Can visual feedback on upper trapezius high-density surface electromyography increase time to task failure of an endurance task?

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A B S T R A C T

We investigate whether visual feedback on the spatial distribution of upper trapezius muscle activity can prolong time to task failure of sustained shoulder abduction. Surface electromyographic signals were acquired with a 13x5 grid of high-density electromyography (HDEMG) electrodes from the right upper trapezius muscle of 12 healthy volunteers as they performed sustained isometric shoulder abduction at 20% of their maximum voluntary contraction torque (MVC) until volitional exhaustion. Data were collected in two sessions; one with HDEMG visual feedback on the spatial distribution of upper trapezius activity and one without feedback. Although the HDEMG amplitude maps could be voluntarily modified by the participants during the feedback condition (significant shift in the barycenter of activity towards the cranial direction, P = 0.038), this did not influence endurance time (total endurance time with HDEMG feedback: 149.01 ± 77.07 s, no feedback 141.74 ± 60.93 s, P = 0.532). Future studies should assess whether endurance performance can be enhanced by allowing changes in arm position during the task (changing fiber tension-length relationships), by providing a more individual motor strategy, and/or by manipulating the colours used for the HDEMG maps (lighter colours for higher contraction intensities).

1. Introduction

Muscle fatigue is defined as an exercise-induced reduction of motor performance (i.e. maximal force or power production) (Bigland-Ritchie et al., 1984) which is thought to reflect both myoelectric (peripheral and central adaptations) and mechanical phenomena (Merletti et al., 1996).

During the development of fatigue, the central nervous system (CNS) adapts by changing the recruitment, firing rates and peripheral properties of motor units (Martinez-Valdes et al., 2016; Samani et al., 2017). HDEMG studies have shown that the spatial distribution of EMG amplitude is not homogenous (i.e. the muscle's electrical activity is distributed in a non-uniform manner across the muscle) and that the spatial distribution of muscle activity varies over time during a sustained fatiguing contraction (Falla and Farina, 2007a; Falla and Farina, 2008; Farina et al., 2008). Additionally, studies have found a correlation between changes in the spatial distribution of muscle activity and total endurance time (i.e. a more heterogeneous activation allowed longer times to task failure) during sustained contractions (Farina et al., 2008; Sanderson et al., 2019). This finding was interpreted as an efficient neural strategy by decreasing the demand on any specific muscle region.

Real-time processing of HDEMG signals and the generation of EMG amplitude maps to display the spatial distribution of muscle activity could be used as feedback to encourage people to activate specific regions of a muscle, with the aim of minimizing muscle fatigue. In previous work by Gaffney et al. (2016), HDEMG feedback was provided to asymptomatic people with the aim of altering their cervicoscapular posture by shifting the spatial distribution of trapezius muscle activity caudally during a computer-typing task. This study showed that the HDEMG biofeedback was more effective than verbal feedback alone in improving cervicoscapular posture (in the short-term) by inducing a caudal shift of trapezius muscle activity. Whilst this work provides preliminary evidence on the usefulness of HDEMG biofeedback for modifying motor control strategies, no study has examined whether HDEMG biofeedback can be used to enhance endurance or minimize muscle fatigue.

Since a greater redistribution of upper trapezius muscle activity during sustained contractions has been associated with longer times to task failure (Farina et al., 2008), we hypothesized that endurance could be prolonged by voluntarily inducing a redistribution of upper trapezius muscle activity with the assistance of visual feedback on HDEMG.

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Therefore, we investigated the influence of real-time feedback on upper trapezius HDEMG versus no feedback on endurance time when performing a sustained isometric voluntary contraction of shoulder abduction until volitional exhaustion.

2. Methods

2.1. Participants

Twelve healthy participants (6 men, mean ± standard deviation; age: 25.08 ± 1.98 years; height: 167.67 ± 6.40 cm; weight: 67.33 ± 12.44 kg) were recruited from the University of Birmingham staff / student population and local contacts via leaflets and e-mail.

Inclusion criteria were: (1) men or women aged between 18 and 35 years old with (2) full, pain-free passive and active range of motion of the neck and shoulder. The exclusion criteria were: (1) current history of neck or arm pain which required treatment of a health professional (2) history of orthopaedic disorders affecting the shoulder or neck region and (3) history of neurological disorders. All participants were right handed.

