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Force Perception At the Shoulder After a Unilateral Suprascapular Nerve Block

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RESEARCH ARTICLE

Force perception at the shoulder after a unilateral suprascapular nerve block

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Abstract
There are two key sources of information that can be used to match forces—the centrally generated sense of effort and afferent signals from mechanical receptors located in peripheral tissues. There is currently no consensus on which source of information is more important for matching forces. The corollary discharge hypothesis argues that subjects match forces using the centrally generated sense of effort. The purpose of this study was to investigate force matching at the shoulder before and after a suprascapular nerve block. The nerve block creates a sensory and muscle force mismatch between sides when matching loads. The torque matching accuracy did not change after the nerve block was administered. Directionally, the torque error was in the direction proposed by the corollary discharge hypothesis. However, the mismatch between deltoid EMG was substantially greater compared to the changes in the torque matching error after the block. The results support that sensory information is used during force matching tasks. However, since the nerve block also created a sensory disruption between sides, it is not clear how sensory information is reweighted following the nerve block and a role for sense of effort is still implicated.

Keywords Supraspinatus · Isometric ramp contraction · Deltoid · EMG · Suprascapular nerve · Force perception

Introduction

To successfully perform any movement, muscle forces must be carefully judged so that the movement outcome is what was intended. The sensory information that provides feedback on the status of a limb in space comes from a variety of mechanical receptors located in muscles, tendons, joint capsules, ligaments and the skin (Riemann and Lephart 2002). In addition, the centrally generated sense of effort is also used to accurately judge the force produced (Carson et al. 2002; McCloskey et al. 1974). The sense of effort is thought to be generated parallel to the motor corollary discharges and passed directly to the somatosensory cortex (Christensen et al. 2007; de Morree et al. 2012; Zenon et al. 2015). There are three interrelated terms related to force matching experiments: sense of effort, sense of heaviness, and sense of force. Sense of effort is considered to be largely derived from central signals, the sense of heaviness from mainly central signals with some evidence of peripheral contributions, and sense of force a combination of central and peripheral signals (Proske and Allen 2019). It remains debatable what information, sensory afferents or centrally generated, is more important when matching force between sides.

In case studies of deafferentation, subjects are still able to match forces between sides, using only the centrally generated sense of effort (Lafargue et al. 2003; Luu et al. 2011). In healthy subjects, the sense of effort has been demonstrated to be important in situations where a side is matched to an eccentrically fatigued side or when forces are matched between sides with the joints at different angles (Cafarelli and Bigland-Ritchie 1979; Carson et al. 2002; Proske et al. 2004). When matching forces in contralateral

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joints at different joint angles, different force levels would be produced, but the level of activation between the muscles was the same (Cafarelli and Bigland-Ritchie 1979). With one side eccentrically fatigued, subjects overshot the reference force when the reference side was the fatigued side and undershoot when it was the matching side (Carson et al. 2002).

An important observation made by Carson et al. (2002) was that the undershoot was absent when the force produced was normalized to the muscle's post eccentric fatigue maximal voluntary contraction. They proposed that the damage to muscle fibers from the eccentric contraction altered the gain of the relationship between the motor command sent to the muscle and the corresponding sense of effort. The same level of effort was perceived with a higher muscle activation and an underestimation of the absolute loads. Additional evidence supporting the hypothesis that these changes come from a central source is that after eccentric exercise, no abnormal function of Golgi tendon organs or muscle spindles was found in cats (Gregory et al. 2002; Gregory et al. 2004). However, Luu et al. (2011) found that after fatigue, deafferented subjects overestimated the load, consistent with sense of effort based predictions, but healthy subjects did not. This prediction was that if maximum muscle force is reduced by 50%, perceived effort should double.

While deafferented subjects have only the sense of effort to match forces, healthy subjects still have afferent proprioceptive information available to them. With recent evidence that tendon vibration to disrupt Ia and Ib afferents (Monjo et al. 2018), it is therefore likely that both pathways contribute to force matching ability in healthy subjects. Also in the study by Luu et al. (2011), healthy subjects performed a matching protocol while the reference side underwent a sustained isometric contraction to fatigue, while matching with the opposite side. Subjects gradually increased the amount of force applied by the matching side but did not overestimate the target to the extent that would be predicted if only the sense of effort were used. Since the matching side did gradually increase the overestimate of the load, this provides evidence for the role of central sense of effort role or reafferent information comparison with the efferent copy.

The purpose of the fatigue protocol was to create a sensory and muscle force imbalance between sides. However, another branch of physiological research argues that group III and IV afferents, responsible for the detection of metabolic, thermal, and ionic changes, also contribute to the conscious perception of effort (e.g., Adreani et al. 1997; Amann 2013; Tucker 2009). Although evidence from pharmacological studies blocking these afferents show that the perceived effort does not change in either static or dynamic exercise (Barbosa et al. 2016; Smith et al. 2003), group III and IV afferents may still have modulating effects on a centrally generated signal during fatiguing contractions. Along with the possible changes in the gain of the signal, absolute force matching undershoot error increases but with a concurrent overshoot in EMG error (Carson et al. 2002), fatigue paradigms may change the way effort is perceived for the same external load.

