Modulation of chest wall intermuscular coherence: effects of lung volume excursion and transcranial direct current stimulation

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Chest wall muscle recruitment varies as a function of the breathing task performed. However, the cortical control of the chest wall muscles during different breathing tasks is not known. We studied chest wall intermuscular coherence during various task-related lung volume excursions in 10 healthy adults (34 ± 15 yr; 2 men, 8 women) and determined if transcranial direct current stimulation (tDCS) could modulate chest wall intermuscular coherence during these tasks. Simultaneous assessment of regional intercostal and oblique electromyographic activity was measured while participants performed standardized tidal breathing, speech, maximum phonation, and vital capacity tasks. Lung volume and chest wall kinematics were determined using variable inductance plethysmography. We found that chest wall area of intermuscular coherence was greater during tidal and speech breathing compared with phonation and vital capacity (all P < 0.05) and between tidal breathing compared with speech breathing (P < 0.05). Anodal tDCS increased chest wall area of intermuscular coherence from 0.04 ± 0.09 prestimulation to 0.18 ± 0.19 poststimulation for vital capacity (P < 0.05). Sham tDCS and cathodal tDCS had no effect on coherence during lung volume excursions. Chest wall kinematics were not affected by tDCS. Our findings indicate that lung volume excursions about the midrange of vital capacity elicit a greater area of chest wall intermuscular coherence compared with lung volume excursions spanning the entire range of vital capacity in healthy adults. Our findings also demonstrate that brief tDCS may modulate the cortical control of the chest wall muscles in a stimulation- and lung volume excursion task-dependent manner but does not affect chest wall kinematics in healthy adults.

respiratory muscles; chest wall kinematics; EMG-EMG coherence; motor control

ACTIVE EXPIRATION REQUIRES the recruitment of the chest wall muscles, which includes the rib cage and abdomen (Hixon et al. 1976; Hoit et al. 1988). The magnitude of recruitment of respiratory muscles is dependent on the prevailing lung volumes, force of expiration, and purpose of expiration. For example, speech breathing requires slightly greater muscle recruitment, based on electromyographic (EMG) analysis, compared with tonic or passive activity associated with tidal breathing. Whereas, compared with tidal and speech breathing, sustained (maximal) phonation and vital capacity requires breathing across a larger range of lung volumes and subsequently greater chest wall muscle recruitment. During maximal sustained phonation, maintenance of a constant tracheal pressure against a decreasing lung volume requires additional muscular control (Hixon et al. 1976; Hixon and Weismer 1995). The findings of Macelfield and Gandevia (1991) suggested that conscious control of respiratory muscles originates, in part, from the primary motor cortex, but the cortical control of the chest wall muscles during respiratory tasks spanning various lung volumes is not understood.

Coherence analysis of EMG activity between synergistic muscles during a movement task yields the degree of linear dependence between respective signals in the frequency domain. Thus intermuscular coherence is interpreted to reflect, in part, motor unit synchronization between two muscles. Within the approximate 15- to 35-Hz frequency range (β-band), the coupling between motor unit firing is under primary motor cortex control (Farmer et al. 1993; Feurra et al. 2011; Grosse et al. 2002; Kilner et al. 1999). Subsequently, an alteration in motor cortex excitability following a limb muscle task intervention, for example, can be determined by a change in intermuscular coherence (Halliday et al. 2003; Norton and Gorassini 2006; Perez et al. 2012). Given the findings of Macelfield and Gandevia, an alteration in cortical excitability may confer a change in coherence between chest wall muscles. This may provide an important first step in understanding mechanisms associated with chest wall dysfunction and rehabilitation in individuals with neuromuscular deficits that affect voluntary respiratory control. This notion is further supported by previous work showing that intermuscular coherence has some degree of inherent “adaptability” (positive or negative) reflective of corticospinal tract development (Farmer et al. 2007) or pathology (Fisher et al. 2012; Grosse et al. 2003).

