Shoulder muscle activation strategies differ when lifting or lowering a load

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Abstract

Purpose Lowering a load could be associated with abnormal shoulder and scapular motion. We tested the hypothesis that lowering a load involves different shoulder muscle coordination strategies compared to lifting a load.

Methods EMG activity of 13 muscles was recorded in 30 healthy volunteers who lifted and lowered a 6, 12 or 18 kg box between three shelves. Kinematics, EMG levels and muscle synergies, extracted using nonnegative matrix factorization, were analyzed.

Results We found greater muscle activity level during lowering in four muscles (+1-2% MVC in anterior deltoid, biceps brachii, serratus anterior and pectoralis major). The movements were performed faster during lifting (18.2 vs. 15.9 cm/s) but with similar hand paths and segment kinematics. The number of synergies was the same in both tasks. Two synergies were identified in ~75% of subjects, and one synergy in the others. Synergy #1 mainly activated prime movers' muscles, while synergy #2 coactivated several antagonist muscles. Synergies structure was similar between lifting and lowering (Pearson's r ≈ 0.9 for synergy #1 and 0.7–08 for synergy #2). Synergy #2 was more activated during lowering and explained the greater activity observed in anterior deltoid, serratus anterior and pectoralis.

Conclusions Lowering a load was associated with an increased activation of a "multiple antagonists" synergy in the subjects with the greatest motor control complexity. The others subjects cocontracted all shoulder muscles as a unit in both conditions. These interindividual differences should be investigated in the occurrence of shoulder musculoskeletal disorders.

Keywords. electromyography –eccentric– ergonomics –shoulder injuries – musculoskeletal– intramuscular

Abbreviations

BB: *biceps brachii*

DeltA: *deltoid* (anterior part)

DeltM: *deltoid* (middle part)

DeltP: *deltoid* (posterior part) of the deltoid

LD: *latissimus dorsi*

MVC: maximal voluntary contraction

Pect: *pectoralis major*

SerrA: s*erratus anterior*

TB: *triceps brachii*

TraS: *trapezius* (superior part)

TraL: *trapezius* (lower part)

VAF: variance accounted for

Introduction

The shoulder complex is subject to high risks of musculoskeletal injuries, such as tendon tears, impingements or joint subluxation, especially in sports and manual occupations in which extreme joint positions (e.g., overhead tasks) and high muscle loads are common (Ludewig and Lawrence 2017). The shoulder complex is put in motion by a large and redundant set of muscles (Ebaugh and Finley 2017; Ebaugh and Spinelli 2010), and an inappropriate coordination of these muscles may play a role in the occurrence of shoulder musculoskeletal injuries (Magarey and Jones 2003; Labriola et al. 2005; Madeleine 2010). Pain and muscles fatigue associated with repetitive use have been pointed out as potential sources of alteration in shoulder muscle coordination (Nieminen et al. 1995; Moraes et al. 2008). However, lengthening contractions might contribute to altering muscle coordination as well.

Lengthening contractions are commonly encountered in sports or occupational activities when resisting or lowering a load, or in the deceleration phase of throwing, for example. One major difference with shortening or isometric contractions is that muscles are intrinsically stronger during lengthening contractions, and that partly explains why the muscle activity needed to perform such actions is generally lower (Kronberg and Brostrom 1995; Hawkes et al. 2012a; Gaudet et al. 2018). In addition, inhibitory and excitatory influences on the motoneurons associated with lengthening contractions are distinct from those observed during shortening or isometric contractions (Duchateau and Enoka 2016). Lengthening contractions have been associated with poor force control, variable motor output, and altered and variable movement kinematics (Borstad and Ludewig 2002; Christou and Carlton 2002; Duchateau and Enoka 2008). Lowering the arm was associated with abnormal scapular kinematics in individuals with shoulder pain (Rossi et al. 2018) and lowering a load has been reported as being more painful than raising it in individuals with impingement syndrome (Borstad and Ludewig 2002), suggesting that lengthening contractions might bring specific biomechanical constraints to the shoulder complex.

Surprisingly, however, few studies have specifically investigated the differences in coordination of the shoulder complex muscles between movements involving shortening and lengthening contractions such as when lifting or lowering a load (Ebaugh and Spinelli 2010). The coordination of the shoulder complex muscles is characterized by parallel changes in the level of activity of all the muscles of this system (Kronberg and Brostrom 1995; Hawkes et al. 2012b; Hawkes et al.

2012a) and by a high level of antagonist coactivation (Blache et al. 2015). As for the timing of muscle activity, a previous study (Hawkes et al. 2012a) reported different behaviors only for the elbow flexors during a weight lifting task, i.e., the peak of activation of these muscles occurred earlier compared to other muscles (i.e., deltoids, adductor of the shoulder and rotator cuff muscles). Regarding the relative level of activity of the shoulder muscles, we found no study that explicitly compared the lifting and lowering movements.

Previous studies used correlations, ratios or the common areas between the activities of pairs of muscles to investigate the synergies involved in shoulder movements (Cools et al. 2007; Faria et al. 2009; Hawkes et al. 2012a). Here a synergy is defined as a group of muscles activated in a fixed balance (e.g., Tresch and Jarc, 2009). However, these techniques only compare one pair of muscles at a time, while muscles are commonly organized into functional groups of more than two muscles (d'Avella et al. 2008; Roh et al. 2012). Linear decomposition methods such as nonnegative matrix factorization may provide an interesting alternative in this regard (Hug 2011; Safavynia et al. 2011). These methods were effective in identifying the covariations in structure underlying multiple muscle activations (i.e., the functional muscle groups) in various motor tasks, including shoulder movements (Roh et al. 2012). However, no previous studies focused on the synergies associated with lifting and lowering using nonnegative matrix factorization. Using deafferentation in frogs or by inducing temporary pain in humans, previous studies suggested that muscle synergies may be tuned by afferent feedback (Cheung et al. 2005; Muceli et al. 2014). Therefore, it can be hypothesized that the complex afferent flow associated with lengthening contractions (Duchateau and Enoka 2008) could affect the structure of muscle synergies during shoulder movements.

In the present study, we used the synergy analysis to specifically investigate the muscle activation strategies associated with the tasks of lifting and lowering a loaded box with the arms, which involve mainly shortening and lengthening contractions, respectively. Given the biomechanical and neurophysiological differences between these two modes of contractions, we hypothesized that lifting and lowering a load would be associated with different muscle synergies and different activation of these synergies.

Method

Subjects

Thirty subjects (16 men and 14 women) aged of 20-30 years (24.0 ± 3.6) participated in the study after signing informed consent forms. The protocol was approved by the University Ethics Committee (Nº11-068-CERSS-D). None of the participants were ever diagnosed with musculoskeletal disorders of the upper limbs or reported significant disability related to their upper extremity (Disabilities of the Arm, Shoulder and Hand scores (Hudak et al. 1996) > 23) or their back (Quebec Back Pain Disability Scale score (Kopec et al. 1996) < 3). Readiness for physical activity was confirmed in all participants (Physical Activity Readiness Questionnaire (Thomas et al. 1992)).

Tasks description

The main task consisted in lifting or lowering a 6, 12 or 18 kg box from one shelf to another. Women performed the tests with only the 6 and 12 kg boxes. The shelves were placed at three different heights, adjusted to the hip, shoulder and eye levels of each participant. The distances were 47.4±8.3 cm and 71.0±15.4 cm between the lower and middle shelves and between the lower and upper shelves, respectively. The dimensions of the boxes were $0.345 \times 0.395 \times 0.08$ m (length, width and height). Subjects were standing in front of the shelves at their preferred horizontal distance, with their feet parallel and naturally spaced. They were instructed to hold the box using the left and right tubular handles. No specific instructions were given regarding the speed at which they should lift or lower the boxes, or the technique that they should use. Three trials were randomly performed for each height, weight and direction (lifting or lowering) condition with a 30 s rest period in-between and 3 minutes between conditions.