A muscular paralysis approach, that was also included as part of Luu et al. (2011) experiments, may avoid this altered perception. The curare compounds used act competitively at the neuromuscular junction. A central command can, therefore, still be issued but the muscle simply cannot respond. The magnitude of the central command may not be represented in the activation of the muscle. During the recovery from paralysis subjects underestimated the reference load indicating that absolute load or torque was not the reference signal. A central matching mechanism is therefore proposed. Luu et al. (2011) notes the complex nature of the pharmokinetics of the curare compound. Combining a paralysis approach with EMG measurement may provide additional insight into the signal used to match loads between sides.

More recent evidence shows that sense of effort is associated with muscle activation and both are related to movement related cortical potentials (de Morree et al. 2012). In dynamic movements sense of effort is tied to psychophysiological (EMG) and psychophysical (size, mass, and acceleration) attributes (Mangalam et al. 2018; Toma and Lacquaniti 2016; Waddell and Amazeen 2017; Waddell et al. 2016). Taken together, muscle activation can be used as a measure of both perceived sense of effort as well as representative of the descending efferent command. However, this does not mean there is a simple scaled relationship between muscle activation and movement related cortical potentials. Lastly it should also be noted that although deafferented subjects likely use a centrally generated sense of effort, they still demonstrate poorer psychophysical decisions regarding sensations of heaviness. Muscle activity is not a unique predictor of correct decisions in all deafferent subjects (Sanes and Shadmehr 1995).

The purpose of the present study is to use a contralateral force matching task to examine how the central nervous system accounts for a difference in shoulder torques and loss of afferent sensations from the supraspinatus muscle, caused by a suprascapular nerve block. Our hypotheses are based on the corollary discharge hypothesis, that subjects will match EMG activation rather than torque about the shoulder joint. We hypothesize that: (1) the force matching accuracy will be worse after the nerve block with either side as the reference. In addition, (2) when the unblocked shoulder is the reference side, we hypothesize that subjects will produce less torque with no change in deltoid EMG error on their blocked side. (3) When the blocked shoulder is the reference we hypothesize that subjects will produce more torque and no change in deltoid EMG error on their unblocked side. Lastly, (4) we hypothesize that the any change in error will be dramatically greater for torque error than EMG error with either side as the reference.

Methods

Subjects

Eight subjects initially enrolled in the experiment but data were only utilized from seven subjects (3 males, 4 females, age: 22.4 ± 3.6 years, weight: 67.0 ± 10.0 kg, height: 172 ± 8 cm, all right hand dominant). One subject was excluded to an error when assigning target loads. Exclusion criteria were: (1) previous shoulder or neck injuries, (2) current shoulder or neck pain, (3) humeral elevation range of motion (ROM) less than 135°, (4) previous syncope due to needle insertion, (5) known allergic reaction to anaesthetic, (6) body mass index (BMI) greater than 30 and (7) pregnancy. Subjects were briefed on the purpose and the experimental procedure prior to the start of the experiment and provided informed consent. All subjects were naive to the experimental procedures before participating. The experiment received ethical clearance from the Internal Review Board at the University of Oregon.

Experimental set up

The compression force acting on the forearms of both upper extremities immediately proximal to the ulna styloid processes were recorded using uni-axial load cells (Lebow Products, Troy, MI. Model 3397-50). Force data were sampled at 2000 Hz with custom LabVIEW software (LabVIEW v12.0, National Instruments, Austin, TX). The forearms were flush with the surface of the load cell and loosely secured with custom non-elastic lifting Velcro™ straps. The load cells were angled 20° with respect to vertical. This achieved an estimated 70° humeral elevation when the wrist was flush with the surface of the load cell. The subject's foot positions were marked once the subject was secured to both load cells with the elbows fully extended. Subjects' matched shoulder flexion torques in the sagittal plane by pushing up and compressing the load cells (Fig. 1).

Surface EMG signals from the anterior deltoid, middle deltoid and posterior deltoid were recorded bilaterally with oval, bipolar Ag/AgCl, conductive solid gel electrode pairs (Bio Protech Inc, Wonju, Korea). The skin surface was cleaned with rubbing alcohol. On the anterior deltoid, the electrodes were placed 4 cm below the clavicle on the anterior aspect of the arm; on the middle deltoid, electrodes were placed 2 cm below the acromion process; and on the posterior deltoid, electrodes were placed 2 cm below the lateral border of the scapula spine and angled obliquely. The electrodes were position

Fig. 1 Experimental set up

along the muscle fiber direction with an inter-electrode distance of 2 cm. The ground electrode was fixed over the right patella. EMG data were collected with the Myopac Jr unit (Run Technologies, Mission Viejo, CA) and sampled at 2000 Hz. This unit provided signal amplification (gain = 1000), band pass filtering (10–1000 Hz) and CMMR of 110 dB. Both EMG and force signals were collected simultaneously in LabVIEW.

Subjects were presented with three force targets with the upper extremities secured to the load cells. Each target force was repeated four times, for a total of 12 trials. The force targets were presented in a randomized order. The force targets were calculated as 120%, 140% and 160% of baseline torque using anthropometric equations from Winter (2005). Baseline torque represents the torque at the shoulder due to the weight of the arm at 70° humeral elevation.

Vision of the environment during the protocol was occluded with a head mounted display (Z800, eMagine, Bellevue, WA) with modifications to prevent influence of external light sources. The display provided visual guidance (see "Contralateral force matching protocol") to targets during the force matching protocol, while blocking all vision of the upper extremities.

