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Electromyographic activity in the shoulder-neck region according to arm position and glenohumeral torque

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Summary. The electromyographical (EMG) response to isometric ramp contractions of the right arm, the left arm, and both arms was studied using four pairs of surface electrodes above the right upper trapezius muscle (UT) of six men and six women. Contractions were made against gravity with the active arm(s) in eight positions, ranging from flexion to abduction. To describe arm positions, a new, simple terminology was developed. Root mean square (rms)-converted EMG-signals were normalized (EMG_{norm}) with respect to a reference contraction. The EMG_{norm} corresponding to a 15 N·m torque in the right glenohumeral (GH) joint was strongly related to the position of the right arm ($P<0.001$). The shape of this relationship depended on the electrode position ($P<0.001$). The ratio between EMG_{norm} at 30 N·m and 15 N·m GH torques was related to arm position ($P<0.001$) and differed between electrodes ($P<0.001$). A left-side GH torque resulted in right-side (contralateral) EMG activity, typically corresponding to 20%–30% of that obtained during similar right-side GH torque. Bilateral GH torque implied 0%–50% increase in EMG activity as compared to that obtained with the right arm alone. The results have shown that signals from one pair of surface electrodes above UT cannot be taken as representative of the EMG activity from electrodes located elsewhere above UT. The EMG recordings reflected a complex pattern of muscular activation, significantly related to both outwardly visible factors (arm position, GH torque), and within-body servosystems (motor control reflexes).

Key words: Electromyography – Neck-shoulder – Isometric – Posture – Co-activation

Introduction

Electromyography (EMG) has been used extensively in studies of various vocations to estimate shoulder-neck muscle loads during work tasks involving the arms. This applies to both laboratory (e.g. Hagberg 1981a; Schüldt et al. 1987) and field studies (e.g. Kadefors et al. 1976; Westgaard 1988; Winkel and Gard 1988). The aim has often been to estimate the load on the upper trapezius muscle (UT), as the region covered by UT is a common site of musculoskeletal disorders in monotonous, repetitive arm work (Hagberg and Wegman 1987). Frequently in these studies the shoulder-neck region has been represented by only one pair of surface EMG electrodes, placed at the upper, bulky edge of UT. It is not known if the activity from these is typical of the EMG response of the whole region.

The shoulder-neck muscles are involved in the complex patterns of elevation and rotation of the scapula accompanying arm movements – the so-called "shoulder rhythm" (Inman et al. 1944). In this context, UT acts as an elevator and outward rotator of the scapula. It has been shown that the activity of UT depends on glenohumeral (GH) joint torque (Hagberg 1981a) and arm position (Herberts et al. 1980; Hagberg 1981b; Sigholm et al. 1984). According to their anatomical lines of action, different fibre portions within UT have different potentials regarding scapular elevation and rotation (Högfors et al. 1987). The UT may thus have a multipartite function, with a load sharing between fibres related to arm position and GH torque. Evidence of a multipartite function of the biomechanically much simpler soleus muscle has been shown by Jørgensen and Winkel (1987). Changes in arm position and external load can thus be expected to effect EMG activity differently, according to the recording site above UT.

The major part of UT is attached to the spinal column. Unilateral forces developed in these fibers (e.g. in response to a load held in one hand) must thus be counterbalanced by either passive or muscular forces on the contralateral side, if the spinal column is to remain vertical. Contralateral muscle activity may also be expected as a consequence of excitation overflow in the central nervous system (CNS) (Moore 1975). Schüldt and Harms-Ringdahl (1988) found contralateral muscle

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activity during maximal isometric contractions but no attempts have been made so far to quantify this activity and relate it to arm position or external load. A considerable contralateral co-activation would cause one to dispute the validity of biomechanical models estimating muscular activity from external loads only, as proposed by Högfors et al. (1987).

Our aim was to study the problems mentioned above by investigating the EMG response of the shoulder-neck region to uni- and bilateral GH torque, according to electrode location and arm position.

Methods

Subjects. Twelve healthy subjects (6 men and 6 women) participated in the study (Table 1).

Experimental design

The subjects were seated in an experimental chair to which they were secured by a belt around the chest (Fig. 1). Vertical straps connected each wrist to a horizontal bar below the chair. A strain-gauge force transducer (KRG-4, Bofors, Sweden) was mounted in each of the straps. Tests were made in eight arm positions (Fig. 1). The horizontal angle (H) was controlled by rotating the bar in the horizontal plane, the vertical angle (V) by adjusting strap length and strap-to-bar attachment according to trigonometric calculations based on the anthropometry of the subject. In all tests the elbow joint was kept fully extended, the forearm pronated and the back of the hand turned upwards. Each subject carried out three trials separated by at least 24 h.

Trial 1. The subject was familiarized with the set-up and maximal isometric GH torques were determined in each of the eight arm positions (Fig. 1). The subject was encouraged to exert maximal, vertical pulls for 5 s: (1) using the right arm alone, left arm hanging relaxed; (2) using the left arm alone, right arm hanging relaxed; or (3) using both arms simultaneously. Each of the 24 determinations of maximal torque was made as the best of three closely separated trials and followed by 2 min of rest. The order of arm positions and of arm choice (right, left or both) was balanced between subjects.