Transcranial direct current stimulation (tDCS) also has been shown to modulate cortical excitability, and the effects of tDCS may be directionally controlled. Data from previous studies have shown that anodal tDCS increases cortical excitability via increased neuronal depolarization following acute administration, whereas cathodal tDCS may depress cortical excitability via neuronal hyperpolarization (Ardolino et al. 2005; Liebetanz et al. 2002; Nitsche et al. 2003). Evidence in human studies suggests that tDCS can modulate corticospinal function (Reis et al. 2009; Stagg et al. 2012). In terms of directional effects of tDCS, Ardolino et al. (2005) showed that cathodal tDCS decreased the size of transcranial magnetic stimulation-induced motor-evoked potentials recorded from an intrinsic hand muscle compared with sham tDCS. Nitsche et al. (2003) demonstrated that anodal tDCS increased corticospinal excitability and cathodal tDCS decreased corticospinal excitability in hand muscles, based on stimulus-response (input–output) data. Therefore, tDCS parameters and electrode placements may be optimized for the type of clinical intervention being considered.
output) curves derived from varied transcranial magnetic stimulation intensities and subsequent motor-evoked potentials. Finally, in a hand muscle model, Power et al. (2006) found that cathodal tDCS decreased motor-evoked potential amplitude and $\beta$-band (15–35 Hz) intermuscular area of coherence, anodal tDCS increased motor-evoked potential amplitude and $\beta$-band intermuscular area of coherence, and sham tDCS had no effect on motor-evoked potential amplitude and $\beta$-band intermuscular area of coherence. Moreover, the authors reported a significant correlation between the change in motor-evoked potential amplitude and change in area of coherence. Together, the cumulative evidence provided by these prior reports suggests the likelihood for directionality effects of tDCS on intermuscular coherence. However, it remains unknown whether tDCS can modulate cortical drive to chest wall muscles.

If tDCS can modulate cortical drive to the chest wall measured as a change in intermuscular coherence, this may reflect a potential intervention for modifying synchronicity between chest wall muscles that could favor enhanced voluntary respiratory control. We determined if differences exist in cortical drive to muscles of the chest wall during respiratory tasks spanning a range of lung volume excursions. We then investigated whether tDCS could modulate cortical drive to the chest wall muscles. On the basis of previous findings that chest wall muscle recruitment is breathing task specific during expiration, we tested the hypothesis that intermuscular area of coherence in the $\beta$-band would differ among tidal, speech, phonation, and vital capacity breathing tasks. We also tested the hypothesis that 10 min of 1-mA anodal tDCS over the motor cortex would enhance chest wall intermuscular coherence during expiration compared with sham or cathodal stimulation.

**METHODS**

**Subjects.** Ten healthy adults (age: 34 ± 15 yr; age range: 18–57 yr; sex: 2 men, 8 women; body mass index: 24 ± 3 kg/m$^2$) free from respiratory disease who were nonsmokers participated in this study. Participants were excluded if they had a history of neurologic disorders including epilepsy or if they had an acute or chronic skeletal or muscle condition affecting the abdominal, thoracic, neck, or head region that would prevent the completion of the experimental protocol. Participants were otherwise healthy and self-reported being free from a known chronic health condition or neuromuscular disorder.

The University of Alberta Research Ethics Board approved this study and associated experimental procedures. The Board also approved the use of our tDCS device (details of our tDCS device are provided in the Appendix). All participants provided written and informed consent before commencing the study. This study was carried out in accordance with the Declaration of Helsinki (World Medical Association 2008).

**Experimental overview.** Participants reported to our laboratory in a rested state, were familiarized with our protocol, and underwent surface land marking for subsequent connection of recording and tDCS devices. Testing procedures included subjects performing prestimulation baseline breathing conditions consisting of resting tidal breathing, a speech task, a maximum phonation task, and a vital capacity maneuver. Chest wall surface EMG and kinematic recordings were made during these conditions. Participants were then randomized to one of three stimulation conditions (sham, cathodal, and anodal tDCS) in a single-blinded fashion. Breathing conditions were performed again with simultaneous EMG recordings 5 min following the stimulation condition. A minimum of 24 h separated testing between stimulation conditions. Data from Power et al. (2006) in healthy adults have shown that increases in $\beta$-band coherence following 1-mA cathodal or anodal stimulation over the motor cortex for 10 min begin to return to baseline 5 min following stimulation. The return to baseline effect is even greater at 10 min. Similarly, the time course of motor-evoked potential amplitudes for the abductor mini muscle following tDCS has been shown to have a stimulation duration-dependent recovery time. Specifically, 9 to 13 min of 1-mA anodal stimulation over the motor cortex have been shown to result in raised motor-evoked potential amplitudes above baseline lasting 30 to 90 min, respectively (Nitsche and Paulus 2001). Given these findings, we are confident that any change in cortical excitability normalized well within several hours following testing and that 24 h between experimental sessions served as an adequate washout period.