Maximal voluntary contractions

Prior to the contralateral force matching protocol, a series of 5 s maximal voluntary contractions (MVCs) were taken. Subjects were verbally instructed on how to perform MVCs and a practice attempt was given prior to recording. MVCs were recorded for external rotation on the right/blocked side only. The shoulder was slightly adducted and elbow flexed to 90°. A towel was placed under the arm to help prevent the subject from abducting their arm during the MVC. If the arm did abduct, the towel would fall to the ground and the MVC was repeated. MVCs were then recorded for both sides shoulder flexion at 70° humeral elevation in the sagittal plane (the testing position for the force matching protocol). Subjects were given two attempts for each MVC position with a 2-min rest between each attempt. If the MVC was performed incorrectly, feedback was given to the subject and a third MVC taken. The first 2.5 s and the last 1 s of force data were trimmed. The mean of the remaining 1.5 s was averaged and used to represent the subject's MVC. The MVC with the largest force was considered for further analysis.

Contralateral force matching protocol

Following anthropometric measures (mass and arm length), MVCs before the suprascapular nerve block were recorded. The subjects were positioned so that they stood with their forearms flush with the surface of the load cells and arms parallel to each other in the sagittal plane. The load cells were positioned shoulder width apart. A custom non-elastic strap was used to loosely secure the wrists to the load cells. The subject was presented with a black screen with 2 white horizontal lines across the middle two-thirds of the head mounted display which would represent the target with a 2 N tolerance. A dynamic read line represented the force applied to the reference load cell. The target represented randomly changed to one of the three force targets after each trial and the subjects was provided with no knowledge of their results.

Subjects were verbally instructed on how to perform the protocol. Subjects were asked to maintain both their arms in the 'thumbs up' position. Subjects were instructed to maintain the red line between the two white lines to become accustomed to the target force. The acclimation time was 1.5 s. After the 1.5 s, the program would initiate a 'find target' command and the subject would attempt to reproduce the reference force target with the contralateral arm while maintaining the reference force with the other. Visual feedback for the force generated by the reference arm remained on the heads up display. When the subject felt they had replicated the force, they verbally signaled the researcher to press the trigger and record the force level. A 'relax' cue was initiated by the computer and a 15 s count down timer appeared on the display, which indicated the time until the next trial began. Prior to the first instance of the force matching protocol six practice trials were given. During the practice trials, the researcher provided verbal feedback and answered any questions. The force output of these trials was visually inspected to ensure that the subject had understood the instructions before recording the experimental trials. No feedback was given to subjects regarding their performance.

The left arm was used as the reference for the first set of 12 trials. The right arm was used as the reference for the second set of 12 trials. Following the collection of both sets of 12 trials, the nerve block was performed on the right side. External rotation was then tested until a 50% drop in MVC was observed (see 'Suprascapular Nerve Block Procedure').

Following this, 2 maximal shoulder flexion contractions were recorded at 70° humeral elevation in the sagittal plane for the right/blocked side. The force matching protocol was then repeated. Upon completion of the protocol, 2 maximal shoulder flexion contractions were recorded at 70° humeral elevation in the sagittal plane for the right/blocked side and another 2 external rotation maximal contractions measurements. These contractions were used to ensure that the block was still effective at the conclusion of testing.

Suprascapular nerve block procedure

A suprascapular nerve block was performed by a board certified anesthesiologist. The subject was seated for the procedure with the head flex slightly to the contralateral side. Ultrasound imaging was used to visualize the scapula notch where the suprascapular nerve travels. The ultrasound gel served as a conductive medium and surface preparation. A 3.5 inch 23 gauge quincke needle was advanced toward the scapular notch in a medial to lateral direction using an in plane technique. The advancing needle was observed on the ultrasound until it reached the scapula notch. At this point the lidocaine and epinephrine (1.5%, 1:200,000, 5 ml) was injected. The needle was removed and the subject was allowed to remain seated for 5 min.

The external rotation MVC was used to determine whether the block had been effective and the post-block force matching protocol could proceed. The criteria to proceed with testing was a 50% reduction in external rotation for two consecutive external rotation MVCs. Five minutes after the block was completed, the subject's external rotation was tested. If the external rotation force was still above 50% MVC, the subject was retested after another 5 min. From that point on, the subject's external rotation was tested every 2 min until 2 consecutive external rotation maximal contractions measurements were below 50% MVC.

EMG normalization

EMG amplitude for each part of the deltoid was normalized to its highest recorded amplitude during the pre-block MVCs at 70° humeral elevation for each side. The raw EMG from each part of the deltoid (anterior, middle and posterior) was smoothed using a 300 ms RMS window. The first 2.5 s and the last 1 s was trimmed. The mean of the remaining 1.5 s was used for normalization. EMG recorded during the trial was also smoothed using a 300 ms RMS window.

Data analysis

Forces from the load cells were converted to torque (Nm). The torque measured from each load cell was added to baseline torque for the arm at 70° humeral elevation. The total torque for the reference side was termed reference torque and the total torque for the matching side was termed matching torque at the time the trigger was pressed. Likewise, EMG for the reference side was termed reference EMG and the matching side was termed matching EMG at the time the trigger was pressed.