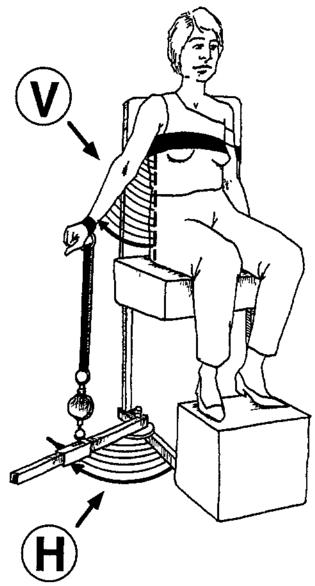


Fig. 1. Experimental set-up and arm position terminology. Horizontal arm position (*H*) is determined as the angle between the vertical perpendicular projections of (a) the upper arm in pure flexion, and (b) the upper arm in the actual position. Vertical arm position (*V*) is determined as the angle between the upper arm and a vertical line through the ipsilateral shoulder joint. Arm position shown: H60V45

Table 1. Data on subjects tested

	Age (years)	Height (cm)	Mass (kg)	MVC⁺ (N·m)	SD⁺⁺
Females					
CF	24	173	55	33.0	2.5
BL	32	173	67	61.2	7.8
NL	31	174	59	45.7	2.5
EG	38	168	62	46.7	2.7
LE	32	168	68	49.9	5.4
IR	26	175	60	49.9	6.5
Mean	30.5	171.8	61.8	47.7	
SD	5.0	3.1	5.0	2.1	
Males					
BS	43	186	80	62.4	9.6
BW	44	180	81	64.7	9.4
TW	38	175	61	53.8	7.4
GJ	37	180	78	67.7	6.3
TJ	38	176	88	51.6	7.2
SR	26	174	62	62.5	10.2
Mean	37.7	178.5	75.0	60.5	
SD	6.4	4.5	11.0	4.3	

⁺, Mean of eight arm positions
⁺⁺, SD of MVC between arm positions, within subject

Trial 2. In each of the 24 combinations of arm position and arm choice (order as on day 1) the subject was asked to pull vertically at the strap, linearly increasing the effort over 8–12 s, from just holding the arm/arms up, to a torque level about 60% of maximal voluntary contraction (MVC); a so-called ramp contraction. The subjects controlled the course of the ramp via visual feed-back of the target level and current strain-gauge force output. Each ramp contraction was preceded by 2-min rest and a subsequent 15-s period of conscious shoulder girdle relaxation, both arms hanging down. Before and after the 24 ramp contractions, the subjects performed a reference contraction, i.e. holding both arms in the H90V90 position for 10 s with a 1.5 kg weight in each hand.

Trial 3. The measurements of day 2 were repeated, reversing the order of arm positions and arm choice.

Registration of EMG and force

In trials 2 and 3, four pairs of surface electrodes (Ag-AgCl, Medicotest A/S, type E-05-VS, Ølstykke, Denmark; interelectrode distance 20 mm) were attached to the skin in the right shoulder-neck region (Fig. 2):

1. (a) At the level of cervical vertebra C1/C2, 25 mm from the spinal column; (b) caudal to 1 (a).
2. (a) Seventy percent of the distance, measured along the skin, between electrodes 1 (b) and 3 (b); (b) lateral to 2 (a).
3. (a) In the sulcus between the most distal parts of acromion and clavicula; (b) medial to 3 (a).
4. (a) Midway between 2 (b) and spina scapula; (b) medial to 4 (a).

The four EMG-signals were pre-amplified close to the electrodes, transmitted for filtering (bandwith 5 Hz–1 kHz) and further amplification, and recorded on an FM tape recorder (TEAC R-71, Tokyo, Japan).

In all three trials the outputs of the force transducers were amplified and recorded on the FM tape recorder. The EMG- and

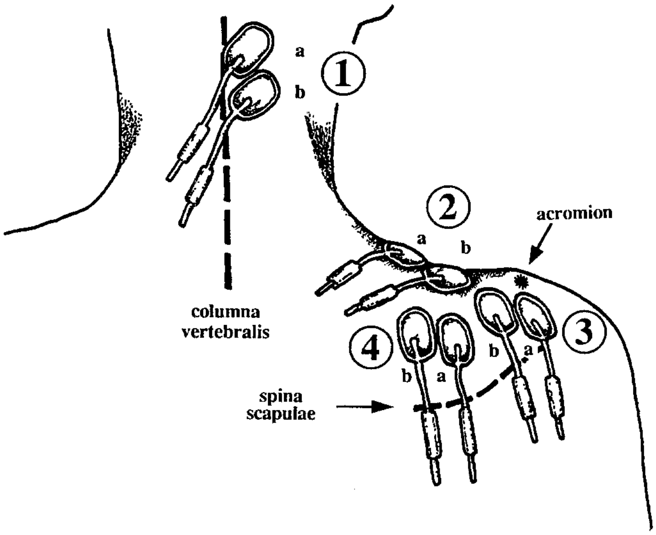


Fig. 2. Positions of the four pairs of surface electrodes. For anatomical description see text