**Lung volume excursion conditions.** The lung volume excursion conditions performed were designed to replicate lung volumes and muscular effort encountered during varied breathing, and thus included speech and “nonspeech” conditions ranging from relatively low to high lung volume and muscle recruitment. Specifically, speech and tidal breathing occurs about midrange of vital capacity, and maximum phonation and vital capacity span the entire lung volume excursion range. All breathing conditions were performed with participants sitting upright, feet flat on the floor, and with their back unsupported. Resting acquiescent breathing (tidal breathing) was performed for 1 min. For the speech condition, volunteers read a standardized passage (“The Rainbow Passage”) as detailed previously (Bernthal and Bankson 1993). Participants then performed maximum duration phonation. For this condition, an investigator instructed participants to take a deep breath and say “ah” for as long they could at what the participant perceived to be a normal conversational loudness. Finally, participants performed a vital capacity maneuver with an investigator providing coaching and verbal encouragement. In terms of breathing depth before a task, we had participants follow a standardized protocol. This entailed participants performing resting tidal breathing before speech and the rest breathing tasks. Before vital capacity and phonation tasks, participants were instructed to take a deep breath beginning at their end-expiratory level (end-tidal volume) and then begin the task (expiration or phonation). Three trials of each task, aside from resting breathing, were collected and used for analyses.

**tDCS.** After performing the baseline breathing conditions, participants were randomized to receive sham, cathodal, or anodal tDCS from a battery-driven custom-built direct current stimulator (Appendix). Sodium chloride solution-soaked sponges (5 × 7 cm each) with a conductive-rubber electrode insert were positioned on the scalp over the left motor cortex region and contralateral supraorbital region. Sham tDCS was applied by ramping up the stimulator over 10 s and maintaining stimulation for 10 s at a target of 1 mA. Afterward, stimulation was turned off for 10 min of sham condition, and then a 10-s duration at 100 $\mu$A was applied to simulate the ramp-down procedure and sensation typically experienced with cathodal and anodal tDCS. Both the cathodal and anodal tDCS conditions entailed a 10-s ramp-up to 1 mA that was followed by 10 min of stimulation at 1 mA and a 10-s ramp-down stimulation termination procedure to 0 mA. Sponge electrodes were kept wet throughout the duration of each stimulation condition.

**Chest wall EMG and intermuscular coherence analysis.** Surface EMG recordings were made from the right side of the body over the sixth intercostal space and abdominal oblique regions using a standardized electrode surface placement protocol. Specifically, intercostal electrodes were placed ventrally 12 cm from midline, and oblique muscle electrodes were placed midway between the anterior superior iliac spine and caudal border of the rib cage. Two electrodes (Kendal Soft-E H69P; Tyco Healthcare Group, Mansfield, MA) were positioned 2 cm apart (center to center) and oriented parallel to fiber direction for the respective expiratory muscles (internal intercostal...
and external oblique). We are aware that within these regions other muscle groups may also contribute to surface EMG signals (e.g., external intercostal, internal oblique). Because external intercostals (inspiratory function) are superficial to the internal intercostals, we cannot exclude the possibility that some inspiratory muscle activation was captured during portions of expiratory maneuvers (Hixon et al. 1976). Our electrode placement protocol optimized ventral-dorsal EMG location to avoid the intercartilaginous region. EMG surface placement for the oblique muscle region was standardized as the midway point between the anterior superior iliac spine and the caudal border of the rib cage. Before testing, we confirmed appropriate EMG positioning for the intercostal and oblique muscles by inspection of EMG activity during inspiratory and expiratory maneuvers and trunk rotation tasks. Body movement during vital capacity and maximum phonation tasks would contaminate EMG recordings from non-respiratory-related muscle activation. To control for this potential methodological issue, we had participants perform multiple practice trials to avoid trunk flexion and extension. In addition, we used video recordings taken throughout the entire experimental protocol to ensure quality control in our selection of speech and nonspeech tasks. This procedure enabled us to only include vital capacity and maximum phonation trials that were performed without extraneous trunk (or limb) movement.