Error was calculated as a percent of the reference for each trial.

$$\%\ \text{Error} = \frac{T_{\text{match}} - T_{\text{ref}}}{T_{\text{ref}}} \times 100,$$

where T is torque, anterior deltoid EMG, middle deltoid EMG or posterior deltoid EMG. When torque is the variable, a positive value indicates that subject overestimated the target and a negative an underestimate of the target. When EMG is the variable, a positive value indicates that the matching muscle was more active than the reference muscle and a negative value indicates the matching muscle was less active than the reference muscle. The average error of the 4 trials at each load were calculated when the left and right were the reference. To assess the accuracy of subjects the root mean square (RMS) error was calculated and normalized to baseline torque.

$$\text{RMS error} = \sqrt{\sum \left(\frac{\left(T_{\text{match}} - T_{\text{ref}}\right)}{b} \times 100 \right)^2 / n},$$

where T is torque, b is baseline torque of the arm at 70° humeral elevation and n is the number of trials. Again, the average error of the four trials at each load were calculated when the left and right were the reference.

Statistical analysis

Statistical analysis was performed using SPSS version 22.0 (SPSS Inc., Chicago, IL). One subject was removed from all analysis due to an error during target load assignment reducing the total subjects to seven as reported in subject demographics. To quantify the effects of the suprascapular nerve block on maximal voluntary contraction forces, a paired *t* test was conducted on maximal external rotation and shoulder flexion, before and after the block. A third paired *t* test was conducted to compare the sagittal flexion MVC between left and right sides.

The following statistical analyses were conducted first with the left/unblocked side as the reference side and then repeated with the right/blocked side as the reference. If the assumption sphericity was violated a Greenhouse–Geisser adjustment was performed. To test the first hypothesis on force matching accuracy, a two-way repeated measures ANOVA was used to assess the effect of condition (non-blocked vs blocked) and load (120%, 140%, and 160% of baseline torque) on torque root mean square (RMS) error normalized to baseline torque.

To test our second and third hypotheses on the direction of error (undershoot and overshoot) before and after the nerve block, a 2-way repeated measures ANOVA was used to assess the effects of the condition (non-blocked vs blocked) and load (120%, 140%, and 160% of baseline torque) on each dependent variable (torque error, anterior deltoid EMG error, middle deltoid EMG error and posterior deltoid EMG error).

To test our forth hypothesis on the magnitude of the change in error due to the nerve block, a two-way repeated measures ANOVA was used to assess the effect of each error parameter (torque, anterior EMG, middle EMG, and posterior EMG) and load (120%, 140%, and 160% of baseline torque) on the change in error. Change in error is calculated by subtracting the pre-block error from the post-block error. In the case of a significant main effect for change in error, a simple contrast comparison between torque error change and EMG error change (anterior deltoid, middle deltoid and posterior deltoid) was planned. Effect sizes (ES) for the differences between pre-block and post-block error are reported using Cohen's *d*.

In addition to the above statistical analyses, a Wilcoxon signed-rank test was run for each dependent variable (torque error, anterior EMG error, middle EMG error, posterior EMG error and force mating accuracy) to assess differences before and after the block. This is done as statistical tests of normality do not provide certainty of the normal distribution assumption in small sample sizes. The Wilcoxon signed-rank tests will help confirm effects observed from the ANOVA and help account for potential Type I error.

Results

Maximal voluntary contraction

The suprascapular significantly reduced the maximal voluntary contractions (MVC). There was a 60% reduction in external rotation MVC after the nerve block (M = 30.0 N, SD = 29.9 N) compared to before (M = 74.2 N, SD = 14.4 N), p <0.001, and a 52% reduction is sagittal plan flexion MVC after the nerve block (M = 37.6, SD = 23.3 N) compare to before the block (M = 78.7 N, SD = 29.9 N), $p < 0.001$ (Fig. 2).

Force matching

In almost all cases, the mean of the four viable trials were calculated. However, in some instances subjects did not perform the protocol correctly (e.g., relaxing during acclimation or the trigger did not register) resulting in 3 viable trials in 4/48 (8%) cases and 2 viable trials in 2/48 (4%) cases. In these cases, the means of the viable trials were calculated.

Left/unblocked side as reference

Torque % error

The interaction between the condition (blocked vs non-block) and load was significant, p = 0.02. Follow up simple effects for condition at each load only demonstrated significance at 160% where subjects significantly undershot the blocked (M = − 10.3%, SD = 10.5%) compared to non-block (M = 1.3%, SD = 15.3%) condition, p = 0.02 (Fig. 3). The Wilcoxon signed-rank test indicated that torque error was undershooting the pre-block error (p = 0.016, ES = 0.5).

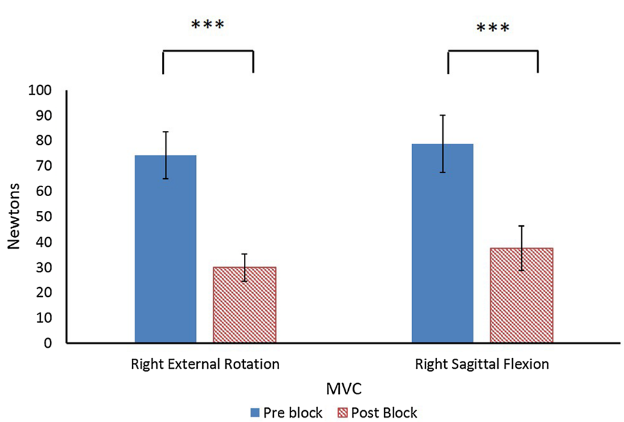

Fig. 2 Changes in maximal voluntary contraction after the nerve block. ***$p < 0.001$