2.2. Study design and setting

This cross-sectional study with a cross-over design was conducted in May 2018 at a laboratory within the Centre of Precision Rehabilitation for Spinal Pain (CPR Spine), University of Birmingham, UK. The Ethics Committee of the School of Sport, Exercise and Rehabilitation Sciences, United Kingdom, approved the study (CM 190218-1). All participants provided written consent prior to participation. The study was conducted according to the Declaration of Helsinki and the study is reported according to the STROBE guidelines.

All participants attended two experimental sessions which lasted approximately 1 h each which were separated by seven days. Their maximal voluntary shoulder abduction force was recorded at baseline and immediately following isometric sustained bilateral shoulder abduction at 20% MVC until failure. HDEMG was acquired from the right upper trapezius in the same manner in both sessions, but in one session (randomly chosen to reduce bias) a topographical map of the HDEMG amplitude was displayed as real-time feedback to the participants whereas in the other session no visual feedback was provided. The procedure is presented schematically in Fig. 1.

2.3. Measurement set-up

All participants were seated comfortably with their back resting against a chair in front of an isokinetic dynamometer as shown in Fig. 2. Their hips and knees were in 90° of flexion, and their feet were flat on the floor at a distance equal to the distance between the two acromia. Both shoulders were positioned in 90° of abduction with elbows fully extended and their palms facing down. In this position, the isokinetic dynamometer was adjusted according to each participant so that the axis was aligned with the axis of rotation for shoulder abduction. A wooden pole was used as an extension of the dynamometer attachment to ensure that both shoulders were positioned at the same level. Additionally, the location of resistance was standardised and positioned at the distal end of the humerus. The settings and position of the set up (i.e. chair, isokinetic device and electrodes placement) were marked so that the participants' position and electrode placement were similar in each session.

2.4. HDEMG

Prior to electrode placement, the skin was gently abraded (Nuprep Skin Prep Gel, Weaver and Company, Aurora, Colorado) to reduce skin impedance, and then cleaned with water. If necessary, the skin was shaved to remove hair.

In the first session, the location of the main innervation zone of the upper trapezius was determined using a linear array of 16 electrodes (silver bars, 93 mm long, 20 mm width and 5 mm inter-electrode distance, OT- Bioelettronica, Torino, Italy) along the line from C7 to the acromion as previously described (Farina et al., 2002) as the subjects held both shoulders in 90° of abduction. Differential acquisition mode was used during this process and all signals were converted from analog-to-digital by a 16-bit converter (Quattrocento- OT- Bioelettronica, Torino, Italy). The sampling frequency used, was 2048 Hz and the amplifier gain was set to 150. EMG signals were digitally filtered with a bandwidth set up to 10 Hz for high pass cut frequency and to 500 Hz for low pass cut frequency.

Following the identification of the innervation zone, the linear array was removed and a two-dimensional (2D) adhesive grid (SPES Medica, Salerno, Italy) of 13 × 5 equally spaced electrodes (each of 1 mm diameter, with an inter-electrode distance of 8 mm) was positioned over the upper trapezius muscle. The fourth row of electrodes aligned along the line between C7 and the acromion with the most lateral electrode column positioned 10 mm from the identified innervation zone as previously described (Falla and Farina, 2007b). Conductive paste (AC-CREAM, SPES Medica, Genova, Italy) was inserted into the cavities of the grid. HDEMG was acquired in monopolar mode with reference electrodes (WhiteSensor WS, Ambu A/S, Ballerup, Denmark) positioned over C7 and the acromion. The electrode grid and reference electrodes were all connected to the same bioelectrical amplifier (Quattrocento-OT- Bioelettronica, Torino, Italy).

The torque exerted by the participants was assessed with an isokinetic
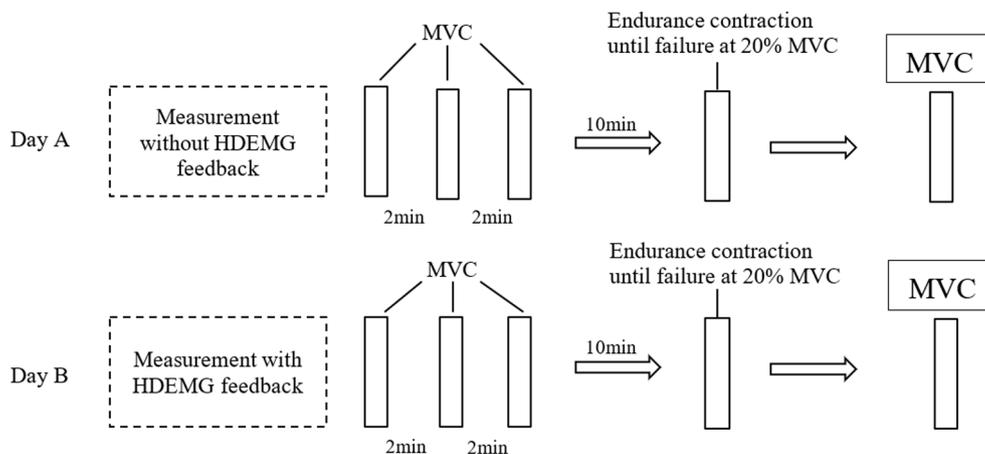