We employed differential EMG recordings to take advantage of the common mode rejection capabilities of the differential input connection. A third grounding electrode was used to minimize 50- or 60-Hz line interference. EMG signals were amplified (Grass P511; Quincy, MA), bandpass filtered (3–3,000 Hz), and sampled at 10 KHz. EMG data were collected with a multichannel data acquisition system (PowerLab 16SP ML795; ADInstruments, Colorado Springs, CO) and stored on a computer in the Chart software environment (v5.5.6; ADInstruments).

EMG activity was analyzed for the expiratory limb of breathing conditions with the EMG window selected from the onset of expiration to end of expiration. These data were then passed through a Tukey window to reduce erroneous high-frequency signals at the borders of adjoining breath trials, concatenated, and rectified. Intermuscular coherence was then calculated as

$$\text{MSC} = \left| C_{xy}(w) \right|^2 = \frac{\left| G_{xy}(w) \right|^2}{G_{xx}(w) \cdot G_{yy}(w)},$$

where MSC is the magnitude squared coherency (coherence), $G_{xy}(w)$ and $G_{xy}(w)$ are the averaged power spectra of x and y throughout the segments for a given frequency w, and $G_{xy}(w)$ is the averaged cross-power spectrum of signals x and y at frequency w (Halliday et al. 1995; Rosenberg et al. 1989). The phase relationship and cumulative density were also calculated as described by Halliday and Rosenberg (Halliday and Rosenberg 1999). The average total length of the concatenated data (windows) used for analysis was as follows: vital capacity, 58 s; maximum phonation, 70 s; tidal breathing, 57 s; and speech, 55 s. There was no overlap in the windows. The total number of segmented data in our study varied according to task. For the three conditions (pre- and postanodal, cathodal, and sham stimulation), an average of 142 (vital capacity), 170 (maximum phonation), 138 (tidal breathing), and 133 (speech) segments were analyzed for the respective tasks. Area of coherence was calculated as the area under the curve within the 15- to 35-Hz frequency band and above the 95% confidence limit to account for the combined frequency range and magnitude effect of breathing task and tDCS (Norton and Gorassini 2011, Ushiyama et al. 2011, 2012) and calculated as

$$\text{Confidence limits for each condition were Bonferroni-adjusted for multiple comparisons as previously described (Kilner et al. 2000; Ushiyama et al. 2011, 2012) and calculated as}

$$\text{CL}(\alpha) = 1 - \frac{1}{N} \left(1 - \frac{\alpha}{100}\right)^{1/(L-1)},$$

where $N$ is the number of frequency bins in the interrogated region of interest (i.e., $\beta$-band) and $L$ is the number of epochs. Thus, based on an $\alpha$ of 95, the confidence limits were determined to be 0.0356 for vital capacity, 0.0296 for maximum phonation, 0.0362 for tidal breathing, and 0.0375 for speech.