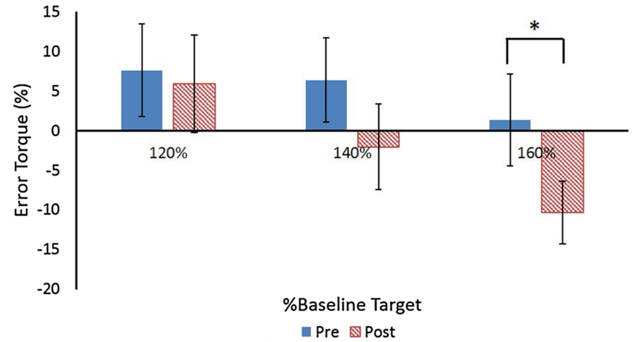

Fig. 3 Torque error (%) before and after the suprascapular nerve block with the left/unblocked side as the reference. *$p < 0.05$

Anterior deltoid EMG % error

The interaction between the condition and load was not significant (p = 0.68). The main effect for condition was significant (p = 0.039). The matching anterior deltoid demonstrated higher activation error in the blocked condition (M = 71.8%, SD = 24.7%) than the non-blocked condition (M = 19.3%, SD = 17.3%) (Fig. 4). The main effect for load was significant (p = 0.02). The Wilcoxon signed-ranks test indicated that anterior deltoid EMG error was more positive after the block ($p < 0.001$, ES = 0.96).

Middle deltoid % EMG error

The interaction between the condition and load was not significant (p = 0.55). The main effect for load was not significant (p = 0.07). The main effect for condition was significant (p = 0.038). The matching middle deltoid demonstrated higher activation error in the blocked condition (M = 133.0%, SD = 51.7%) than the non-blocked condition (M = 10.0%, SD = 15.0%) (Fig. 4). The Wilcoxon signed-ranks test indicated that middle deltoid EMG error was more positive after the block ($p < 0.001$, ES = 1.4).

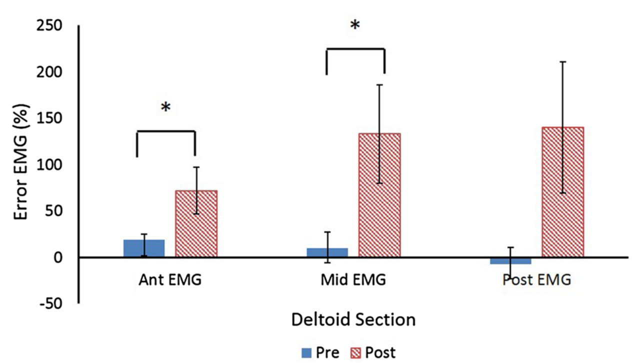

Fig. 4 EMG error (%) before and after the suprascapular nerve block with the left/unblocked side as the reference. *$p < 0.05$

Posterior deltoid % EMG error

The interaction between the condition and load was not significant ($p = 0.64$). The main effect for load was not significant ($p = 0.19$). The main effect for condition was not significant ($p = 0.05$) (Fig. 4). The Wilcoxon signed-rank test indicated that posterior deltoid EMG error was more positive the block ($p < 0.001$, ES = 1.34).

Change in error

The interaction between the error parameters and load was not significant ($p = 0.63$). The main effect for load was not significant ($p = 0.56$). The main effect for error parameter was significant ($p = 0.03$). The follow up simple contrast demonstrated a significantly larger change in anterior deltoid EMG error ($M = 52.5\%$, SD = 19.9%, $p = 0.01$); middle deltoid EMG error ($M = 123.0\%$, SD = 46.4%, $p = 0.025$); and posterior deltoid EMG error ($M = 147.2\%$, SD = 59.1%, $p = 0.04$) than the change in torque error ($M = -7.3\%$, SD = 4.2%) (Fig. 5).

Force matching accuracy

The interaction between the condition and load was not significant ($p = 0.96$). The main effect for load was not significant ($p = 0.43$) and the main effect for condition was not significant ($p = 0.78$) (Fig. 6). The Wilcoxon signed-rank test indicated there was no change in accuracy after the block, $p = 0.43$.

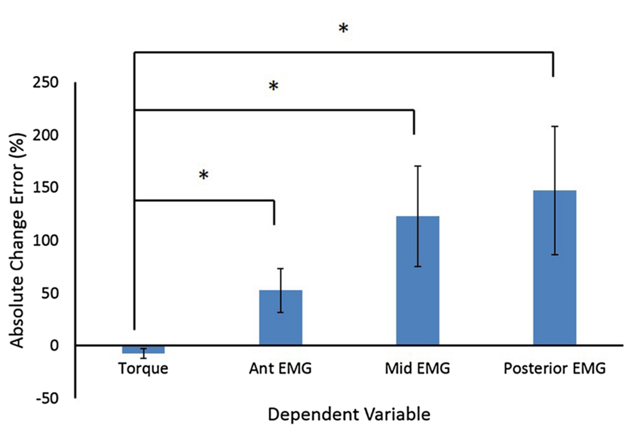

Fig. 5 Comparison of the absolute change in error (%) from before to after the suprascapular nerve block between torque and EMG amplitude with the left/unblocked side as the reference. *$p < 0.05$