Fig. 1. Schematic representation of the measurement sessions. Maximum voluntary contraction, MVC. The two measurement days were randomised.

Fig. 2. Participants' standardised position for the two sessions. Seated on the chair in front of the isokinetic dynamometer which was individually adjusted. A wooden pole was used as an extension of the dynamometer attachment to ensure that both shoulders were positioned at the same level.

dynamometer (System 3 Pro, Biodex Medical Systems, New York), which was synchronized with the HDEMG signals. The synchronization was obtained by recording the torque signals generated by the isokinetic dynamometer through the auxiliary input of the EMG amplifier.

2.5. Procedure

At the beginning of the experiment, the participants were familiarized with the dynamometer and practiced the required contractions at low force levels. They then performed three MVCs of right shoulder isometric abduction of 5 s duration. Each trial was separated by 2 min of rest. The highest MVC value was used as the reference maximal torque. In each of the two experimental sessions, the submaximal torque was relative to the MVC measured during the same session. All participants were given a 10-min rest after the MVC measurements. They then performed sustained isometric shoulder abduction at 20% of the MVC until failure. Task failure was defined as the time instant when the participant exerted a torque 10% MVC below the target level for a time interval of 2 s (Castronovo et al., 2015; Martinez-Valdes et al., 2017). During this contraction, both arms were positioned in 90° of shoulder abduction (to avoid lateral flexion of the trunk) with full elbow extension and with the forearms pronated. The wooden pole provided tactile position feedback to the participant to ensure that both arms were maintained at 90° (Fig. 2). However, it was not possible for the participants to exert torque through the wooden pole (i.e. through the contralateral arm/shoulder). Immediately after task failure, the participants performed one further MVC to quantify the decrease in MVC torque.

In both sessions, the participants received auditory feedback of their target %MVC torque (20%) using a customized computer software. Participants were allowed to visualize the target torque (20% MVC) for 5 s and then the visual torque feedback was removed, and they were instructed to solely use auditory feedback. The absence of sound meant that the participants were at the desirable target ± 3% MVC error. A deviation out of this range triggered an audible sound with either a higher or lower frequency. The participants were then encouraged to push more when the audible sound frequency was low or push less when the audible sound frequency was high.

For HDEMG feedback, the signals were presented to the participants in monopolar mode to obtain higher sensitivity to changes in the spatial distribution of activity. The refresh rate of the visual HDEMG feedback was set at 4 Hz. The colours used in the visual feedback were normalized according to the peak value (µV) of the three MVCs which was set as the maximum Average Rectified Value (ARV). As presented in the colour bar (Fig. 3) the blue or cold colours represented lower ARV values (i.e. low EMG amplitude) while the red colour or hot colours represent higher EMG amplitude. The map was displayed on a computer monitor in the same direction as it was positioned on the skin. To improve the participants' understanding of the position of the electrodes in relation to the displayed muscle activity, the investigator used tactile pressure over different regions of the electrode grid, as a way to induce noise to the signals which appeared as red (high EMG amplitude) to the participants in the corresponding region of the grid. Once the participants were fully familiarised with the visual feedback, they were instructed to shift the centre of gravity of the HDEMG map towards the portion of the upper trapezius that was less activated which was represented with blue or cold colours (i.e. if they were activating a lower portion of the muscle, they were instructed to activate the upper portion of the muscle).