![Fig. 1. Top: raw data tracing of rib cage and abdominal kinematics and intercostal and oblique muscle electromyographic (EMG) activity during a vital capacity maneuver for a representative subject. Rib cage and abdominal kinematic signals are processed to yield a lung volume as described in METHODS. Bottom: coherence spectra are yielded from processing intercostal and oblique EMG signals as described in METHODS. Resulting coherence spectra calculated between the intercostal and oblique muscle regions for the same subject are shown. The area of coherence calculated under the curve and above the Bonferroni-adjusted 95% confidence limit (significance level) is highlighted in gray. The associated phase relationship corroborating significant coherence is shown in the inset. V, volts; AU, arbitrary units; rad, radian.](http://jn.physiology.org/ Downloaded from http://jn.physiology.org/)
Chest wall kinematics and analysis. Variable inductance plethysmography (Respiratr; Ambulatory Monitoring, Ardsley, NY) was used to measure chest wall kinematics to demonstrate volume-dependent differences across behavioral tasks and to determine potential changes associated with tDCS. The rib cage transduction band was placed with the top border positioned at the level of the 7th rib, and the abdominal transduction band was positioned with the lower border within 1 cm above the iliac crests and the top border below the costal margin. Chest wall kinematic signals were calibrated against a simultaneously measured known volume for each participant. The volume was derived from a bidirectional pneumotach (Hans Rudolph, Kansas City, MO) that measured flow at the mouth and was transduced (model DP45-14; Validyne Engineering, Northridge, CA), time-integrated, and calibrated against a 3-l syringe. Pneumotach-derived volume measurements taken at airway opening were time-locked to the chest wall kinematic signals for several resting tidal breaths. Isovolume maneuvers were performed by each participant to establish the relationship of the rib cage to abdomen under a closed system (Konno and Mead 1967). Subsequent gain adjustments were applied to the rib cage and abdomen signals to account for their respective relationship to lung volume displacement. After calibration, lung volumes were derived from the sum of rib cage and abdomen displacements. Lung volume initiation and termination were referenced to end-expiratory level. Lung volume excursion (initiation—termination) was expressed as a percentage of predicted vital capacity. The relative rib cage contribution was calculated by dividing rib cage excursion by lung volume excursion.

Statistics. The effect of lung volume excursion (controlled via tidal breathing, speech, phonation, and vital capacity maneuvers) from all prestimulation trials (baseline testing) was assessed using Friedman repeated-measures ANOVA on ranks because unequal variances were found. Given a significant lung volume excursion condition effect, multiple comparisons were made using the Student-Newman-Keuls method. The effect of tDCS on chest wall intermuscular coherence was then determined by one-tailed (a priori) paired t-tests based on the theoretical rationale that sham tDCS would have no effect on intermuscular coherence and because of directional expectations associated with cathodal vs. anodal stimulation (Ardolino et al. 2005; Nitsche et al. 2003; Power et al. 2006). Wilcoxon signed-rank testing was performed when tDCS data were not normally distributed (Shapiro-Wilk test, P < 0.05) (SigmaPlot 12.0; Systat Software, San Jose, CA). Data are means ± SD, and findings are considered significant when P < 0.05.

RESULTS

Effect of lung volume excursion condition on chest wall intermuscular coherence. Figure 1, top, shows raw chest wall kinematic and EMG tracings for a representative subject during vital capacity expiration. There was a significant effect for lung volume excursion (breathing condition) on area of coherence (P < 0.001). Multiple comparison analysis indicated that the areas of coherence for tidal breathing and speech breathing were significantly greater than for phonation and vital capacity breathing. In addition, the area of coherence for tidal breathing was greater than for speech breathing (all P < 0.05; Fig. 2, top). These findings suggest that the oscillatory coupling between the expiratory muscles that we interrogated is associated, in part, with the volume of excursion. The decrease in area of coherence from low volume (tidal breathing) to high volume (vital capacity) excursions was observed in a majority of participants (Fig. 2, bottom).

Effect of tDCS on chest wall intermuscular coherence. Figure 3 shows the area of chest wall intermuscular coherence for each breathing condition and for each pre- and poststimulation condition. Sham and cathodal tDCS had no effect on chest wall intermuscular coherence for any condition (all P > 0.05). However, coherence was more variable among subjects during tidal breathing and phonation noted by the standard deviations for those conditions shown in Fig. 3, thus making it difficult to detect a potential cathodal stimulation effect. Anodal tDCS increased area of coherence from 0.04 ± 0.09 prestimulation to 0.18 ± 0.19 poststimulation during vital capacity expiration (P < 0.05). Figure 4 shows coherence spectra for a representative subject following each stimulation condition (anodal, cathodal, and sham) during vital capacity expiration. In the subject presented, anodal stimulation caused a notable increase in intermuscular coherence in the 15- to 35-Hz range, whereas cathodal stimulation and sham stimulation had no effect on coherence.