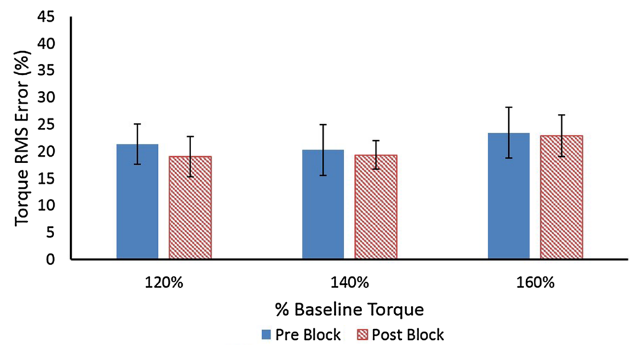

Fig. 6 RMS torque error (normalized to baseline torque) before and after the suprascapular nerve block with the left/unblocked side as the reference. *$p < 0.05$

Right/blocked side as reference

Torque % error

The interaction between the condition and load was not significant ($p = 0.064$). The main effect for load was not significant ($p = 0.38$). The main effect for condition was significant ($p = 0.02$). Subjects had a significantly higher overshoot error in the blocked condition ($M = 20.6\%$, SD = 4.9%) compared to the non-blocked condition ($M = 6.1\%$, SD = 1.2%) (Fig. 7). The Wilcoxon signed-rank test indicated that torque error was more positive than the pre-block error ($p = 0.003$, ES = 1.06).

Anterior deltoid EMG % error

The interaction between the condition and load was not significant ($p = 0.67$). The main effect for load was not significant ($p = 0.28$). The main effect for condition was significant ($p = 0.02$). The matching anterior deltoid demonstrated higher activation error in the blocked condition ($M = -44.4\%$, SD = 12.4%) than in the non-blocked

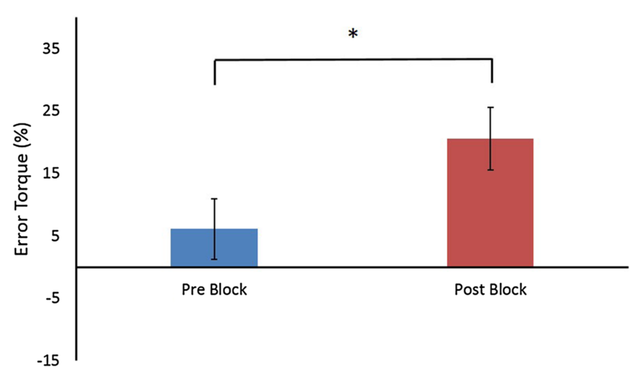

Fig. 7 Torque error (%) before and after the suprascapular nerve block with the right/blocked side as the reference. *$p < 0.05$

condition (M = − 12.4%, SD = 12.0%) (Fig. 8). The Wilcoxon signed-rank test indicated that anterior deltoid EMG error was more negative after the block ($p < 0.001$, ES = 0.88).

Middle deltoid EMG % error

The interaction between the condition and load was not significant (p = 0.49). The main effect for condition was significant ($p = 0.01$). The matching middle deltoid demonstrated higher activation error in the blocked condition (M = − 68.7%, SD = 22.5%) than the non-blocked condition (M = 13.9%, SD = 14.4%) (Fig. 8). The main effect for load was significant ($p = 0.01$). The Wilcoxon signed-rank test indicated that middle deltoid EMG error was more negative after the block ($p < 0.001$, ES = 0.9).

Posterior deltoid EMG % error

The interaction between the condition and load was not significant (p = 0.79). The main effect for condition was significant ($p = 0.005$). The matching poster deltoid demonstrated higher activation error in the blocked condition (M = − 70.0%, SD = 35.0%) than the non-blocked condition (M = 7.2%, SD = 23.4%) (Fig. 8). The main effect for load was not significant ($p = 0.058$). The Wilcoxon signed-rank test indicated that posterior deltoid EMG error was more negative after the block ($p < 0.001$, 0.83).

Change in error

The interaction between the error parameter and load was not significant ($p = 0.56$). The main effect for load was not significant (p = 0.89). The main effect for error parameter was significant ($p < 0.001$). The follow up simple contrast found a significantly larger change in anterior deltoid EMG error (M = − 32.0%, SD = 10.3%, p <0.001); middle deltoid EMG error (M = − 54.8%, SD = 15.26%, p = 0.001); and posterior deltoid EMG error (M = − 77.2%, SD = 18.0%, p =0.001) than the change in torque error (M = 14.5%, SD = 4.7%) (Fig. 9).

Force matching accuracy

The interaction between the condition and load was not significant ($p = 0.83$). The main effect for load was not significant ($p = 0.39$) and the main effect for condition was not significant (p = 0.08) (Fig. 10). The Wilcoxon signed-ranks test indicated accuracy was worse after the block, $p = 0.006$.