2.6. Data processing

Monopolar signals were band-pass filtered (20–450 Hz) and differentiated in the direction of the muscle fibers (in the direction of the
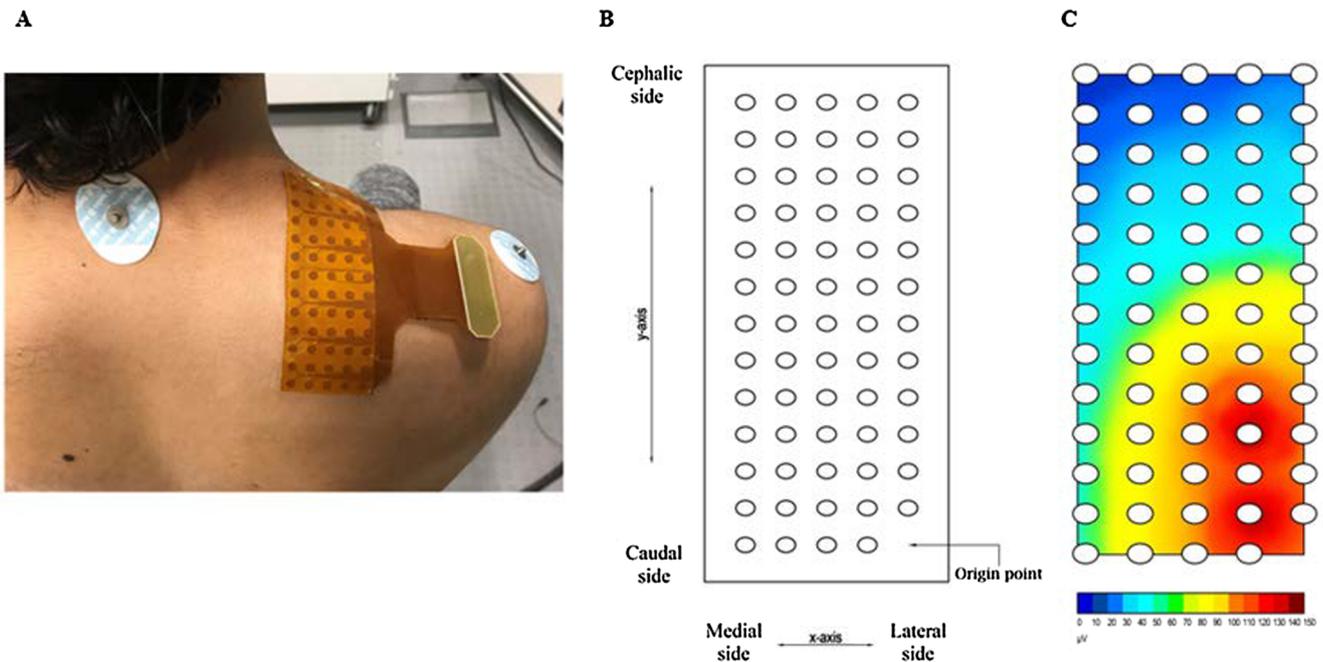
A   **B**   **C**

Fig. 3. Electrode placement over the upper trapezius (A), representation of the electrode with the axes of coordination and the origin point which was used to define the electrode's position (B), visual feedback of the upper trapezius EMG amplitude map (C).

electrode rows) to form 51 adjacent bipolar channels. HDEMG amplitude (ARV_{mean}, averaged across the 51 bipolar channels), the mean power spectral frequency (MNF_{mean}, averaged across the 51 bipolar channels), heterogeneity of muscle activity (Entropy) and displacement of the centre of activity (Y and X-axis barycentre) were calculated from the single differential signals as shown previously (Farina et al., 2008) (Fig. 3). Additionally, in order to check for the influence of auditory feedback across days, we quantified the coefficient of variation of torque and mean torque across conditions. HDEMG and torque variables were then analysed in 10 normalized segments according to total endurance time (10, 20, 30, 40, 50, 60, 70, 80, 90 and 100% of endurance time). All HDEMG and torque variables were calculated offline by using a custom algorithm on MATLAB (R2018a).

2.7. Statistical analysis

The IBM SPSS Statistics 24 computer software was used for statistical analysis of the data. Shapiro-Wilk tests were first performed to confirm that the data (endurance time, MVC and EMG data) followed a normal distribution. Once this was confirmed, paired t-tests were performed to compare the endurance times between sessions, the MVCs before the sustained contraction for both conditions and for the pre and post MVCs for each condition (i.e. with or without HDEMG feedback). Two-way repeated measurements analysis of variance (ANOVA) was performed for the ARV_{mean}, MNF_{mean}, Y-axis barycentre, X-axis barycentre, entropy, coefficient of variation of torque and mean torque variables followed by a Bonferroni post hoc test when appropriate. More specifically, the data were analysed from the whole group of participants using as within-subjects factors the time (10 levels) and condition (with or without HDEMG feedback). Additionally, the Y-axis barycentre total displacement was further investigated by calculating the mean absolute value of change during the fatiguing task (i.e. total displacement of the barycentre during the task, which was quantified as the difference between epochs 10 and 1) for each condition and then comparing their means with an additional paired t-test. The level of significance, α was set to 0.05. The data are presented as mean values and standard deviations.