Effect of tDCS on chest wall kinematics. Initiation and termination volumes during expiration for each breathing task were not affected by any tDCS condition (Fig. 5). The total lung volume excursion normalized as a percentage of predicted vital capacity was also not affected by any of the tDCS conditions (Fig. 5). Additionally, tDCS had no effect on rib cage contribution to lung volume excursion (Fig. 5).

DISCUSSION

The major novel findings of this study were that chest wall intermuscular coherence varied as a function of the type of lung volume excursion and that 10 min of 1-mA anodal tDCS...
increased chest wall intermuscular coherence during vital capacity breathing. Cathodal stimulation and sham stimulation had no effect on area of coherence on any breathing condition. Brief tDCS did not affect respiratory kinematics. These findings support our hypothesis that the neurally mediated coupling between the same chest wall muscles varies as a function of the type of lung volume excursion performed. Another finding was that the modulation of chest wall intermuscular coherence by tDCS was stimulation and lung volume excursion task specific.

**Effect of expiratory task on chest wall intermuscular coherence.**
Total lung volume excursion was likely the primary modulating factor accounting for differences in intermuscular coherence. Indeed, Hoit et al. (1988) found in healthy subjects that abdominal EMG activity about the oblique muscle region was incrementally greater with lower lung volumes during speech. Our finding that speech and tidal breathing occurred in the midrange of vital capacity, whereas maximum phonation and vital capacity occurred throughout the entire range of expiration, coupled with previous reports (Hoit et al. 1988) supports our notion that lung volume excursion dictates, in part, the degree of coherence between chest wall muscles.

Intermuscular coherence during breathing tasks occurring in the midrange of vital capacity was greater compared with that during breathing tasks spanning the entire range of vital capacity. This finding is not surprising given that task dependency has been established to affect β-band intermuscular coherence in limb muscles. For example, Kilner et al. (1999) found the presence of intermuscular coherence between intrinsic and extrinsic hand muscles during a sustained hold task, but not during the initial positioning of the hold task (during "dynamic" movement). Our finding suggests that greater common cortical drive to chest wall muscles is required for speech and conscious tidal breathing compared with that required for breathing that spans the full potential range of lung volume excursion. Specifically, the increased cortical drive to intercostal and abdominal oblique muscles may occur, in part, because of the increased complexity of motor task. To produce "effective" speech, speech breathing requires maintaining a relatively constant tracheal pressure while performing complex articulatory movements. Subsequently, this process typically requires activation of the internal intercostal followed by the abdominal oblique muscles to carefully regulate tracheal pressure. In this regard, lung volume excursions during speech may be considered to require more carefully regulated motor control relative to vital capacity and maximum phonation lung volume excursion. Indeed, Perez et al. (2012) have also found that precision grip tasks, despite increasing contraction effort, are also associated with increased coherence. Together, these findings suggest that relatively fine motor control in distal and proximal muscles may be reflected by greater neural coupling between synergistic muscles. However, given that maximum phonation also requires careful regulation of a constant tracheal pressure for sound production throughout expiration, our findings suggest that the increasing contraction strength that occurs with decreasing lung volumes during phonation may not increase neural coupling between the intercostal and oblique muscles.

The smaller area of coherence during greater lung volume excursions may reflect less reliance on oscillatory drive during such tasks (phonation and vital capacity). This finding is similar to previous reports (Perez et al. 2012), where cortico-muscular coherence was found to decrease with increasing contraction effort during finger abduction. A relative reduction in coherence may also be explained by the functional role of the muscle and/or its level of skilled training and subsequent fine motor control (Ushiyama et al. 2010).

An alternative explanation for our finding of a lower area of coherence associated with maximum phonation and vital capacity breathing compared with speech and tidal breathing may consider differences in muscle recruitment. Greater lung volume excursions may rely on other chest wall muscles not measured in the current study, such as the rectus abdominis and latissimus dorsi, in addition to the internal and external intercostals and oblique muscles that are employed during maximum phonation and vital capacity breathing (Hixon and Weismer 1995). This recruitment likely occurs to a greater degree compared with speech and tidal volume breathing. Given that
a greater lung volume excursion requires greater muscle recruitment, the lower area of intermuscular coherence that we report during greater lung volume excursions may be the result of shift in muscle use during such events. Whether this is reflected by a difference in the cortical drive to other chest wall muscles subsequently requires study.

