was sometimes included for analysis [45].

**POSITIONING OF ELECTRODES**

To accurately record the diaphragm EMG, the oesophageal electrode should be positioned as close to the diaphragm as possible. Unlike chest wall surface electrodes, oesophageal electrodes cannot be positioned simply using fixed anatomical landmarks. Traditionally, investigators [23] positioned the oesophageal electrode by observing the magnitude of the diaphragm EMG during sniffs or during maximal or quiet breathing efforts on the assumption that the maximal EMG signal would be obtained when the catheter was optimally positioned [23,55,73]. Luo et al. [7,38,71] developed a technique to accurately position an electrode based on the amplitude and polarity of the diaphragm CMAP elicited by bilateral magnetic phrenic nerve stimulation [7,71]. We used a catheter with seven metal coils (number 1 being proximal and 7 being distal) created with four electrode pairs (Figure 4), the interelectrode distance within a pair being 3 cm. Electrode 3 was positioned close to the diaphragm based on the amplitude and opposite polarity of the diaphragm CMAP recorded from pairs 1 and 3, since the CMAP shape is sensitive to electrode position [61,74]. The position was adjusted further by observing the amplitude and polarity of the CMAP recorded from electrode pair 2, which was characterized by a small or absent CMAP, since there is a cancellation of the potential when both recording electrodes are equally close to the source of the diaphragm potential [65,71,72]. The ideal position of the electrode was characterized by a large CMAP amplitude from pairs 1 and 3, with opposite polarity, and a small CMAP was recorded from electrode pair 2 (Figure 4). This method can position the electrode accurately, but it requires the use of phrenic nerve stimulation which is a complex technique. Luo et al. [27,28] have demonstrated the positioning of a multipair oesophageal electrode based on a spontaneous EMG recorded simultaneously from four or five pairs of electrodes. The optimal position was characterized by a large EMG from two electrode pairs which shared the middle electrode and a small EMG from the pair straddling the middle electrode (Figure 5).

**ANALYSIS OF EMG SIGNALS**

For the diaphragm CMAP, latency and the amplitude are measured [7,14,37–39,44,58]. The latency is synonymous with the conduction time, which is measured from stimulation artefact to the onset of CMAP [7,14,37–39,44,58]. The amplitude of the diaphragm CMAP is greatly affected by the stimulator output. A supramaximal stimulation of the phrenic nerve is usually desirable, although since submaximal stimuli recruit large (and therefore faster) bundles of axons preferentially the measured nerve conduction time does not greatly depend on the stimulus intensity whereas the amplitude does (Figure 6). For the amplitude of the diaphragm CMAP to be useful, only the signals elicited by demonstrably supramaximal (conventionally 20 % higher than maximal) stimulation are usually analysed. To eliminate the influence of the ECG, CMAPs are usually selected from those with a constant shape and stable baseline before and after stimulation [7,38,39,69]. Similar principles are used in the analysis of the MEP [14–19] following stimulation of the diaphragm motor area of the cortex, but it should be noted that, because the MEP reflects a volley of impulses, deciding where the signal begins (and hence the latency) may be more difficult (Figure 7).

**TIME AND FREQUENCY DOMAIN ANALYSIS OF THE DIAPHRAGM [75]**

**Time domain EMG analysis**

Time domain EMG analysis is used to describe the relationship between EMG strength and time so that, broadly speaking, activity is the product of the amplitude of the signal and its duration. To facilitate computerized processing, the signal is usually rectified and integrated [46,63,76,77]; however, some groups prefer to use a
moving average, peak activity or rate of rise of activity [26]. The RMS of the diaphragm EMG has been used (Figure 8) to quantify the activity of the diaphragm [27,28,31,34,78] because it reflects the number and firing rate of the motor units recruited. It has been demonstrated that the RMS of the diaphragm EMG is independent of a change in lung volume and is a reliable index of global diaphragm activation [45,67,70].

**Frequency domain analysis**

Power spectral analysis of the EMG has been used to assess diaphragm fatigue [20,21,24,25,73,79], although enthusiasm for its use has diminished following the demonstration that EMG indices of fatigue correlate poorly with mechanical indices of fatigue [80] (Figure 9). In brief, this approach seeks to describe the relationship between the frequency and power of the diaphragm EMG. Fc (centroid frequencies) and the H/L ratio (ratio of high-frequency power to low-frequency power) are usually measured [25]. Fc reflect the relative contribution of low- and high-frequency components in a given bandwidth, and its number represents that frequency at which the power above and below it is the same [23,25]. The H/L ratio is the ratio of power at high frequencies (129–246 Hz) to power at low frequencies (31–51 Hz) [25]. Both Fc and the H/L ratio have been used to identify diaphragm fatigue [20,21,23,25].