Discussion

The purpose of the study was to determine how the central nervous system accounts for a muscle force and sensory disruption between sides in a contralateral force matching task. In our hypotheses, we expected EMG amplitude to be matched but torque error to increase after the block

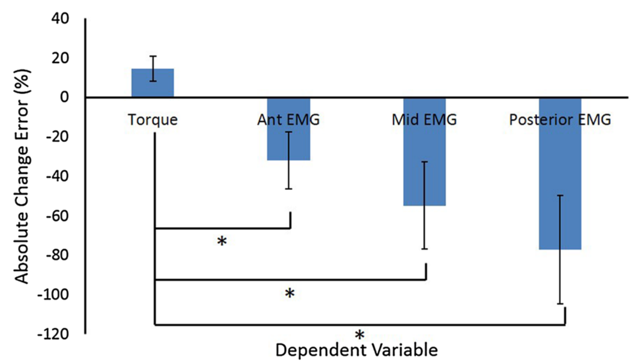

Fig. 9 Comparison of the absolute change in error (%) from before to after the suprascapular nerve block between torque and EMG amplitude w ith the right/blocked side as the reference. *$p < 0.01$

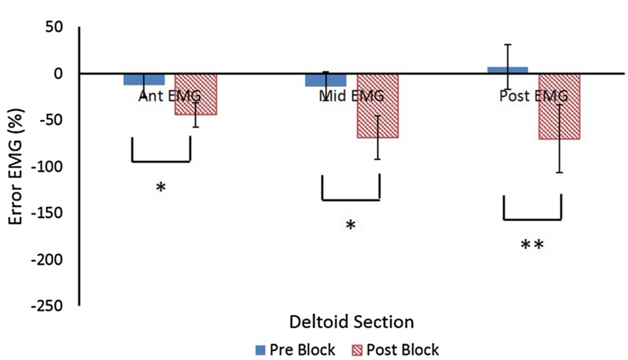

Fig. 8 EMG error (%) before and after the suprascapular nerve block with the right/blocked side as the reference. *$p < 0.05$, **$p < 0.01$

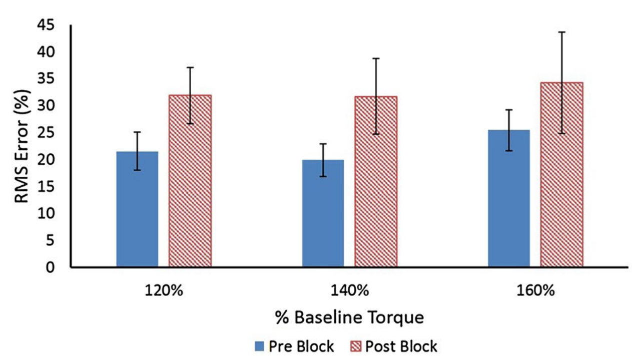

Fig. 10 RMS torque error (normalized to baseline torque) before and after the suprascapular nerve block with the right/blocked side as the reference

(Cafarelli and Bigland-Ritchie 1979; Carson et al. 2002). Our data shows the opposite of this.

For the left side as reference, subjects only significantly undershot the reference when the load target was 160% of baseline torque after the suprascapular nerve block (Fig. 3). A much greater change due to block is observed in EMG matching errors across the deltoid muscle (Fig. 4). In fact, the change score on EMG was seven times greater for the anterior, 17 times greater for the middle and 20 times greater for the posterior deltoid. Even though some effect is observed for torque at the 160% baseline load, the effect is relatively small when considering the EMG changes. This is indeed the case when we calculate the effect sizes.. The effect sizes for the anterior, middle and posterior EMG % error is 1.9, 2.8, and 2.7 times greater than observed for torque % error. The same effect but in the opposite direction is observed when the right/blocked side is the reference. However, the data when the right side is the reference needs to be interpreted more cautiously.

In this case, the change score for the anterior, middle and posterior was 2, 4 and 5 times greater than the change in torque error respectively. The reason for the magnitude difference in error change in EMG between the left and right sides is due to the normalization to the reference value. For the left, the reference value for EMG and torque remains the same before and after the block while for the right, the EMG reference dramatically increases after the block (McCully et al. 2007) but the torque reference remains the same. The method of calculating the EMG % error underestimates the effect of the block on EMG. This is the reason that effect sizes are similar between torque and EMG % error when the right side is the reference. While torque % error does increase after the block, EMG % error is substantially larger.

Our non-parametric (Wilcoxon signed-rank) tests were conducted to help account for the small sample size in the study. When the left side was the reference, there was a consistent response for each dependent variable to the block. Torque % error was more negative while EMG % error was more positive for all portions of the deltoid muscle after the block. The opposite trend was observed when the right side was the reference, torque % error was more positive and EMG % error was more negative.

Unilateral fatigue force matching studies demonstrate that subjects perceived more effort for the same load or produce unequal forces between sides even though they perceive the same level of effort (Jones and Hunter 1983; McCloskey et al. 1974). Since the function of muscle spindles and GTOs should not have been affected by the fatigue protocol (Gregory et al. 2002), the sense of effort appears to be prioritized and arising from motor corollary discharges. In a biceps brachii fatigue force matching model (Jones and Hunter 1983), EMG amplitude from the fatigued side was able to predict the force produced by the matching side. Using an eccentric fatigue model, Carson et al. (2002) observed that subjects significantly undershot the reference load when the unfatigued side was the reference. They also saw an increase in EMG amplitude in the matching side even though it underestimated the target. They hypothesized that the eccentric contractions (through muscle fiber damage) modified the gain between the descending motor command and the perceived amount of effort. This also alters the force-EMG relationship within the muscle. Peripheral nerve blocks can alter motor cortex excitability but not spinal cord excitability (Brasil-Neto et al. 1993). Although the force results from that study and the torque results from the present study occur in the same direction, the effect is smaller in the present study and small in comparison to the changes we observed in EMG when the left side is the reference. If EMG and associated sense of effort were the primary sources of matching information, it would have led to a dramatic underestimation of force when the left was the reference and an overestimation when the right was the reference. Instead we see a dramatic change in motor output by increases in deltoid EMG but no change in force matching error. Our force error observations are very similar to those observed the continuous isometric force matching by Luu et al. (2011). The subjects' error moved in the direction with hypothesized using sense of effort feedback, but not to the extent that would be expected if only sense of effort was used.