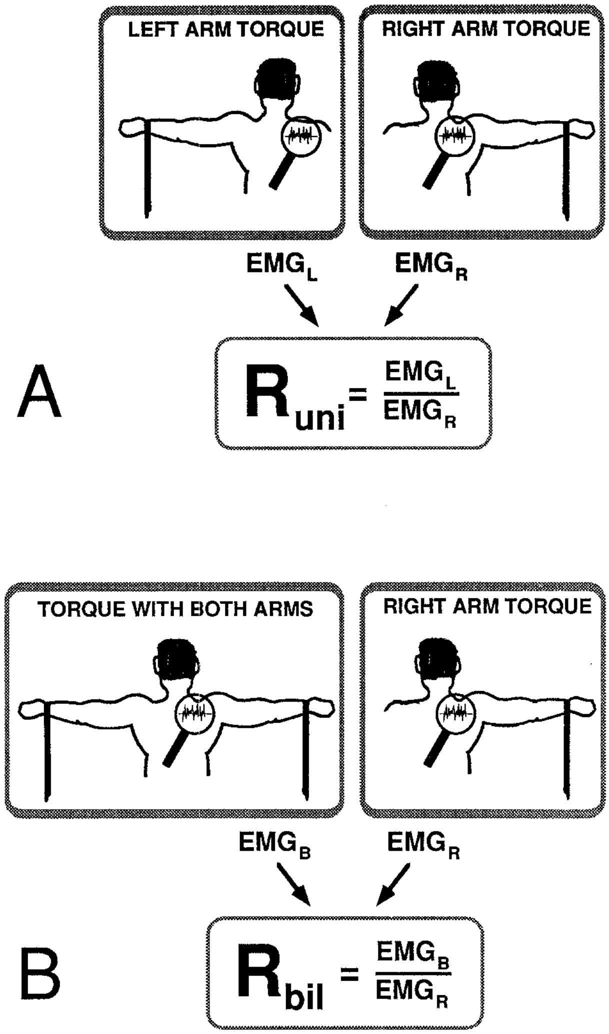


Fig. 3. Definition of the co-activation ratios R_{uni} and R_{bil}. EMG was registered in the right shoulder-neck region during (a) unilateral torque with the left arm (EMG_L), (b) unilateral torque with the right arm in the same arm position (EMG_R), and (c) during a similar bilateral torque (EMG_B). The ratios R_{uni} and R_{bil} are defined as shown

force signal quality was checked by on-line mingograph recording.

Calculations

The EMG signals were rms converted (time constant 100 ms), transmitted to a microcomputer (IBM/PC, AT-2, IBM Corp., USA) through an A to D-converting interface (sampling frequency 20 Hz per channel) and adjusted for relaxation signal level. The EMG signal levels were normalized EMG_{norm} according to the equation:

$$EMG_{norm}\,(\%) = \frac{EMG\,(\mu V)}{R\,(\mu V)} \cdot 100 \qquad (1)$$

where $R = \frac{A1 + A2}{2} \cdot \frac{15}{T}$;

with $\frac{A1 + A2}{2}$, mean of the EMG amplitudes [µV] recorded by the electrode pair concerned during the reference contraction before (*A1*) and after (*A2*) the ramps, and
T, reference contraction torque [N·m], (ranging between subjects from 16.3 N·m to 23.2 N·m)

R is thus an estimate of the EMG level corresponding to a 15 N·m GH torque in the H90V90 position.

Force transducer signals were low-pass filtered (cut-off at 20 Hz) and transmitted to the computer, where the corresponding antigravitational GH torques were calculated and corrected for the additional lift of the arm (Chaffin and Andersson 1984).

The relationship between the GH torque values and the corresponding EMG_{norm} amplitudes was calculated using a linear, an exponential, and a potential least-square regression model. In general the best fit, as judged by r^2, was taken to represent the torque-EMG relationship in question. The chosen equation was then used for calculation of EMG_{norm} values at GH torques of 15 N·m and 30 N·m. About 10% of the left-arm ramps were not considered suitable for estimation by simple regression and were read by eye.

The contralateral (right-side) EMG activity, resulting from a unilateral left arm GH torque, was quantified using the co-activation ratio R_{uni}, defined in Fig. 3. The EMG activity during bilateral contractions was expressed by a ratio, R_{bil}, defined as for R_{uni}, except that bilateral EMG amplitude was used as the numerator.

Statistics

The statistical methods are presented below.

Results

Maximal strength

Maximal isometric GH torque was not significantly related to arm position but differed between sexes (ANOVA). Values are shown in Table 1.

Reference contractions

The EMG signals recorded during the two reference contractions (one before, one after the ramp contractions) were compared. Significant increases in *amplitude* ($P<0.01$, one tailed paired t-test) were seen at all electrodes sites for the female subjects (increase: 6%, 36%, 25%, 40% for electrode pair nos. 1–4, respectively) whereas the male group showed a more inconsistent pattern (increase: −1% ($P=0.41$), 25% ($P=0.09$), 10% ($P=0.01$), 13% ($P=0.10$) for the four electrode pairs, respectively). However, we found no significant differences in the EMG *frequency* spectrum between the two reference contractions, as judged by zero-crossing analysis (Hägg et al. 1987).

Torque with the right arm

In 84% of the calculated torque-EMG relationships, regression explained more than 80% of the total variance of EMG values (i.e. $r^2>0.8$). The quality of the regression did not relate to electrode position. The EMG_{norm} corresponding to 15 N·m and 30 N·m showed between-days-within-subject variation coefficients of 26% and 38%, respectively (mean value over electrode pairs and arm positions). The "classical" electrode position (no. 2 in Fig. 2) showed slightly less variation, 25% and 28% at 15 and 30 N·m. Part of this between-days-within-subject variance is due to random variations in exact electrode location. The between-subject variation coefficient in EMG_{norm} was 19% at both 15 N·m and 30 N·m (mean for all electrode pairs at all arm positions).