**APPLICATIONS**

**Testing the integrity of the phrenic nerve–diaphragm unit**

Accurately recording the diaphragm CMAP elicited by electrical or magnetic stimulation of the phrenic nerve is very useful for assessing diaphragm function and for
selecting patients for diaphragm pacing. Because surface electrodes may be associated with varying conduction times [38,39,81] and are not practical in the obese or those with a chest wall deformity, the oesophageal electrode often offers a better chance of achieving accurate data. Measurement of phrenic nerve conduction time with an oesophageal electrode is useful in differentiating axonal neuropathies (e.g. motor neuron disease) from neuropathies where damage primarily occurs to the axonal myelin [7–9,12,13,69]. Patients with Guillain-Barré syndrome and hereditary motor sensory type 1 usually have a prolongation of phrenic nerve conduction time [8,13]. In contrast, conduction times in patients with motor neuron disease or hereditary motor sensory type 2 are usually normal [13]. The normal range for phrenic nerve conduction time measured by an oesophageal electrode is 7.6 ± 0.7 ms on the left and 6.9 ± 0.9 ms on the right [7]. This reflects the length of the nerve, with the phrenic nerve conduction time on the right side being slightly, but significantly, shorter than that on the left [58].

Oesophageal electrodes can also be used to assess patients who have undergone open chest surgery, including cardiac surgery, to determine whether the phrenic nerve has been injured during the operation [9]. Because the multipair oesophageal electrode can reliably record the diaphragm CMAP elicited by magnetic stimulation, patients with suspected hemi-diaphragm or bilateral diaphragm paralysis can be accurately diagnosed [7,10], although the assessment of the magnitude weakness will usually require the measurement of twitch \( P_{di} \) [7,82].

Assessment of cortical diaphragmatic pathways

Oesophageal electrodes can be used to record the diaphragm EMG response to transcranial electrical [83] or magnetic [14–19] stimulation to investigate neural pathways of the diaphragm. Sharshar et al. [18] found that the diaphragm MEP could be reliably recorded with an oesophageal electrode (Figure 7). By combining transcranial and cervical magnetic stimulation, central conduction time can be estimated by subtraction of the latency of the CMAP elicited by neck stimulation from transcranial stimulation [14–19,84]. On the basis of the MEP, it has been suggested that patients with stroke have impaired central motor conduction to the diaphragm [17]. A possible supplementary diaphragm motor area (3 cm anterior to the vertex) has been identified during transcranial magnetic stimulation mapping with the help of an oesophageal electrode [85]. Motor control of the costal and crural diaphragm can also been assessed with the help of the oesophageal EMG technique [18].

Efficiency of diaphragm contraction and diaphragm fatigue

The EMG recorded from an oesophageal electrode can be used to establish the relationship between EMG amplitude and \( P_{di} \). Similar to the relationship between quadriceps force and EMG amplitude [86], a linear relationship exists between \( P_{di} \) and the diaphragm EMG recorded from needle [87] or oesophageal [88] electrodes.
Figure 8  Recording and analysis of the diaphragm EMG
The diaphragm EMG is recorded with an oesophageal catheter with ten coils. EMG signals are amplified with a gain of 1000 and filtered. RMS of the diaphragm EMG is generated after eliminating the ECG.

Figure 9  Force response to electrical stimulation at 20 Hz/force at 100 Hz following repeated submaximal voluntary contractions in the quadriceps
The Figure illustrates substantial low-frequency fatigue, with the EMG H/L ratio remaining normal. This Figure was reproduced from Moxham, J., Edwards, R.H.T., Aubier, M. et al. (1982) Changes in EMG power spectrum (high/low ratio) with force fatigue in man, Journal of Applied Physiology, 53, 1094–1099, and is used with permission. Copyright American Physiological Society.

It has also been shown that there is a linear relationship between twitch $P_{di}$ and the amplitude of the diaphragm CMAP ($r = 0.8$) [7], which could be clinically important because the diaphragm EMG could be used to predict the strength of the diaphragm, assuming neuromuscular transmission is normal. When $P_{di}$ and the diaphragm EMG are measured simultaneously, contraction efficiency can be assessed and peripheral low-frequency diaphragm fatigue can be detected based on a decrease in the $P_{di}$/integrated diaphragm EMG ratio [89].

The power spectrum of the diaphragm EMG can be recorded from an oesophageal electrode and a shift to lower frequency has been considered to be an indicator that fatigue is occurring [20–25,64,73,79,80]. Fc and the H/L ratio have been used to diagnose diaphragm fatigue in patients, including those with OSA (obstructive sleep apnoea) [24,79], patients undergoing weaning from mechanical ventilation [21] and patients with COPD (chronic obstructive pulmonary disease) during exhaustive exercise [23,32]. From power spectral data, it has been considered that diaphragm fatigue contributes to weaning failure and respiratory symptoms [3,6,90], but findings obtained using magnetic nerve stimulation have established that true low-frequency fatigue is hard to elicit in clinical practice [91–94]. Because the relationship between the power spectral shift and fatigue remains controversial and the underlying cellular mechanisms remain unknown, in particular because power spectral analysis can be affected by the change in lung volume and the distance between the electrode and the diaphragm [70,72], the shift of the power spectrum to lower frequency is of uncertain clinical significance.