Prior to the main task, each subject performed a series of maximal isometric voluntary contractions (MVCs) in which the EMG were recorded. The protocol, which consisted in a series of manual testing, has been presented in detail in Dal Maso et al. (2016). Subjects rested for a period of 5 min after these MVCs.

Data recording and analysis

Surface EMGs were recorded at a sampling frequency of 2000 Hz from 10 shoulder muscles. The task being symmetric, we only recorded the muscles on the right side. Wireless surface electrodes (bipolar; Trigno™ IM, 20-450 Hz bandwidth; 16 bits; Delsys Inc., Boston, MA) were placed over the anterior (DeltA), middle (DeltM) and posterior (DeltP) parts of the deltoid, the long heads of the *biceps brachii* (BB) and *triceps brachii* (TB), the superior (TraS) and lower (TraL) parts of the *trapezius*, the s*erratus anterior* (SerrA), the *pectoralis major* (sternal portion – Pect) and the *latissimus dorsi* (LD) according to the Surface EMG for Non-Invasive Assessment of Muscles (SENIAM project, *www.seniam.org*) recommendations. Prior to electrode application, the skin was shaved and cleaned with alcohol to minimize impedance. Because of their possible role in gleno-humeral stability (Blache and Begon 2017) deep muscles were additionally recorded in a subset of 10 subjects. These EMGs (Trigno™ Spring contact adapter, 20-450 Hz bandwidth; 16 bits; Delsys Inc., Boston, MA) were recorded from the *supraspinatus* (SupS), *infraspinatus* (InfS), and lower *subscapularis* (SubS) using fine-wire intramuscular electrodes (30 mm, 27 gauge; CareFusion) inserted in a single hypodermic needle into the muscles. The subjects in whom 10 muscles (surface EMG only; N = 30) were recorded correspond in the following to **group #1** and the subset of subjects in whom 13 muscles (surface and intramuscular EMG; N = 10) corresponds to **group #2**.

Surface EMG signals were band-pass filtered (4th-order Butterworth) between 20 and 400 Hz and intra-muscular EMG signals between 20 and 1000 Hz. Electrical noise was removed using a notch filter at 60 ± 0.3 Hz. Raw EMG signals were then demeaned to nullify possible bias in the EMG amplifiers. Integrals of the rectified EMG signals were computed over a 35-ms window (trapezoid method) and shifted with each EMG sample interval to obtain the EMG profiles (iEMG).

The resultant forces applied to the right handle were simultaneously recorded (custom-made 6-dof force sensor by Sensix, France) at a sampling frequency of 1 kHz. EMGs were analysed only when external forces (as measured by the right handle) were applied to the box. More precisely onset and offset of each trial were determined as when the norm of the forces reached and returned to the baseline force, computed as the mean ± 2×SD of the background forces.

Then iEMGs for each muscle were normalized by the peak iEMG values extracted from the MVC tests. iEMGs were finally time-interpolated to 200 time samples for each trial using the spline method. Mean EMG activity corresponded to the arithmetic mean of the normalized iEMG over the period of analysis previously described and over all conditions of height and weight.

Kinematic data were acquired with an 18-camera Vicon motion analysis system (Oxford Metrics Ltd, Oxford, UK). These data were available for 17 subjects only. Thirty-five markers were placed on participants' skin over the pelvis (4), trunk (6), clavicle (5), scapula (9), upper arm (7) and forearm (4) (Bouffard et al. 2019). The kinematic data were low-pass filtered with a cut-off frequency of 10 Hz (2nd order Butterworth). The markers used for the present study and the definition of the trunk, arm, forearm and hand segments are presented in Table 1.

Joint angles

Elevation angles (i.e., relative to the horizontal) were computed in the sagittal plane using classic trigonometry for the trunk and arm segments and the relative angles were computed for the elbow (forearm minus arm elevation angles) and wrist segments (hand minus arm and forearm elevation angles). The amplitude was computed as the maximum minus the minimum angle values.

Measure of distance between trajectories

The distance between two functions (or trajectories) **a**(t) and **b**(t) was computed firstly by using the dynamic time warping technique to temporally align the two functions (Matlab *dtw* function; Helwig et al. 2011), and secondly, by computing the arithmetic mean of the distances between the two time-aligned curves. The latter distances were computed at each time sample and corresponded to the l_2-norm when comparing the 3D positions of the hand (i.e., mid-metacarpus; Table 1) and the absolute difference when comparing segment angles. For comparing lifting and lowering movements, the function corresponding to the lowering movement was first revered in time.

Muscle synergy analysis

We used the synchronous synergy model which assumes that the covariation structure of the muscle activations is time-invariant (Tresch and Jarc, 2009). Muscle synergies were extracted using non-negative matrix factorization (Lee and Seung 2001) which iteratively factorizes the EMG matrix (**E**, of dimension t × m) into the matrix of synergy activation coefficients **C** ($t \times s$) and the synergy weightings matrix **W** ($s \times m$) such that the Frobenius norm ($\| \, \|_{Fro}$s) of the residuals is minimized:

$$\min_{C \geq 0, W \geq 0} \left\| \mathbf{E} - \mathbf{C} \times \mathbf{W} \right\|_{Fro}.$$

The synergy activation profiles correspond to the columns of **C** and the synergy vectors to the rows of **W**. Dimensions t, m and s are the number of time points, the number of muscles and the number of synergies respectively (Hug et al. 2011).

We used the update rules provided in Lee and Seung (2001) to factorize E. To hasten convergence, matrices **C** and **W** were initialized using the scores and loadings obtained from a principal component analysis extracted from the correlation matrix of **E**, negative values being replaced by positive random values (Zheng et al. 2007). At each iteration of the update rule the synergy vectors were normalized by their norm. Contrary to previous publications (Hug et al. 2011) the algorithm was not repeated as the initialization and normalization of the synergy vectors allowed the algorithm to converge to solutions that were identical between different runs and with lower cost than using random initializations.

The accuracy of the model reconstruction was measured using the variance accounted for (VAF) which was computed as:

$$\text{VAF} = 1\text{-SSE/SST},$$

where SSE is the sum of squared residuals and SST the total sum of the squared values. We also compute the 95%-confidence interval of the VAF by implementing a bootstrapping procedure in which the matrix **E** was resampled 100 times with replacement. The synergies were extracted each time. The number of synergies was defined as the minimal number for which the lower bound of 95%-confidence interval of the total VAF was greater than 90%. Synergies were extracted separately for the lifting and lowering tasks. The time-interpolated EMG data from the three loads conditions and the three heights – three trials each – were included to ensure that a substantial motor variability was present in the dataset and enhance the ability of the matrix factorisation algorithm to accurately capture the number of activated synergies (Steele et al. 2015). This resulted in EMG data matrices of dimension 3600 (time samples) × m (number of muscles) for women or 5400 × m for men, i.e., 3600 = 3 (trials) × 2 (loads) × 3 (heights) × 200 (time samples for each trial) and 5400 = 3 (trials) × 3 (loads) × 3 (heights) × 200 (time samples for each trial). The number of columns (i.e., number of muscles) was 10 (group #1) or 13 (group #2).

Finally, contribution of a given synergy (s_i) to the EMG signal was retrieved using:

$$\mathbf{E}_{m,t}(s_i) = \mathbf{C}_{t,s_i} \times \mathbf{W}_{s_i,m}$$

The contributions for a given synergy and a given muscle was then averaged (arithmetic mean) across all time samples and all conditions (height and weights). Note that the contributions computed here do not take into account the residuals (representing ~5-7% of total VAF).