3. Results

The data obtained from one subject was excluded from all analyses due to poor EMG signal quality. As a result, data from 11 healthy participants, 6 males and 5 females (mean ± standard deviation; age: 25.27 ± 1.95 years; height: 168.36 ± 6.22 cm; weight: 68.64 ± 12.16 kg) were used for the analysis. The results for all variables are presented in Fig. 4.

3.1. Endurance time and MVC

The MVC performed at the beginning of each session were similar between sessions; 35.08 ± 9.24 N·m and 35.04 ± 10.59 N·m for the no-feedback and feedback session, respectively (P = 0.977).

The reduction of the MVC following the sustained contraction was also similar in both sessions and decreased by 28.99 ± 19.50% versus 23.36 ± 13.96% MVC after the feedback and no-feedback conditions respectively (F = 0.024, P = 0.881).

Mean torque and the coefficient of variation of torque exerted by the participants was also similar across conditions (mean torque condition effect: F = 0.145, P = 0.711 and coefficient of variation of torque condition effect: F = 1.054, P = 0.329).

HDEMG feedback did not influence time to task failure as the participants sustained the contraction for 149.01 ± 77.07 s versus 141.74 ± 60.93 s for the feedback and no-feedback conditions, respectively (P = 0.532). Four out of the eleven participants improved their performance during the feedback condition by an average of 33.63 ± 20.71%.

3.2. Changes in EMG variables

The ARV_{mean} values, increased similarly during the sustained contraction in both conditions (time effect: F = 7.003, P = 0.000) (Fig. 4A). Likewise, the MNF_{mean} values decreased similarly during the contraction in both conditions (time effect: F = 19.666, P = 0.000) (Fig. 4B).
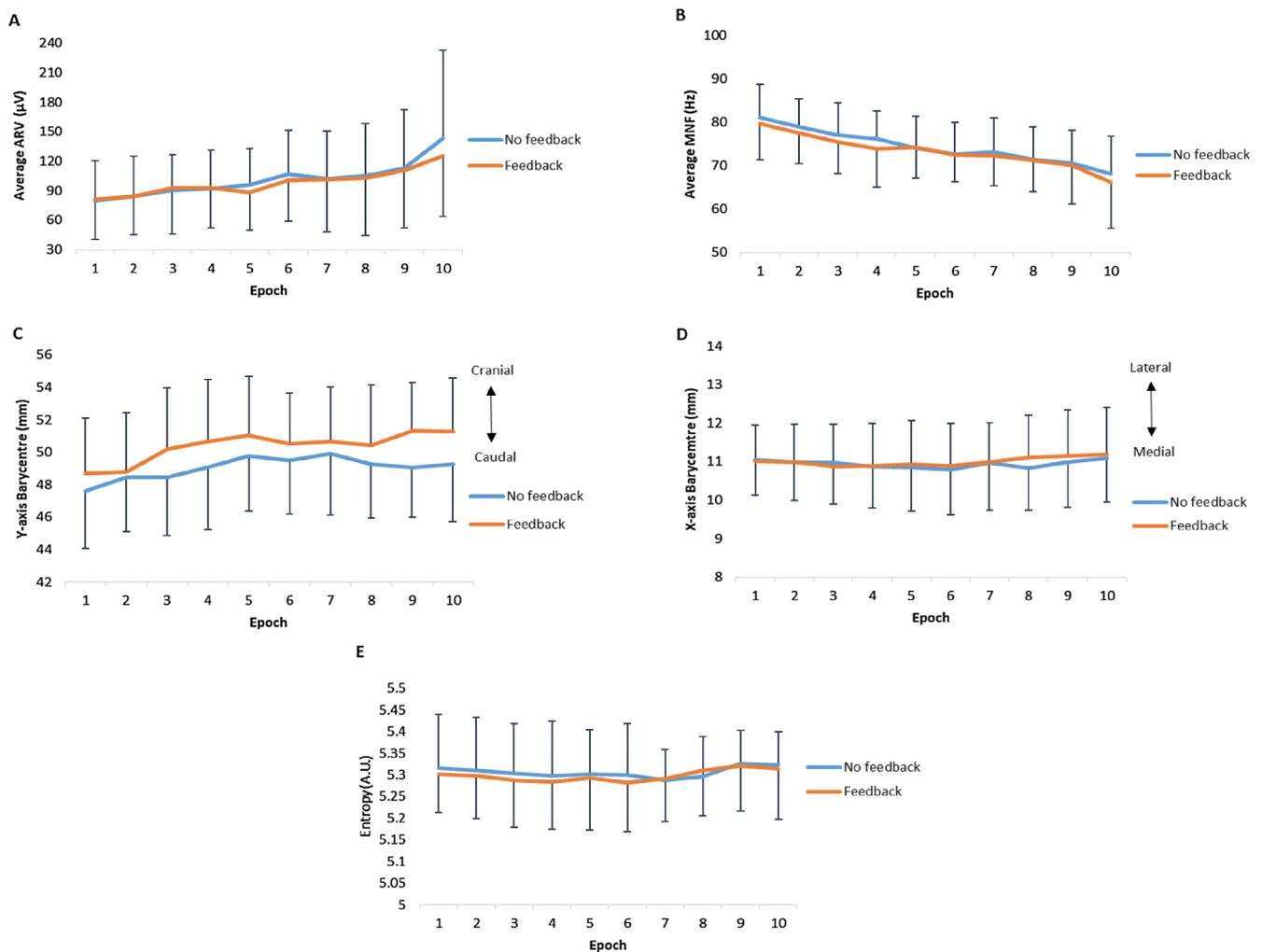
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Fig. 4. Mean (± SD) changes in average rectified value (ARV) amplitude (A), mean power spectral frequency (MNF) amplitude (B), Y-axis barycentre displacement (C), X-axis barycentre displacement (D) and entropy (E) at 0–100% of contraction. Data are presented for both conditions.