Assessment of neural respiratory drive
It is increasingly useful to assess neural respiratory drive, as the product of EMG amplitude against time, in normal subjects and patients with respiratory disease. The diaphragm EMG has long been considered to be a powerful physiological tool to assess neural respiratory drive [26–34,52,56,71]. The diaphragm EMG increases...
Oesophageal diaphragm electromyography

There is a good relationship between end-tidal CO\textsubscript{2} and the amplitude of the RMS. No plateau of EMG is observed. Results are from a patient with COPD. This Figure was reprinted from Respiratory Physiology & Neurobiology, volume 146, Y.M. Luo and J. Moxham, Measurement of neural respiratory drive in patients with COPD, pp. 165–174, Copyright (2005), with permission from Elsevier (http://www.sciencedirect.com/science/journal/15699048).

gradually during CO\textsubscript{2} rebreathing [26–28,46,76], and there is a linear relationship between the RMS of the diaphragm EMG and ventilation in normal subjects [28] and patients with COPD [27] (Figure 10). The observation that pressure support reduces both \(P\textsubscript{di}\) and diaphragm EMG equally in normal subjects [34] suggests that two measures reflect neural respiratory drive. However, inspiratory pressure underestimates neural respiratory drive in patients with COPD because of hyperinflation. For example, Sinderby et al. [32] have shown that diaphragm electrical activity increased progressively during incremental exercise in COPD, whereas \(P\textsubscript{di}\) reached a plateau, indicating that the diaphragm EMG is a better index of neural respiratory drive. Luo and Moxham [27] found neural drive, as assessed by RMS of the diaphragm EMG, increased linearly with an increase in end-tidal CO\textsubscript{2} in patients with COPD. Furthermore, they found that the diaphragm EMG in patients with COPD increased initially but eventually reached a plateau during constant rate treadmill exercise, indicating the importance of the exercise protocol.

**Application in the ICU (intensive care unit)**

An important application of the oesophageal diaphragm EMG is in the assessment of diaphragm function in patients in the ICU [11]. Because patients in the ICU frequently develop a neuromuscular abnormality [12,95,96], including dysfunction of the diaphragm [97], which contribute to weaning failure, measurement of the diaphragm CMAP with an oesophageal electrode can provide useful information about phrenic nerve and diaphragm function [7,8]. Luo et al. [11] measured CMAP with a multipair oesophageal electrode in the ICU and found that the amplitude of the diaphragm CMAP in patients was half that of normal subjects. An attractive potential application of the oesophageal electrode is triggering and adjusting ventilation from the diaphragm EMG [30–34]. Conventionally, ventilators are triggered by flow or pressure, but these methods could fail to detect the onset of breathing causing asynchrony between the patient and the ventilator [33]. Because the diaphragm EMG appears ahead of airflow or a change of pressure, particularly so when there is intrinsic or applied positive end-expiratory pressure, the diaphragm EMG recorded from a multipair oesophageal electrode could be used to trigger the ventilator and improve patient–ventilator synchrony [33]. The diaphragm EMG can also be used to automatically adjust a ventilator in response to varying patient effort [30,33,34]. With this model, ventilatory assistance is given in proportion to the neural drive. Spahija et al. [30] reported that, during loaded breathing in healthy subjects, it was possible to automatically adjust the level of ventilatory assistance from one breath.
to the next and to maintain diaphragm activation within a predetermined target range. Ventilator adjustment using the diaphragm EMG can efficiently unload the respiratory muscles during maximal inspiration [34].

**Applications in sleep medicine**

Vincken et al. [98] used an oesophageal electrode to investigate the mechanism of airflow opening in OSA and concluded that reopening of the airway is not related to respiratory muscle fatigue. Some studies have investigated diaaphragm activity during obstructive events in an attempt to determine whether OSA is associated with diaphragm derecruitment [24,79]. Conventionally, distinguishing obstructive from central apnoea/hypopnoea depends on detection of respiratory effort measured from chest and abdominal movement or oesophageal pressure. The diaphragm EMG can accurately assess neural respiratory drive [27,28,31–33,99], therefore, it could be a useful technique to differentiate types of respiratory events [100,101] (Figure 11); this distinction may be of importance in conditions such as heart failure, where nocturnal ventilatory support is indicated for patients with OSA but not central sleep apnoea.

**CONCLUSIONS**

In conclusion, measuring the diaphragm EMG may be useful for the diagnosis and treatment of respiratory disease. Recent technical advances mean that an oesophageal electrode is simpler to use and, therefore, should now, in most cases, be the technique of choice in preference to surface electrodes.

**ACKNOWLEDGMENTS**

Y.M.L. is supported by a grant from the Natural Science Foundation of Guangdong, People’s Republic of China (No. 6113858) and holds British Council researcher exchange programmes awards.

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