Statistical analyses

Data normality was first verified using Shapiro-Wilk tests. Paired t-tests were used to compare movement durations and velocities, angle amplitudes, mean and peak EMG levels, mean synergy activations, the number of synergies and VAF between lifting and lowering. The hand and segment angle trajectories were compared between lifting and lowering by comparing the intra-condition (Intra) and inter-condition (Inter) distances, with the null hypothesis that similar trajectories would result in similar Intra and Inter. Intra-condition distance corresponded to the distance (defined previously) between each trial trajectory and the average trajectory in a given condition (i.e., averaged across all trials in the lifting or lowering conditions). The intra-condition distances obtained for the lifting and lowering conditions were averaged to get a single value for each subject. Inter-condition distance corresponded to the distance between the average trajectories. Paired t-tests were used to analyze intra- and inter-condition distances. Similarity between synergy vectors were assessed using Pearson's r correlation coefficient. Vectors with a r-value >0.8 were considered as similar. This value corresponds to the 99.6th percentiles of randomly generated unit vectors of dimension 10. Wilcoxon matched-pair tests (repeated measures) were used to assess the effect of the task (lifting or lowering) on the VAF for each synergy. Statistical significance was set initially to $p<0.05$. For multiple comparisons the α-value for rejecting the null hypothesis was adjusted using the Holm–Bonferroni method.

Results

The task and EMGs are illustrated in Figure 1. All of the recorded muscles were activated almost simultaneously during the lifting and lowering tasks.

Kinematics

Hand trajectories and segment angles are depicted in Figure 2. Duration of the lifting and lowering movements were 3.26±0.76 s and 3.43±0.93 s, respectively (p<0.001). These durations corresponded to different velocities of the hand (i.e., 18.2±4.3 cm/s and 15.9±3.8 cm/s for the lifting and lowering movements, respectively; p<0.001). The hand trajectories were similar in the lifting and lowering conditions (Intra = 7.0±8.4 cm and Inter = 2.0±4.4 cm; p=0.162) but they were more variable across trials during lowering (Intra = 6.5±2.4 cm *vs.* 7.6±2.1 cm; p=0.020). Trajectories were different for the trunk angle (Intra = 1.2±0.4° and Inter = 1.8±0.9°; p=0.014) and shoulder angle amplitude was greater during lifting than lowering (i.e., 86.2±23.8° *vs.* 79.6±27.3°; corresponding to +6.6±16.6°; p=0.019).

EMG levels

The mean level of EMG activity was less than 30% MVC in both tasks for all muscles (Figure 3A) and peak iEMG levels were below 50% MVC (Table 1). The most activated muscles were DeltA, DeltM, TraS, SerrA and SupE with mean values of ~10-15% MVC (Figure 3). Mean EMGs were significantly greater during lowering for DeltA, BB, SerrA and Pect (p<0.003–Figure 3B).

VAF and number of synergy

The number of synergies varied between 1 and 3 with a modal value of s=2 observed in ~75% of subjects in both groups (supplemental figure 1). Adding rotator cuff muscles recordings (group #2) did not change the number of synergies. The number of synergies was the same during both the lifting and lowering tasks in 27 out of 30 subjects (90.0%) in group #1 and in 9 out of 10 subjects (90.0%) in group #2. Extracting two synergies in subjects with one synergy (~25% of subjects) resulted in very little differences in the vectors and activation of the two extracted synergies, suggesting that they did not represent independent functional groups.

Total VAF was very similar between the lifting and lowering tasks (p=1) with values of 93.6±1.1% and 93.0±1.7% for groups #1 and #2 respectively. Individual muscle VAF ranged between 86.0±6.0% (BB-lifting) and 94.3±1.4% (DeltM-lifting). When two synergies were present, synergies #1 and #2 accounted for 59.0±13.1% and 33.7±12.1% of the EMG variance respectively in both groups. In group #1 VAF was greater for synergy #1 during lifting than lowering (VAF=63.0±10.4% *vs*.54.9±14.5%; $p<0.001$), while it was greater for synergy #2 during lowering (VAF=30.0±9.8% vs. 37.8±13.0%; $p<0.001$). No effect of the task was found in group #2 ($p>0.60$)

Synergy vectors and synergy activations

Examples of synergy vectors and synergy activations are provided in Figure 4 for group #1 (group #2 presented in supplemental figure 2). The first synergy mainly activated DeltA, DeltM, TraS and SerrA. The second and third synergies co-activated most of the recorded muscles. Synergy vectors were similar between the lifting and lowering tasks in both groups, i.e., r=0.95±0.04 (N=30) and 0.84±0.16 (N=23) for synergy #1 and #2 respectively, in group #1 and the values were r=0.91±0.08 (N=10) and r=0.84±0.11 (N=6) for synergy #1 and #2, respectively, in group #2.

Synergy activation coefficients are displayed in Figure 5 for subjects showing at least two synergies in both tasks and belonging to group #1 (N=23). In this subgroup the averaged activation of synergy #1 was similar in the two tasks ($p=0.306$–Figure 4) while that of synergy #2 was greater during lowering (+27.1±25.8%; $p<0.001$). For subjects with one synergy, the activations of synergy #1 were qualitatively similar as in Figure 5 but with no significant effect of the task ($p=0.340$). In group #2 no effect of the task was found on either synergy ($p>0.187$).

Individual muscle contribution

The contribution of each synergy to the activity of each muscle is presented in Figure 6. In subjects with at least two synergies in group #1, individual muscle contribution was greater during lowering for DeltA, DeltM, SerrA and Pect ($p<0.004$) in synergy #2 (Figure 6B).

Discussion

In this study, we hypothesized that lifting and lowering a load would be associated with distinct activation strategies and distinct muscle synergies. The level of muscle activation was higher during lowering in some muscles (i.e., 1-2% MVC). Meanwhile, the lowering movements were performed more slowly (-2.3 cm/s) and with more variability of the hand paths across trials. The hand trajectories and segment configurations were highly similar during lifting and lowering, despite differences of ~6° for the shoulder angle amplitude and differences of ~2°on average between the trunk angle trajectories. In a majority of subjects (90%) muscle activity was accounted for by the same number of synergies, and these synergies were broadly similar between lifting and lowering. A major finding, however, was the separation of our subjects into two groups. In about 75% of them, two synergies were needed to characterize shoulder muscle coordination and the second synergy was found to be more activated during lowering than lifting. In the remaining subjects, only one synergy was needed, meaning that all shoulder muscles were activated as a unit in both conditions.

How to explain the greater muscle activity during lowering?