Effect of tDCS on chest wall intermuscular coherence. It is clear from previous reports that tDCS can effectively modulate cortical excitability (Ardolino et al. 2005; Jeffery et al. 2007; Liebetanz et al. 2002; Nitsche et al. 2003; Power et al. 2006; Reis et al. 2009; Stagg et al. 2012). For example, tDCS has previously been shown to significantly improve motor-evoked potentials and intermuscular coherence in healthy individuals and in those with neuromuscular injury. In their study, Jeffery et al. (2007) found that 10 min of 2-mA anodal tDCS significantly increased transcranial magnetic stimulation-induced motor-evoked potentials of the tibialis anterior in healthy subjects compared with sham and cathodal tDCS. Similarly, Kaski et al. (2012) found in healthy adults that locomotor adaptation was enhanced during stable-unstable platform stepping protocol following anodal tDCS compared with sham tDCS. To our knowledge, however, no study has determined the effect of tDCS on chest wall intermuscular coherence.

In the present investigation, we found that anodal tDCS increased chest wall area of coherence during vital capacity expiration (Fig. 3). The lack of effect observed with cathodal stimulation is consistent with previous reports (Jacobson et al. 2012). Given our observation that chest wall area of coherence was excursion volume dependent, our tDCS findings suggest that the modulation of cortical excitability may also be related, in part, to lung volume. Specifically, lung volume excursions eliciting relatively less chest wall intermuscular coherence (vital capacity) were only responsive to anodal stimulation. From a clinical rehabilitation perspective, our finding in healthy adults suggests that anodal tDCS is most effective for enhancing chest wall intermuscular coherence during breathing tasks that span the full range of lung volume excursion.

As we have noted, EMG activity was analyzed for only the expiratory limb of breathing conditions (onset of expiration to end of expiration). However, data from Hixon et al. (1976) have shown that inspiratory muscle activity can be present...
during much of expiration, depending on vocal loudness in combination with prevailing lung volume. Thus a limitation of our study is that both inspiratory and expiratory muscle activity reflected in the external and internal intercostal muscles, respectively, may have been introduced into our coherence analyses. Additionally, our study sample was relatively small and our tDCS-related statistical analyses directional, although consistent with previous reports (Ardolino et al. 2005; Nitsche et al. 2003; Power et al. 2006). We note that the repeated-measures design of our experiment adds inherent statistical power. Additionally, surface EMG placement has been shown to affect the magnitude of the subsequent calculated intermuscular coherence (Keenan et al. 2012). However, electrode placement was standardized and consistent between testing days. Conclusion. We have demonstrated that motor cortex control of the chest wall during voluntary expiration, measured as the area of coherence between respiratory muscles, varies as a function of breathing task. Thus we conclude that chest wall intermuscular coherence is, in part, lung volume excursion dependent. Our findings also suggest the possible use of tDCS for potentiating synchronized force development of chest wall muscles during voluntary expiration. This latter finding may be beneficial in its adjuvant application to rehabilitation efforts for neuromuscular deficits that affect voluntary muscles during voluntary expiration. This latter finding may be beneficial in its adjuvant application to rehabilitation efforts for individuals with neuromuscular deficits that affect voluntary respiratory control. The optimal lung volume excursion maneuver and tDCS combination in clinical groups requires investigation.

APPENDIX

Our custom-built tDCS device employs a power supply (2 × 9-V DC batteries in series), adjustable voltage regulator, analog voltmeter, and analog amp meter as shown in Fig. A1. We calibrated the amp meter of the tDCS device with an adjustable voltage across a known resistance to obtain the required DC current. The current value was then checked and confirmed using two different four-digit digital amp meters (Fluke 189 True RMS Meter; Beckman RMS 3080). In both measurements, digital amp meters were found to correspond with the tDCS device current output. With the use of this same procedure, our tDCS device has also been found to maintain a stable and accurate current output for a minimum of 20 min. We have reproduced these findings on separate testing days.

GRANTS

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DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the authors.

AUTHOR CONTRIBUTIONS


REFERENCES


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EXPERIMENTAL TASK AND \( \text{tDCS} \) EFFECTS ON CHEST WALL CONTROL