Figure 4 A–D presents the EMG_{norm} activities for all electrodes in all arm positions, recorded from female and male subjects at external torques of 15 N·m and 30 N·m. The EMG_{norm} values tested by ANOVA differed significantly between electrodes ($P<0.001$) and between arm positions ($P<0.001$). The difference between electrodes was significantly related to arm position (two-factor interaction, $P<0.001$) and the shape of this relationship depended significantly on gender (three-factor interaction, $P<0.002$) and torque (three-factor interaction, $P=0.001$).

Table 2 shows the result of a stepwise least-square regression analysis of the relationship between arm position and EMG_{norm}. Inspection by eye of Fig. 4 justified that vertical, horizontal and squared horizontal arm position should be tested as regressors in the model (acceptance level $P=0.1$). Thus, the activity of the "neck" electrode pair (no. 1) was related primarily to vertical arm position; the "classical" electrode pair (no. 2) showed a similar dependence on vertical position, combined with a marked relationship to horizontal position; and electrode pair no. 4, caudal to the "classical" electrode pair, had a distinct relationship to both vertical and horizontal position. The most lateral electrode pair (no. 3) showed a pattern including quadratic components.

To study further the EMG response to an increase in external load, the ratio between EMG_{norm} at 30 N·m and 15 N·m was calculated (Fig. 5). The ANOVA showed a significant relationship to electrode ($P<0.001$), arm position ($P<0.001$) and gender ($P=0.02$), even when electrode pair no. 1 was excluded ($P=0.02$, $P=0.001$ and $P<0.005$, respectively).

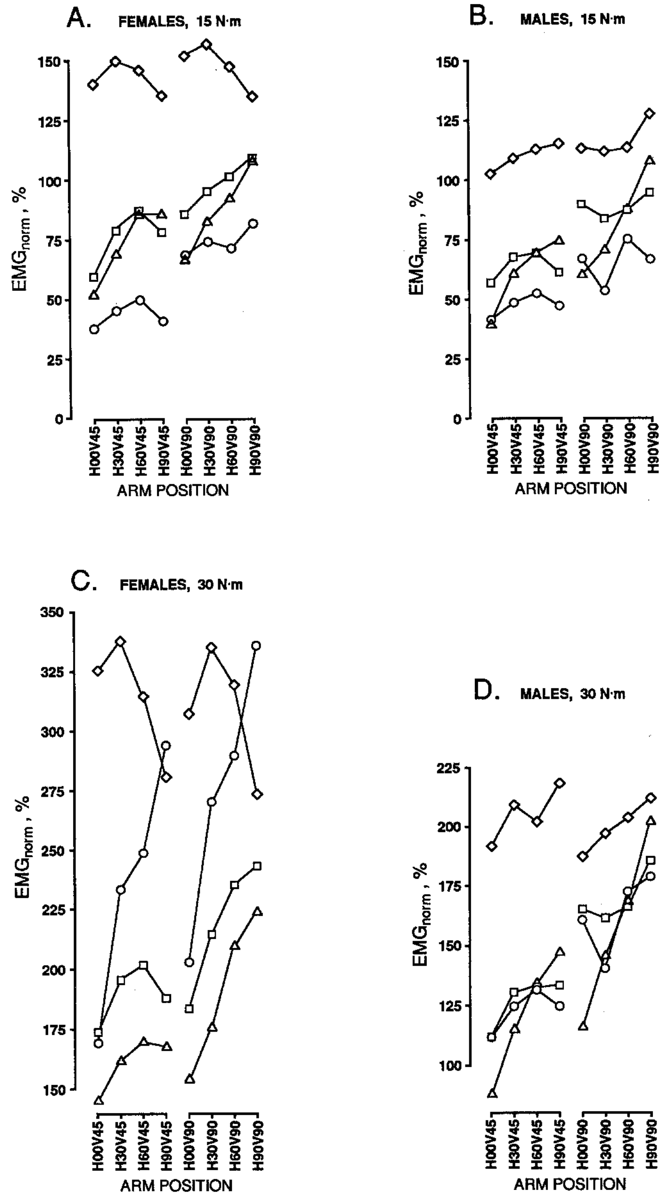


Fig. 4. Right side electromyogram (*EMG*) activity during right arm torque. Normalised EMG (*EMG_{norm}*) amplitude for all four electrode pairs (○: *E1*, □: *E2*, ◇: *E3*, △: *E4*, see Fig. 2), corresponding to glenohumeral torques of 15 N·m **A**, **B** and 30 N·m **C**, **D**, in women **A**, **C** and men **B**, **D** ($n=6$ for each group). *H*, Horizontal angle; *V*, vertical angle

A GH torque of 15 N·m corresponds to a hand-held mass of about 1 kg, when the arm is fully stretched and horizontal (V90). According to Fig. 4 A and B, changing the arm position from 90° abduction (H90V90) to 45° flexion (H00V45) decreased the EMG activity measured at the "classical" electrode position (no. 2) by 35% (male) and 44% (female). The corresponding decrease for the most lateral electrode (no. 3) was only 20% (M) and −5% (F). Doubling the GH torque caused an increase in EMG activity of the "classical" electrode of 198% in 45° flexion (F), but only of 120% in 90° abduction.