Contrary to previous studies, the task of lowering was associated with greater muscle activity than the task of lifting (Hawkes et al. 2012a). The contrary could be expected because muscles are generally stronger (Herzog 2014) and there is a greater amount of motoneuron inhibition during lengthening contractions (Duchateau and Enoka 2016). The loads were the same and cannot explain this greater activity during lowering. EMG can be higher at higher movement velocity, but here, lowering movements were performed more slowly than lifting movements. Therefore, the greater EMG level more likely reflects a difference in coordination strategy. A major difference with the aforementioned study is that the weight of the load was higher, i.e., 6-18 kg in the present study while it was only 1 kg in Hawkles et al. (2012a). Moreover, the finding of lower muscle activity during lengthening contractions have often been made in situations in which no objects had to be held (Gaudet et al. 2018; Kronberg and Brostrom 1995; Ebaugh and Spinelli 2010). Therefore, holding and controlling the movement of an object, as observed here, likely requires a specific control strategy, and actually the data suggest that antagonist coactivation was greater during lowering. It has been shown that the ability to control forces during lengthening contractions is lower than during shortening contractions (Christou and Carlton 2002; Duchateau and Enoka

2008). It is also well established that in tasks requiring precision or in unstable situations, antagonist muscle coactivation is increased, which helps to control the movement by increasing joint mechanical stiffness (Llewellyn et al. 1990; Gribble et al. 2003). Therefore, the inability to precisely control the forces and motor output variability associated with lengthening contractions probably accentuated the need to stabilize the shoulder joints during lowering, hence increasing the level of muscle activity in this condition. The greater variability of the hand path trajectories during lowering is consistent with this hypothesis. The relation between the number of synergy and hand path variability should be investigated further. The present study results also showed that the greater muscle activation during lowering was mainly linked to the activity of synergy #2 (Figure 5). This synergy is constituted of the balanced activity of several antagonist muscles. Therefore, a hypothesis consistent with these observations is that synergy #2 was more activated during lowering to increase shoulder joint stability.

Functional role of the synergies

The major weightings in synergy #1 corresponded to prime mover muscles in both groups of subjects. The roles of these prime movers are to flex the shoulder (medial and anterior deltoids), to upwardly rotate the scapula (serratus anterior), and to elevate the scapula (trapezius superior)(Arborelius et al. 1986). The activation of this synergy also showed more fluctuations during the movement compared to synergy #2 (Figure 5). Therefore, the functional role of synergy #1 was very likely to drive the arm movement.

The role of synergy #2 was likely to stabilize the glenohumeral joint. Firstly, this synergy coactivates several antagonist muscles (e.g., DeltM and LD or Pect and TraL; Figures 4 and 6). Secondly, the amount of activation of synergy #2 is in total agreement with the relative amount of antagonist coactivation observed in previous studies (Blache et al. 2015; Faria et al. 2009). Synergy #2 explained ~30% of EMG variation during lifting and ~37% during lowering. A previous study estimated that the proportion of joint moment that actually contributes to shoulder motion was about 50%, meaning that the remaining 50% (i.e., the joint moments that compensate each other) were dedicated to joint stabilization (Blache et al. 2015). From the results of Faria et al. (2009) who computed the common area in the activity of antagonist muscles pairs, it can be estimated that antagonist muscle coactivation corresponds to ~30-50% of the individual muscle activity level. The present study suggests that non-negative matrix factorization can be used to estimate the amount of muscle activity associated with antagonist co-activation (Figure 7), an application that should be investigated further, using numerical simulations for example (Blache and Begon 2017).

In a non-negligible proportion of our subjects (i.e., ~25%) only one synergy was found. We verified that this could not be accounted for by the method of analysis itself. Firstly, in using different criteria the same number of synergy was found. Secondly, we extracted a second synergy in those subjects and this resulted in highly similar synergy vectors and activations, suggesting that these two synergies did not represent distinct functional groups. Therefore, it must be concluded that all shoulder muscles strongly covaried in amplitude in these subjects. The number of synergies is often taken as a measure of neuromuscular control complexity (e.g., Steele et al. 2015), which suggests that these subjects were less flexible in terms of shoulder muscle recruitment and control. In terms of practical implications, it may be asked whether such inter-individual difference could be correlated with the occurrence of shoulder pathologies.

Changes in intermuscle coordination

In this study we found that the muscle synergy vectors were broadly similar between lifting and lowering. This result is consistent with previous studies showing that changes in mechanical constraints have limited effects on the synergy structure (Hug et al. 2011; d'Avella et al. 2008). However, these results could not be so easily extrapolated to the context of eccentric actions due to the specificity of lengthening contractions regarding the production of muscle force or the complex afferent flow associated with muscle lengthening (e.g., Duchateau and Enoka 2016; Hagen and Valero-Cuevas 2017). A limitation in this reasoning is that the actual movements of the sarcomeres were not recorded (Faulkner 2003).

Although the differences in terms of synergy vector were small, the data suggested that the muscle weightings were altered by the direction of the movement. Firstly, it can be observed that while synergy #1 was similarly activated in both tasks (Figure 5), the activation of biceps brachii in this synergy was more important during lowering (Figure 6). Secondly, the activation of the muscles associated with synergy #2 was greater during lowering for DeltA, DeltM, SerrA and Pect only (Figure 6). If the muscle synergies were perfectly identical in both conditions, increases in activity would have been expected in all muscles of synergy #2 and no changes would have been expected for synergy #1 (Figure 6). This suggests that the relative weightings of the individual muscles in synergy #1 and synergy #2 were altered by the movement direction. The change in the contribution of the biceps brachii is consistent with previous results (Hawkes et al. 2012a) showing decoupled activity of the elbow flexors from the other shoulder muscles during a lifting task. The relatively low *r*-values between the two tasks for synergy #2 (i.e., 0.7-0.8) is also consistent with the hypothesis of altered synergy structure, although the changes are relatively modest. This interpretation is consistent with previous results in the literature suggesting the flexibility of the synergy structure (Muceli et al. 2014). A limitation in the present study is that only the right arm muscles were analyzed.

Conclusions

Contrary to our initial hypothesis, lifting and lowering a load were associated with similar synergy structure. However, the study revealed that lifting and lowering are associated with specific activation of the synergies. In a subgroup of subjects (3/4 of subjects), lowering movements involved greater activation of a “multiple antagonists” synergy than lifting movements. The role of this synergy was very likely to stiffen the shoulder complex joints and ensure their stability. In the rest of the subjects, this second synergy was not observed and all muscles were coactivated as a unit. These results might be of importance to study the link between muscle coordination and interindividual differences in the occurrence of shoulder musculoskeletal disorders.

Declarations

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Conflicts of interest/Competing interests none

Authors' contributions NAT, RM and MB wrote the paper. NAT analyzed the data. RM and MB performed the experiments. MB designed the study. All authors approved the final manuscript.

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Tables

segment	segment definition
trunk	ASIS to shoulder
arm	shoulder to elbow
forearm	elbow to wrist
hand	wrist to mid-metacarpus

Table 1. Segments definition. ASIS, anterior superior iliac spine, *shoulder*, *elbow* and *wrist* correspond to markers placed on the tip of the right acromion, right olecranon and right wrist. *Mid-metacarpus* corresponds to the middle of two markers placed on the 2nd and 5th metacarpal heads.

muscle	lifting (%MVC)	lowering (%MVC)	difference (%MVC)	*p-value*
DeltA	34.9 ± 13.6	37.5 ± 13.7	+2.5 ± 6.1	0.031
DeltM	29.5 ± 15.5	28.9 ± 16.3	-0.5 ± 3.9	0.452
DeltP	12.1 ± 9.9	12.1 ± 7.7	+0.0 ± 4.0	0.972
BB	17.0 ± 14.3	19.7 ± 18.3	+2.7 ± 5.2	0.008
TB	14.6 ± 8.4	13.8 ± 8.2	-0.8 ± 3.3	0.194
TraS	47.8 ± 22.3	46.7 ± 18.4	-1.1 ± 7.4	0.423
TraI	17.9 ± 9.5	18.7 ± 8.7	+0.8 ± 4.5	0.348
SerrA	36.1 ± 13.7	38.3 ± 13.7	+2.2 ± 6.2	0.058
Pect	23.5 ± 15.0	26.5 ± 17.3	+3.0 ± 6.9	0.023
LD	15.9 ± 11.2	16.4 ± 11.3	+0.5 ± 3.6	0.498
SupraE	23.3 ± 16.1	23.2 ± 13.8	-0.1 ± 5.0	0.972
InfraE	19.8 ± 6.2	20.7 ± 7.0	+1.0 ± 6.9	0.667
SubSca	8.2 ± 6.7	8.3 ± 7.3	+0.1 ± 2.1	0.903

Table 2. Peak iEMG levels. The peak values were computed over all conditions and presented in percentage of the maximum obtained during isometric maximal voluntary contractions (MVC). According to the Holm–Bonferroni procedure none of these differences were significant. Values for SupraE, InfraE and SubSca were computed from N=10 subjects. N= 30 for the other muscles.