Even though subjects' torque % error direction changed in accordance with our hypothesis—undershot when the left/unblocked was reference and over shot when right/blocked was reference—their overall accuracy in the performance of the force matching task did not statistically change. This further indicates that torque, and not motor neural input to a muscle or sense of effort, is being used to match forces during this task. It should be noted that when the right/blocked side was the reference, accuracy was worse but not statistically so.

In the absence of vision, the CNS can make use of afferent feedback, motor command or both to estimate the forces. Possible sources of afferent information include: muscle spindles and GTOs within the deltoid muscle; sense of effort in parallel with the motor command and efferent copy; cutaneous receptors located at the wrists; and possibly the lack of information from the supraspinatus muscle on the blocked side. Barring the cutaneous receptors—because error was not greater after the block and pressure on the skin would remain almost the same—the suprascapular nerve block would have cause significant disruption to all other sources of afferent information during the task.

The deltoid muscles generated vastly different tensions and the sense of effort would have been far greater on the right side, as evident from the substantially greater activation. GTO firing rates would have been different between the deltoids because of this tension difference and any GTO impulses would be absent on the blocked side. For muscle spindles, it was previously observed that firing rates do increase during an isometric contraction and it is was hypothesized that these signals are filtered out (McCloskey et al. 1983). In a more recent study on bilateral force matching at the elbow using passive and active tendon vibration of the biceps brachii tendon, subjects underestimated the load during passive vibration and overestimated the load during active vibration (Monjo et al. 2018). This appears to support a centrally generated signal for force matching. However, in the same study they performed a unilateral task and asked subjects to quantify effort according to the Borg CR 10 scale. In both active and passive vibration subjects reported less effort than without the tendon vibration. In this case EMG activation of both the biceps and brachioradialis did not change. If only the central signal were used, the amount of effort would not have changed. In our study muscle spindles were not affected by the block but there would have been a difference in the nature of the response between the muscle spindles of the musculature for the blocked and unblocked sides. Muscle spindle response is completely absent from the blocked supraspinatus muscle but could be elevated in the deltoid on the blocked side.

The last afferent consideration is cutaneous receptors, which should not be affected by the suprascapular nerve block. Removal of cutaneous afferents can result in either underestimation (Jones and Piateski 2006) or overestimation (Monzee et al. 2003) of forces depending on the on whether feedback is attenuated or completely removed. In our study, subjects tended undershot the load with the left/unblocked as the reference and significantly overshot by 21% of reference when the right/blocked was the reference. This overshoot is still well below what would be expected if only sense of effort was being utilized to match the loads. Cutaneous feedback would still have been able to determine a difference between sides. The possibility exists that it may not be sensitive enough or be weighted as heavily by the CNS to prevent error moving in the direction consistent with sense of effort matching but could prevent excessive errors. However, in the previously mentioned fatigue model studies, cutaneous information also remained intact but did not prevent the torque mismatch between sides. This points to sense of effort being upregulated or emphasized after fatigue over sensory afferent information (Carson et al. 2002; Luu et al. 2011; Proske et al. 2004).

Another potential option for the CNS system is to use total afferent feedback or total efferent outflow. This would require interpretation of the total tension in all muscles or the total strength of the motor command (sense of effort) to the contracting muscles. Using this paradigm when the left side (unblocked) is the reference, the summed tension of the left supraspinatus and deltoid would need to be the same as the summed tension from the right deltoid and paralyzed supraspinatus. Summed tension could be represented as number of type Ib afferents per second from a muscle group. The same would apply to sense of effort. Error in the system would increase since GTO and muscle spindle firing in the deltoid may not correlate exactly with the supraspinatus. This may also be the reason why subjects were successful at completing the force matching task after the block.

In this study, we have assumed that there is at least a good correlation between deltoid EMG and sense of effort and that as EMG activation increases, there is a concurrent increase in sense of effort. It is not certain whether central drive to the blocked supraspinatus remained unaffected and could not reach the muscle due to the block, or if it was reduced, or completely absent. Follow up research could employ the suprascapular nerve block, force matching, and transcranial magnetic stimulation to evaluate assumptions in this study and would provide further insight into the signal used to match torques between sides. While the differences observed in this study are large, the sample size is still reasonably small. Additional studies with larger sample sizes would be important to ensure the results are repeated.

Conclusion

Our results do not support a central corollary discharge model for matching forces between left and right sides. Subjects are able to successfully match forces between sides even after the torque producing capacity of one side is reduced by more than 50% and afferent information from the supraspinatus is lost unilaterally. The mismatch between deltoid activation is far greater than torque error. This study provides evidence that afferent information is more heavily weighted when matching contralateral forces. The mechanism is not certain since the afferent information is also mismatched between sides. It is possible that the central nervous system is able to rapidly recalibrate and reinterpret incoming afferent information.

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Compliance with ethical standards

Conflict of interest The authors have no conflicts of interest to disclose.

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