Table 2. Relationship between arm position and normalised EMG (EMG_{norm}). Results of stepwise regression analysis of the curves presented in Fig. 4. Regression was tried against the factors vertical (V), horizontal (H), and squared horizontal (H^2) arm position, using an acceptance limit of $P=0.1$. Table shows regression coefficients of accepted regressors, intercept (i) of the resulting equation and variance explained by regression (r^2). Empty cells denote non-accepted regressors

15 N·m	Females					Males				
	V	H	H^2	i	r^2	V	H	H^2	i	r^2
E1	0.68			13.3	0.91	0.41			29.2	0.69
E2	0.49	0.61	0.0042	40.3	0.94	0.56			38.9	0.89
E3			−0.0023	154.8	0.92	0.15		0.0016	98.4	0.85
E4	0.32	0.42		40.8	0.95	0.46	0.46		20.2	0.95
30 N·m	**Females**					**Males**				
	V	H	H^2	i	r^2	V	H	H^2	i	r^2
E1	0.85	1.35		138.2	0.96	0.89			83.5	0.75
E2	0.65	0.41		142.1	0.76	0.94	0.22		75.4	0.95
E3	1.02		0.016	318.6	0.93	0.26			191.4	0.77
E4	0.66	0.53		107.9	0.84	0.82	0.80		49.0	0.97

E1, E2, E3, E4, abbreviations of electrode pairs, see Fig. 2.
Example: the EMG_{norm} amplitude of E3, males, 15 N·m can be predicted by the equation:
$EMG_{norm} = 0.15 \cdot V + 0.0016 \cdot H^2 + 98.4$
15% of the variance of EMG_{norm} amplitude is then left unexplained by the regression

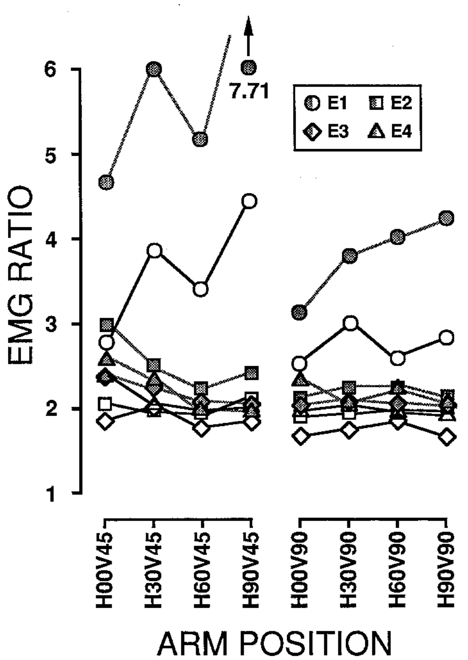


Fig. 5. Ratio between normalised electromyogram activities recorded at right arm glenohumeral torques of 30 N·m (denominator) and 15 N·m (numerator). *E1–E4*: electrode pairs 1–4, ref Fig. 2. *Filled symbols, grey lines*, women ($n=6$); *unfilled symbols, black lines*, men ($n=6$)

Torque with the left arm

Unilateral GH torques with the left arm caused, in general, substantial EMG activity in the right shoulder-neck region (see below). However, the shape of the torque-EMG relationship was not uniform. Three types – with some overlap – appeared, as shown in Fig. 6. About 80% of all 768 registered relationships could be described by type A. Type B (about 10% of all curves) was horizontal until a torque value corresponding to about 50% MVC, above which the EMG activity increased steeply. Type C (about 10%) was steadily horizontal, usually at a signal level close to the noise level. There was a tendency for curve types B and C to appear more frequently in flexion than in abduction but great individual variations masked possible systematic relationships.

Figure 7 presents the contralateral co-activation, expressed through the ratio R_{uni}. The between-days-within subject variation coefficient in R_{uni} was 70%. The R_{uni} showed a significant relationship (ANOVA) to electrode ($P<0.001$) and to arm position ($P<0.001$). When electrode 1 was excluded these relationships persisted ($P<0.01$ and $P<0.001$, respectively) and only one interaction term was significant, the influence of gender on the relationship between torque and R_{uni} ($P<0.01$). Thus a complete picture of the variation of R_{uni} with respect to electrode, arm position, gender and torque can be obtained by first calculating the mean R_{uni} for any desired combination of electrode and arm position (plotted in Fig. 7) and then adding multiplication factors, expressing how gender and torque influence these mean values (table in Fig. 7).

No correlation was seen between R_{uni} and maximal strength.

The extent of contralateral activity can be illustrated by pointing out, that a *left* arm GH torque of 30 N·m exerted in abduction (H90V90) induced an EMG activity in the "classical" electrode (no. 2) on the