Figures

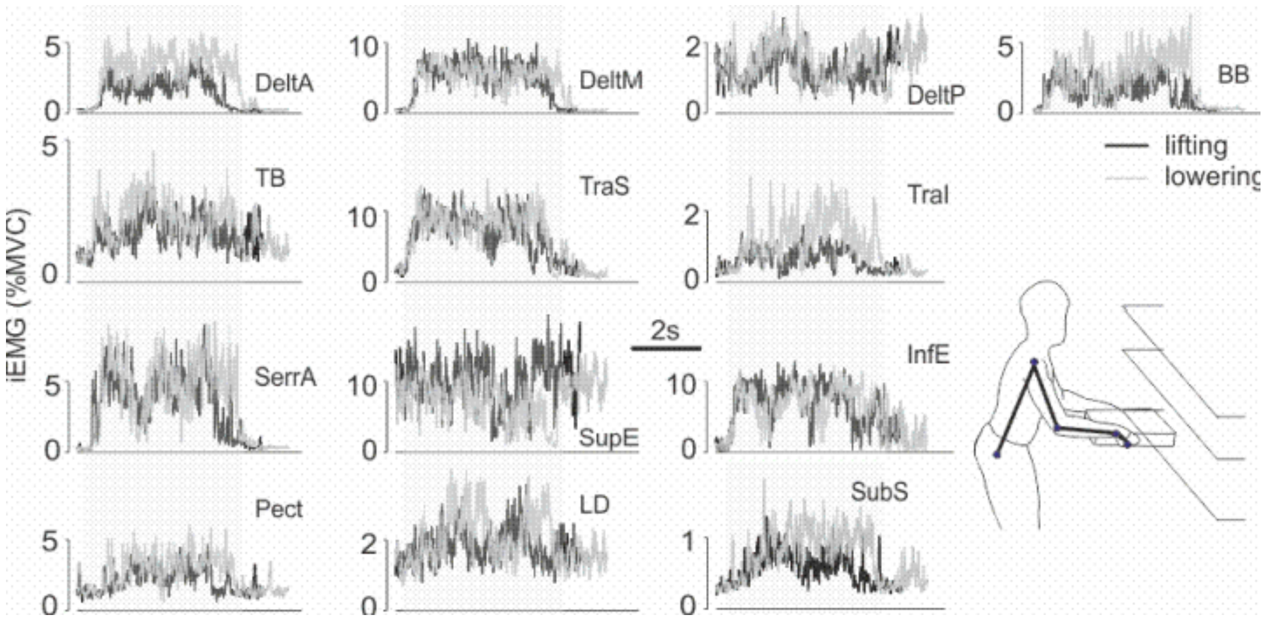


Figure 1. Integrated EMG profile. Example of iEMG time histories for one subject. The tasks in these examples corresponded to lift (black) or lower (gray) a load of 12 kg between shelves, from the hip to the eye levels. The other conditions (with differences in height and load) showed qualitatively similar profiles. The task is illustrated on the lower right corner of the figure. The three shelves were placed at the level of the hip, shoulder and eye. **DeltA**: *anterior deltoid*; **DeltM**: *middle deltoid*; **DeltP**: *posterior deltoid*; **BB**: *biceps brachii*; **TB**: *triceps brachii*; **TraS**: *trapezius–superior part*; **TraL**: *trapezius–lower part*; **SerrA**: *serratus anterior*; **SupS**: *supraspinatus*; **InfS**: *infraspinatus*; **SubS**: *subscapularis*; **Pect**: *pectoralis major–sternal portion*; **LD**: *latissimus dorsi*.

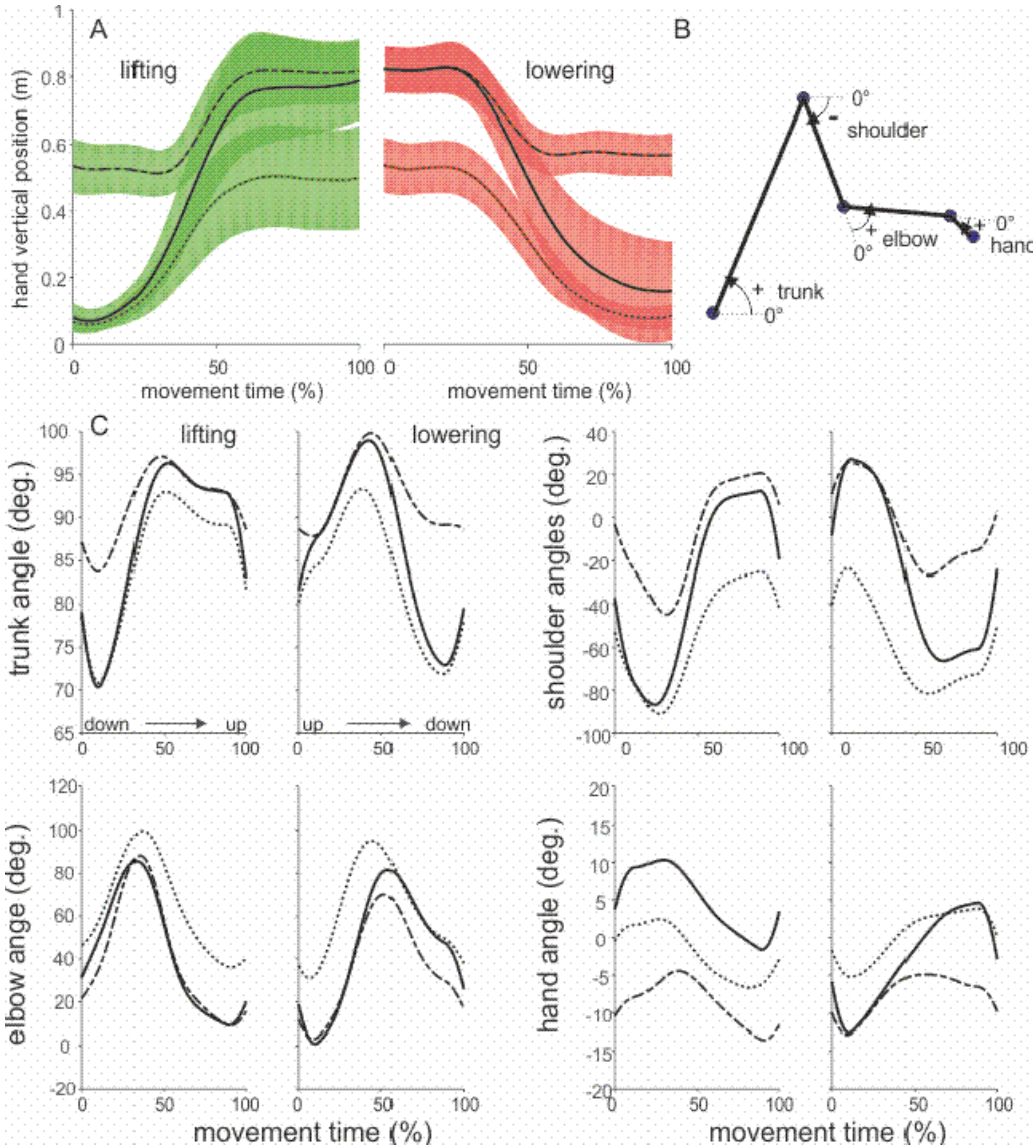


Figure 2. Lifting and lowering kinematics. In A and C, the solid, dashed and dashed-dot lines are the average patterns over all subjects for the movements made between the lower and upper shelves, between the lower and middle shelves and between the middle and the upper shelves, respectively. **A.** Hand (mid-metacarpi) position in the vertical direction (in meter) – see also Table 1. Shaded areas correspond to one SD (inter-individual variability). **B**. Kinematic model. **C.** Segment angles. For each angles the left and right panels correspond to the lifting and lowering movements, respectively. Angles are in degree.