3.3. Changes in the position of the barycentre and entropy

According to the ANOVA test results, the displacement of the centre of activation along the Y-axis (across the 10 epochs) differed significantly between the two conditions with a more cranial displacement during the condition with feedback (condition effect: $F = 5.704$, $P = 0.038$) (Fig. 4C). Displacement along the X-axis (condition effect: $F = 0.022$, $P = 0.886$) and entropy variables (condition effect: $F = 0.051$, $P = 0.826$) were similar in both conditions (Fig. 4D and 4E, respectively).

3.4. Total displacement along the Y-axis during the contraction

The results from the additional test that we performed, showed that the total displacement of the barycentre along the Y-axis (i.e. difference between epoch 10 and epoch 1) was similar for both conditions (no feedback: 2.28 ± 1.60 mm, feedback: 2.80 ± 1.53 mm) ($P = 0.426$).

4. Discussion

We investigated whether healthy individuals could enhance their endurance performance during a sustained contraction until volitional exhaustion by using real-time visual feedback of upper trapezius HDEMG. Although HDEMG feedback resulted in a significant cranial displacement of the barycentre of the upper trapezius activity, relative to the condition with no feedback, this did not influence endurance time. Thus, although participants were able to alter the spatial distribution of their upper trapezius muscle activity, specifically by engaging more cranial regions of the muscle, this did not alter their motor performance.

4.1. Changes in ARV, MNF and entropy variables

The average ARV values and MNF values (indicators of muscular fatigue), increased and decreased respectively during the sustained contraction, similarly as in previous studies evaluating fatiguing contractions (Farina et al., 2008; Gallina et al., 2013). The (1) recruitment of additional motor units, (2) the reduction in propagation velocity of the depolarization across muscle fibers (leading to increased duration of action potentials) and (3) the shape changes of the intracellular action potentials can determine this increase in EMG amplitude (Farina et al., 2010). Likewise, factors 2 and 3 in addition to the accumulation of metabolic by-products contribute to the fall in MNF values (Falla et al., 2014).