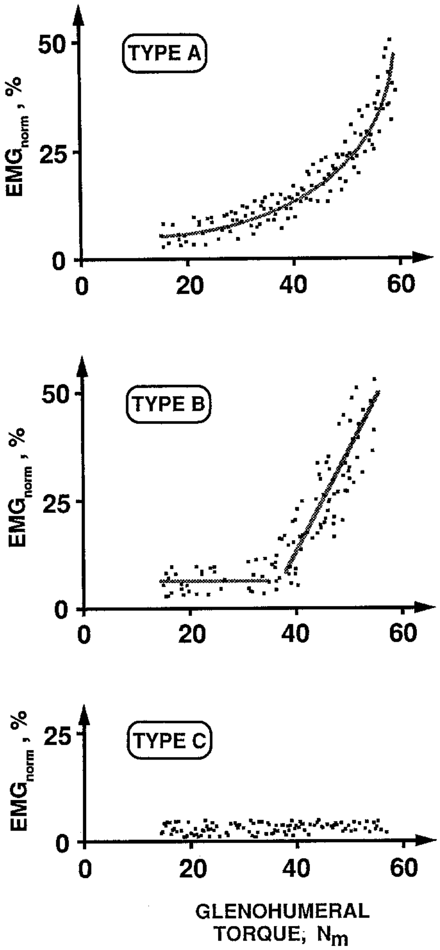


Fig. 6. Examples of the three types of relationship observed between left arm glenohumeral torque and normalized electromyographic activity (EMG_{norm}) in the right upper trapezius muscle

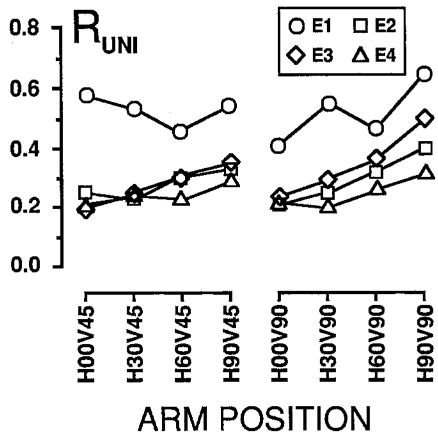
right side corresponding to a *right* arm GH torque close to 15 N·m, although the subject was instructed to relax his right arm and shoulder.

Torque with both arms

Figure 8 presents values of the ratio R_{bil}. A value of R_{bil} larger than 1.0 indicates that the EMG activity in the *right* shoulder-neck region is increased when adding left arm activity to an otherwise unilateral right arm torque.

The between-days variation coefficients in R_{bil} were 33% and 23%, including and excluding electrode 1, respectively. The R_{bil} did not relate significantly to gender or external torque (ANOVA); the data were, therefore, pooled across these variables. The R_{bil} was significantly related to electrode ($P<0.001$). This significance persisted when electrode pair no. 1 (the "neck" electrode pair) was excluded and in addition significant dependence on arm position was seen ($P<0.001$). The most lateral electrode (no. 3) was unaffected by position changes, whereas electrodes 2 and 4 showed a greater bilateral "superactivation" in abduction as compared to flexion (two-factor interaction between electrode and arm position, $P<0.001$). In most arm positions R_{bil} was significantly greater than 1.0 (one-tailed *t*-test, Fig. 8).

During the bilateral ramp contraction in the H90V90 position the subjects performed a GH torque



Conversion factors from mean values shown:

	Electrode 1 15Nm	Electrode 1 30Nm	Electrode 2,3,4 15Nm	Electrode 2,3,4 30Nm
F	0.73	0.63	0.72	1.23
M	1.77	0.88	0.96	1.10

Fig. 7. Contralateral co-activation during unilateral glenohumeral torque. Values of the ratio R_{uni} (defined in Fig. 3). *E1–E4*, electrode pairs 1–4, see Fig. 2. Values shown are means of 12 subjects (6 males and 6 females) and two glenohumeral torques (15 N·m, 30 N·m). To estimate values specific to torque level and sex, use the multiplication factors shown in the Tables. Example: R_{uni} corresponding to electrode pair no. 3 at a 30 N·m torque, position H60H90 in females (*F*): 0.37 (Fig.) × 1.23 (Table) = 0.46

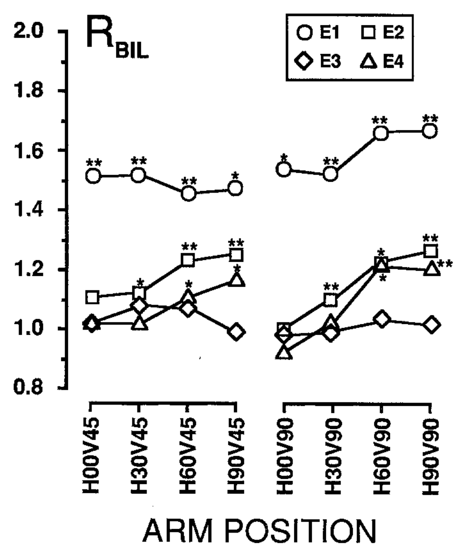


Fig. 8. Right side EMG activity during bilateral torque. Values of the ratio R_{bil} (defined in Fig. 3). *E1–E4*: electrode pairs 1–4, see Fig. 2. Values shown are means of 12 subjects (6 males and 6 females) and two glenohumeral torques (15 N·m, 30 N·m). Ratios significantly larger than 1.0 are indicated as follows: *: $0.01<P<0.05$, **: $P<0.01$ (one-tailed *t*-test)

similar to the one exerted during the reference contractions before and after the ramps. However, the EMG_{norm} corresponding to this GH torque was on average 26.5% larger during the ramp contraction than during the reference contraction ($P<0.01$, two-tailed paired t-test).