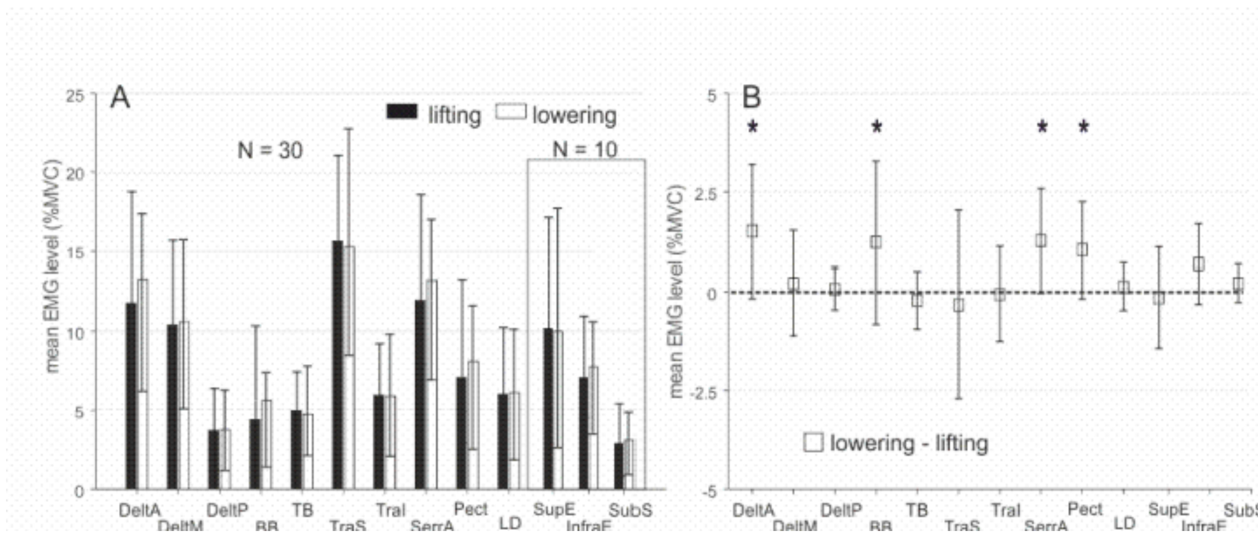


Figure 3. Mean iEMG levels. **A**. iEMG averaged over time and over all conditions (group data). Data are presented as mean ± SD. Note that the number of subjects was only 10 for SupS, InfS and SubS (enclosed histograms). **B**. Difference in the mean iEMG levels between lowering and lifting. A positive difference indicates greater activity during the lowering task. Values are given in percentage of the maximal iEMG obtained during maximal voluntary contractions (**MVC**). *****: $p<0.003$.

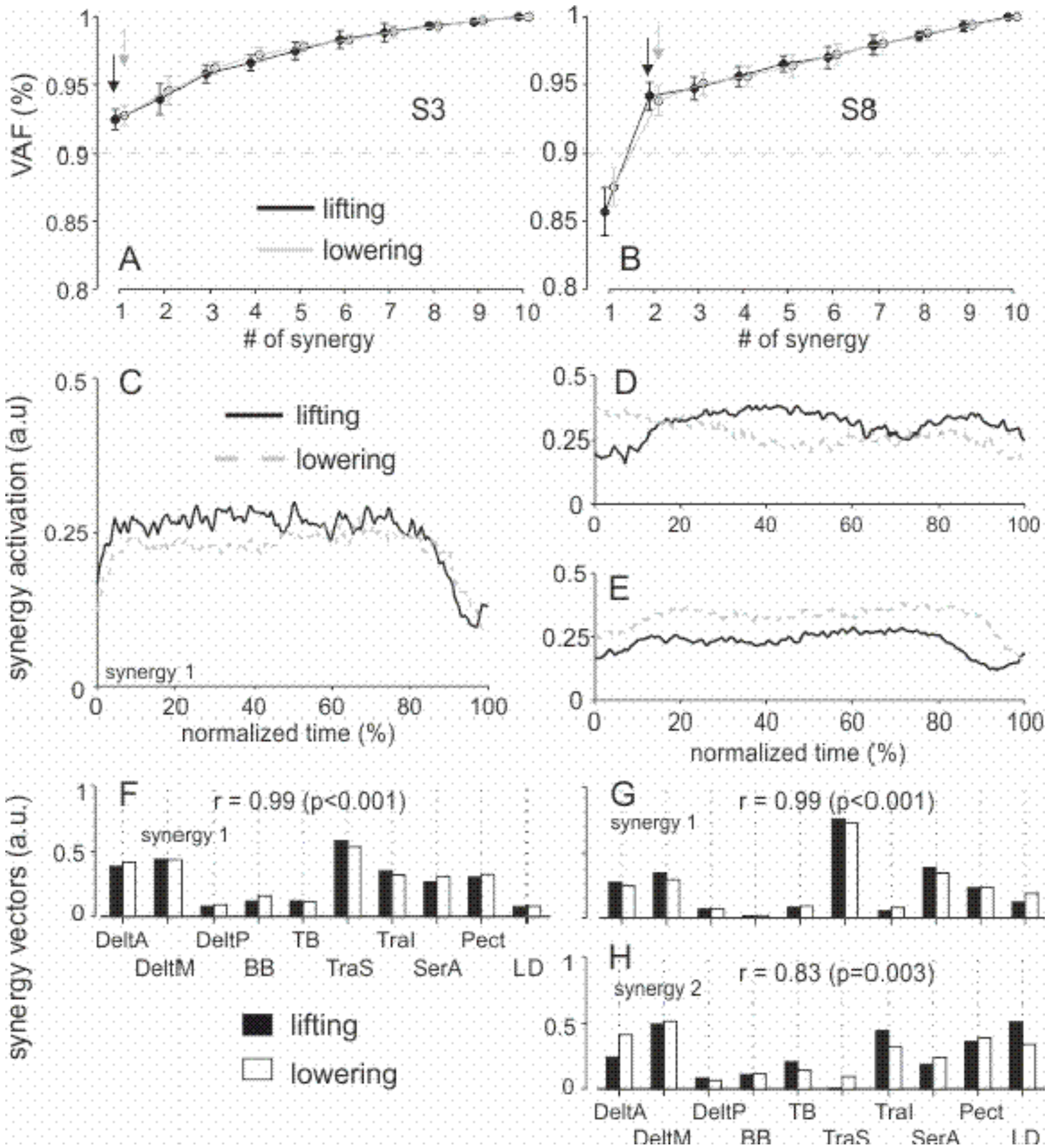


Figure 4. Results of the synergy analysis in two representative subjects (group #1). Panels A, C and F are the data for subject #3 and panels B, D, E, G and H are the data for subject # 8. **A and. B**. Variance accounted for (VAF*)* as a function of the number of synergies. The arrows indicate the number of synergy that were kept for further analysis. **C, D and E**. Synergy activation coefficients, representing the time-varying activation profile of the synergies. **F, G and H**. Unit synergy vectors, representing how muscles are weighted in each synergy. Note that the synergy vectors (in matrix W) transform the synergy activations (matrix H) into muscle activity (matrix E) such that E≈WxH. Pearson's r and their corresponding *p*-values between the lifting and lowering synergy vectors are indicated.