Conversely, entropy values remained stable and similar between conditions. Higher entropy values reflect a more uniform spatial distribution of muscle activity, while lower values represent the opposite (Hyngstrom et al., 2018; Martinez-Valdes et al., 2019). Farina et al., (2008) observed a reduction in entropy values during sustained shoulder abduction, while Hyngstrom et al., (2018) observed the opposite. Heterogeneity between studies regarding the measurement set-

up and the sample used could partly explain such differences. More specifically, in the first study, participant's force output was not standardised with visual feedback and their arm was not entirely restricted. Thus, some participants could have exerted higher levels of force and/or move their arm more freely, leading to higher heterogeneity of muscle activity. The second study measured post-stroke patients and the EMG data was acquired from the knee extensors during sustained-isometric contractions after ischemic conditioning, which restricted limb positioning and likely reduced the heterogeneity of muscle activity. It is possible that our setup also induced similar restrictions in limb positioning, which likely influenced the lack of change in entropy values during the feedback condition.

4.2. Lack of changes in endurance with HDEMG feedback

We expected that the HDEMG feedback would prolong endurance time. Such a hypothesis was reasonable, as endurance time has been shown to be correlated with a change in the extent of spatial distribution of muscle activity over time (Farina et al., 2008). Even though the participants were able to influence and change the spatial distribution of upper trapezius muscle activity, their performance did not change.

This was the first study to investigate if HDEMG feedback has an effect on endurance. As a result, comparisons with other studies regarding performance cannot be made. However, Farina et al. (2008) and Falla et al. (2010) observed a similar redistribution of muscle activity towards the cranial region of upper trapezius in healthy participants during sustained shoulder abduction. It was reasoned that avoiding overload of the same part of the muscle could be an efficient way to maintain motor output and therefore improve performance (i.e. prolong endurance time). However, the chosen sensitivity and colours of the HDEMG feedback, or the specific fixed arm position could have influenced performance and prevented the participants from enhancing their endurance. Indeed, the threshold value used for the representation of the highest ARV values could have been too low (making HDEMG feedback highly sensitive), and thus resulting in faster appearance of the red/hotter colours. The reported relation between colour and psychological functioning could have been a barrier in participants' performance (Elliot et al., 2007). Red colour which was used as an indication of greater muscle activation and fatigue, has been associated with danger (Elliot et al., 2007) which possibly could have impacted on participants' affect, cognition and behaviour without their conscious awareness and more specifically it could have evoked an avoidance of motivation as they could have thought that they were already fatigued when visualising the red colours (Bargh, 1990). The use of a different colour or/and a less sensitive HDEMG feedback could have led to different results and should be explored in future studies. Another factor that might have influenced our results is the standardised, fixed arm position used during the fatiguing task. In the feedback condition, the participants were able to redistribute their activity towards a different portion of the muscle (cranial region) as shown by the shift of the barycentre along the Y-axis (caudal-cranial direction). However, since they could not move their extremity and thus place their muscle in positions of more mechanical advantage (i.e. by changing the length-tension relationship of the muscle), they failed at a similar time as the no-feedback condition (Gordons et al., 1966). The upper trapezius is functionally subdivided with complex muscle fibre architecture and depending on the arm posture there is a different line of action of individual upper trapezius muscle fibres (Lindman et al., 1990; Jensen and Westgaard, 1995; Holtermann et al., 2008; Mendez-Rebolledo et al., 2018). Thus, subtle movement of arm position may have provided a better strategy for prolonging endurance.

4.3. Limitations

A small convenience sample of participants were recruited for this

study which could have increased the risk of selection bias and limited the external validity of the study (Higgins et al., 2011). A further limitation as described above, is that the task may have been too constrained especially since brief increases in force (i.e. ramp contractions) from 20% to 25% MVC during an isometric shoulder contraction have been shown to induce a greater change in the spatial distribution of upper trapezius muscle activity (towards the cranial direction) and reduce fatigue compared to sustained force contractions (Falla and Farina, 2007a). Furthermore, the level (20%MVC) and the nature of the contraction (isometric), could have limited the real-time visualization of displacements in the distribution of muscle activity as any variation may have been too small to be meaningful to the participant.

5. Conclusion

This study applied novel HDEMG feedback of upper trapezius activity to evaluate the influence of feedback on endurance time during a sustained shoulder abduction task. Although the participants were able to shift the centre of upper trapezius activity to a more cranial position when presented with visual feedback on HDEMG, this did not affect their endurance time. The specific HDEMG parameters used (i.e. sensitivity and arm position) could have influenced our results and this should be investigated further.

Declaration of Competing Interest

The authors declare no conflicts of interest and no financial disclosure.

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