Discussion

Methodology

Arm position terminology. No generally accepted consensus exists regarding the terminology of (upper) arm position in relation to the GH joint centre. A rather complicated description including anatomical terms in combination with angle measurements has been proposed as a standard (American Academy of Orthopaedic Surgeons 1965; Hagberg 1988). However, authors have tended to adapt their own ad-hoc terminology (Herberts et al. 1980; Armstrong et al. 1982; Sigholm et al. 1984; Högfors et al. 1987). The descriptive terminology proposed in this study covers all (upper) arm positions in a comprehensible way, consistent with conventional polar geometry. It does not possess the exactness required for refined biomechanical calculations (Högfors et al. 1987) but serves well as an instrument for description of work postures.

Electrode positioning. The positions of the four electrode pairs used in this study were aimed at picking up activity from different parts of UT. Cross-talk from underlying and adjacent muscles may, however, have affected the EMG recordings, especially in slender subjects. According to Basmajian and DeLuca (1985) muscle fibres down to a depth from the skin surface equalling the inter-electrode spacing will contribute meaningfully to the EMG signal. Thus, recordings from electrode pairs nos. 1 and 3 may be considerably influenced by activity from muscles other than UT e.g. the erector spinae, splenius capitis, and semispinalis capitis muscles (electrode 1), and supraspinatus muscle (electrode 3). Electrode pair no. 4 may, to a minor extent have picked up activity from the supraspinatus muscle, whereas electrode pair no. 2 should have produced a recording purely from the underlying UT fibres. For further discussion see Schüldt et al. (1987). Changes in arm position inevitably change the sample of motor units recorded by surface electrodes in the shoulder-neck region; (a) due to sliding of the skin over the muscle surface and (b) due to changes in the shape of the underlying muscle belly. When the arms are kept below shoulder level as in this study, systematic effects of skin sliding are judged less important. And it seems unlikely that changes in muscle shape alone could account for the observed influence of arm position on EMG activity.

Some of these problems of validity could be met by using intramuscular electrodes. However, this would introduce new methodological problems, due to the high selectivity of such electrodes, e.g. difficulties in interpreting signal differences related to changes in arm position (Basmajian and DeLuca 1985). Furthermore, it would be difficult to draw conclusions concerning vocational surface electromyography.

Test regimen. One trial in this study included 24 submaximal isometric contractions up to 60% MVC, lasting 8–12 s each, interspersed by 2–2.5 min of rest. The regimen did not seem to induce fatigue, as judged by the zero-crossing analysis of the reference contractions. The observed increases in EMG amplitude of the reference contraction may, however, indicate a changing pattern of coordination in the course of a trial. However, possible systematic errors owing to fatigue are neutralized through the balancing of arm position order between and within subjects.

No systematic differences in EMG activity were seen between the two repetitions of a given ramp for a given subject (two-tailed t-test). This suggests that the tests had no training effect on muscle patterns of coordination.

Torque. In this study external exposure (GH torque) is quantified using an absolute term (N·m) rather than a relative one (e.g. %MVC). The latter parameter is the more common in vocational studies of EMG and requires a translation of EMG activity into %MVC by means of a pre-work recording of the relationship between external (relative) torque and EMG amplitude (e.g. Jonsson 1982; Hagberg and Sundelin 1986). Several arguments, however, speak in favour of using absolute external load as the exposure parameter:

1. The validity of relative measurements relies upon a correct determination of MVC. This introduces important methodological problems, especially when unaccustomed or less motivated subjects, or those with a disease, are tested (Westgaard 1988).
2. If the muscle investigated is part of a complex agonist-antagonist system – as is the trapezius muscle – maximal activation of the whole muscle complex does not necessarily imply maximal activation of that individual muscle (Schüldt and Harms-Ringdahl 1988). It is not known whether UT is activated maximally in any voluntary effort. The true UT MVC may thus be impossible to determine by voluntary activation.
3. In working life, loads are typically presented in absolute rather than relative terms (e.g. in manual materials handling).
4. Some load-related physiological reactions may be more closely related to absolute than to relative strain, e.g. intramuscular pressure (Barnes 1980; Järvholm et al. 1988).

Physiology and biomechanics

Torque with the right arm. This study has shown that a certain external GH torque causes different levels of EMG_{norm} at different electrode positions in the shoulder-neck region. The signal level is strongly related to either vertical or horizontal arm position, or both, depending on electrode location (Fig. 4). This relationship cannot be ascribed to interposition differences in MVC, as MVC did not vary significantly with arm position.

The response of a single EMG-recording above UT to a change in arm position has been studied by several authors. Unfortunately, the studies differ substantially in experimental set-up and data processing. (Järvholm et al., accepted for publication) recorded intramuscular EMG from the belly of UT and showed the relationship between GH torque and EMG activity to be dependent on arm position in a pattern consistent with our findings. Sigholm et al. (1984) also used intramuscular recording from the UT belly. Raising the arm and moving the arm from flexion to abduction was found to increase EMG activity. However, Sigholm et al. did not take into account that different arm positions per se impose different GH torques. We, therefore, recalculated their data (as read from their figures) to give approximate EMG values corresponding to a constant 15 N·m GH torque. Raising the flexed arm from 45° to 90° (H00V45→H00V90 in our terminology) led to a 50% increase in this recalculated EMG. This is consistent with our findings from electrode pair no. 2. A similar elevation of the arm in abduction (H90V45→H90V90) did not change EMG activity in their study, in contrast to the ≈60% increase observed by us. Differences between intramuscular and surface recording and inaccuracies in the recalculation procedure might explain this discrepancy.