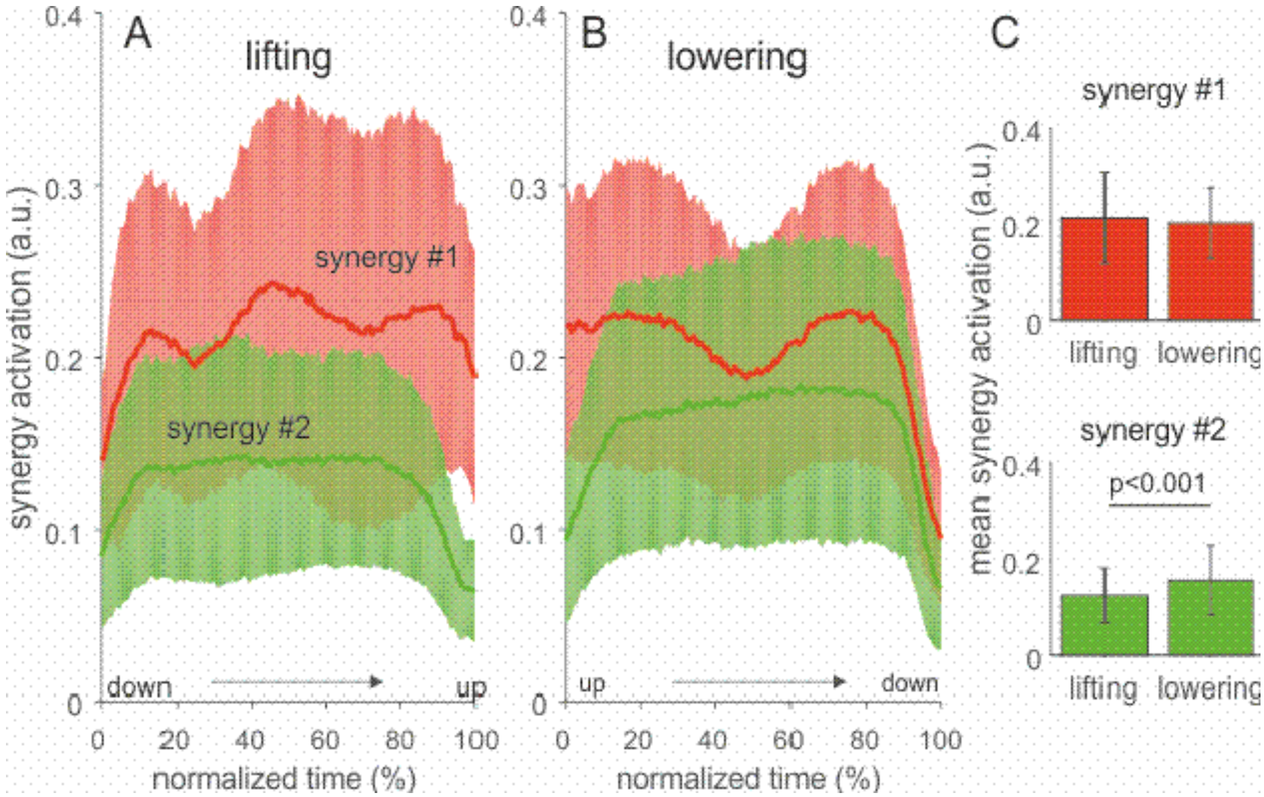


Figure 5. Synergy activation coefficient (group data). A and B: Activation of the synergies. Data presented as the average (tick lines) ± SD (shaded areas) computed across all conditions. **C**. Averaged activation of synergies #1 and #2. Data are for subjects with at least two synergies in both tasks and belonging to group #1 (23 out of 30 subjects).

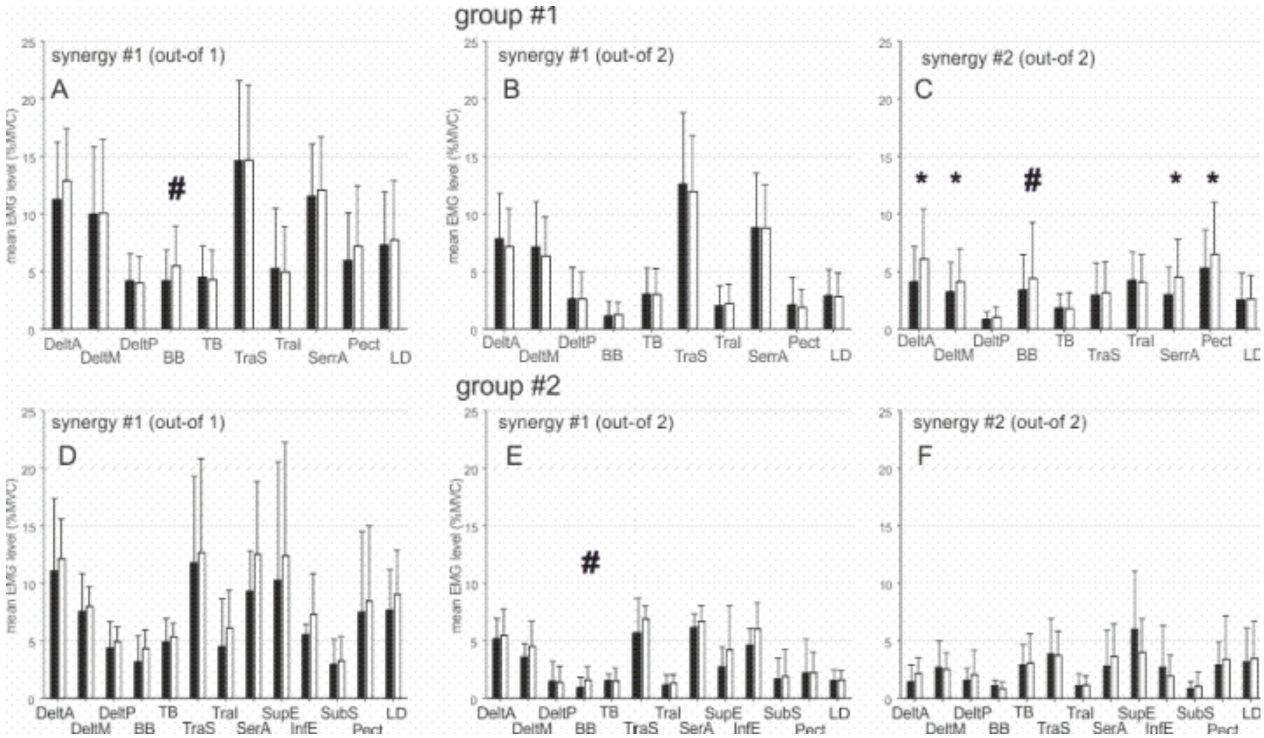
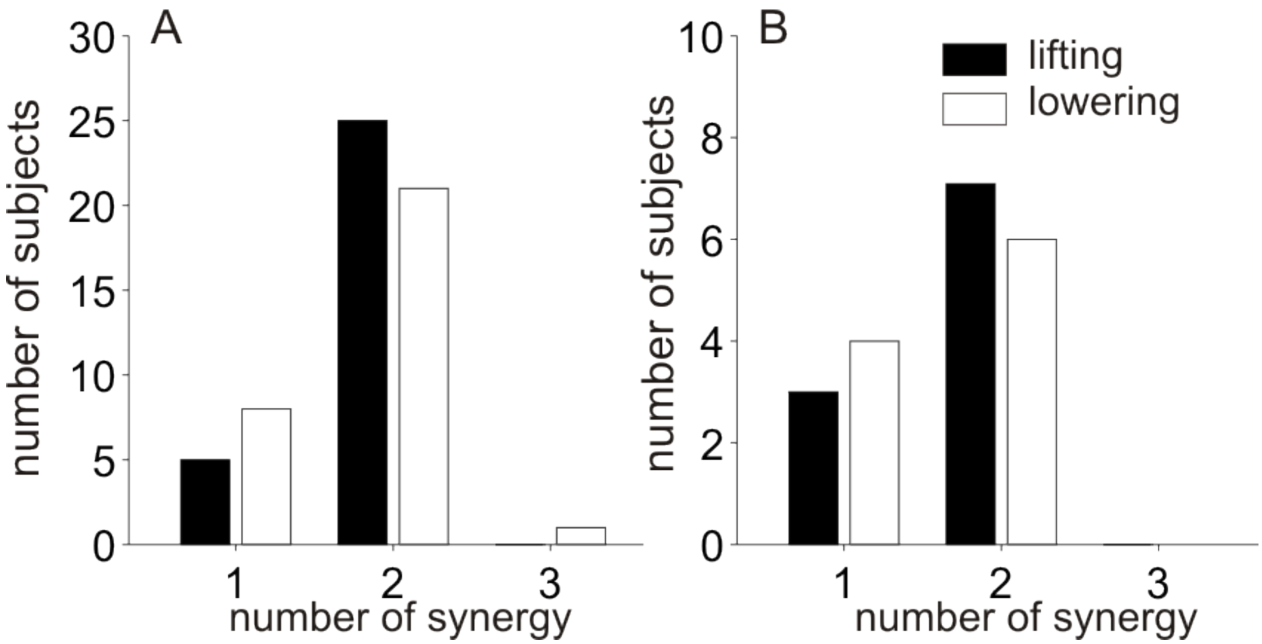
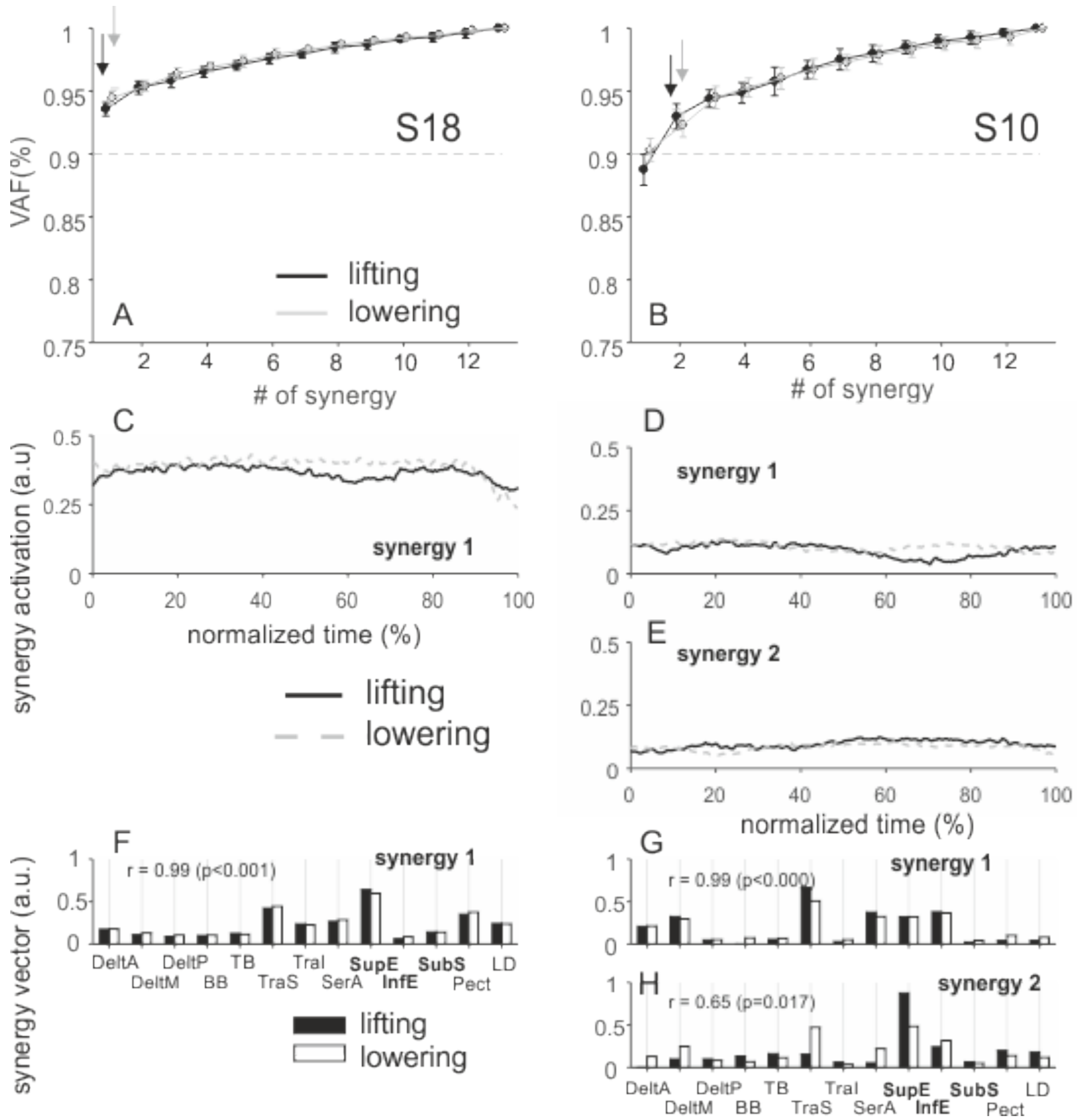


Figure 6. Individual muscle contribution within a synergy. Data represent the averaged muscle activity associated with each synergy for participants in group #1 (**A**, **B** and **C**) and in group #2 (**C**, **D** and **E** – group with intra muscle recording). A and D correspond to the data for subjects with one synergy in both tasks (N = 5 and 4 respectively) and B, C, E and F correspond to the data for subjects with at least two synergies in both tasks (N=23 and N=6 for group #1 and #2 respectively). Significant differences according to the Holm–Bonferroni procedure are indicated by stars (*: $p<0.004$). # indicates a p-value <0.05.

Supplemental Figures and their legends



Supplemental figure 1. Distribution of the number of synergies. A. Number of synergies in the first group of subjects in whom 10 muscles (surface EMGs only) were recorded (N = 30). **B.** Number of synergies in group #2 in whom 13 muscles (surface + intra-muscular EMGs) were recorded (N = 10).



Supplemental figure 2. Results of the synergy analysis in two subjects (group #2). Panels A, C and F are for subject #18 and panels B, D, E, F and G are for subjects #10. Description is the same as in Figure 4. **A and. B**. Variance accounted for (VAF) *vs.* number of synergy curve. The arrows indicate the number of synergy that were kept for further analysis. **C, D and E**. Synergy activation coefficients, representing the time-varying activation profile of the synergies. **F, G and H**. Unit synergy vectors, representing how muscles are weighted in each synergy. Note that the synergy vectors (in matrix W) transform the synergy activations (matrix H) into muscle activity (matrix E) such that E≈WxH. Pearson's r and their corresponding *p*-values between the lifting and lowering synergy vectors are indicated.