Inman et al. (1944) investigated the responses of an intramuscular UT electrode – with unspecified location – to flexion and abduction of the arm. Signal level was about 15% higher in H90V90 than in H00V90 but no difference was seen between H90V45 and H00V45. Raising the arm from V45 to V90 increased the EMG activity – adjusted for changes in GH torque – by about 35% in abduction (H90) and by about 15% in flexion (H00). These results are consistent with our data from electrode pair no. 2. Schüldt et al. (1987), investigating hand-arm movements, reported a significantly greater EMG activity from the belly of UT with abducted upper arm as compared to flexed. Herberts et al. (1980) and Hagberg (1981b) have shown faster EMG-signs of fatigue in UT (electrode position no. 2) in sustained arm abduction (H90V90) as compared to sustained arm flexion (H00V90), giving indirect evidence of a greater involvement of UT in abduction than in flexion.

Thus, the literature supports the results obtained in the present study as far as the influence of changes in horizontal arm position on EMG activity is concerned, whereas the evidence concerning vertical changes in arm position is sparse and conflicting.

The studies by Wiedenbauer and Mortensen (1952), Schüldt et al. (1987), and Schüldt and Harms-Ringdahl (1988) indicate that the EMG response to a change in arm position depends on the location of the electrode pair. This is consistent with the interaction between electrode pair and arm position found by us. Major differences in experimental design and data presentation make closer numerical comparisons futile.

The positions of the arm and the scapula are closely linked. This so-called "shoulder rhythm" was described by Inman et al. (1944). Raising the arm (an increase in the angle V) is brought about by simultaneously rotating the scapula outwards and increasing the GH angle, in a 1:2 angular ratio. At the same time the scapula is elevated, i.e. slides cranially (and in flexion even laterally) across the thorax wall. The exact interplay of movement between humerus and scapula differs substantially between individuals (Högfors, personal communication). Scapular movements are accomplished by a number of muscles (including UT) attached to the scapula and the clavicula. Table 3 states the synergist/antagonist relationships between muscles linking the scapula to the body. Changes in arm position are accompanied by changes in the lever arms of several of these muscles. Besides their roles in positioning the scapula, some of these muscles (among them UT) act as anti-gravity muscles with respect to external loads on the shoulder girdle. The force distribution pattern between the shoulder muscles can thus be expected to be complexly related to arm position and external load. Thus, it hardly seems worthwhile trying to compare observed surface EMG levels to predictions based on functional anatomy.

Furthermore, several (anatomically defined) muscle entities in the shoulder-neck region may be organized functionally in separate subunits. The UT most probably has a multipartite function such as this, as shown by EMG observations (Westgaard 1988), and fibretype characteristics (Lindman et al., accepted for publica-

Table 3. Synergist/antagonist patterns of muscles linking scapula/clavicula to the body. Based mainly on Inman et al. (1944)

	Arm abducted (H90)		Arm flexed (H00)	
	scapular elevation	scapular rotation	scapular elevation	scapular rotation
trapezius muscle, middle part	O	+	O	+
trapezius muscle, lower part	–	+	–	+
serratus anterior muscle	–	+	–	+
rhomboidei muscles	+	–	O	O
levator scapulae muscle	+	–	+	–

+, Synergist in relation to upper trapezius; –, antagonist in relation to upper trapezius; O, no significant action

tion). The concept is supported by our study: changes in arm position and GH torque had different effects on EMG activity depending on electrode location, also as regards the electrode pairs nos. 2 and 4 considered to represent mainly UT.

Torque with the left arm and with both arms. The present study showed that unilateral arm activity increased the tension level of shoulder-neck muscles on the other side of the body. The increase is specifically related to electrode location and arm position (Fig. 7). Bilateral arm activity produced more EMG activity than did unilateral (Fig. 8).

Contralateral co-activation has been demonstrated during maximal activity in shoulder-neck muscles (Schüldt and Harms-Ringdahl 1988), and in elbow flexors (Gregg et al. 1957; Moore 1975). All studies have reported a co-activation amounting to 10%–30% of the ipsilateral activity level, as measured by EMG amplitude. This value corresponds well to our submaximal observations.

At a 30 N·m GH torque, the contralateral muscle activity is sufficient to raise that arm against gravity. However, no arm movement is seen, indicating that the muscle activity is part of a co-contraction pattern in the contralateral shoulder-neck region.

An external, unilateral GH torque imposes a sideward bending force on the vertebral column. Contralateral shoulder-neck muscle activity may seem biomechanically justified as a reflex compensatory measure taken against this potential bending. However, a purely biomechanical explanation of the contralateral co-activation is not sufficient. Firstly, some of the subjects managed without co-activation, or with a co-activation not directly related to the absolute ipsilateral torque (Fig. 6 B and C). Secondly, the demonstrated co-activation in bilateral arm activity (Fig. 8), where bending forces on both sides of the vertebral column balanced out each other, had no biomechanical justification. Neither could the co-activation be ascribed to inadequate motor learning, as it persisted unaffected throughout all trials of a subject. Thus, contralateral co-activation may be primarily related to sensomotor control processes in the CNS, related to e.g. postural reflexes or excitation overflow (Moore 1975).

The *sum* of the *ipsilateral and contralateral* EMG activities was greater than the corresponding *bilateral* EMG level (i.e. $1+R_{uni}>R_{bil}$) in 88% of the 768 observation sets ($768=12$ subjects $\times$ eight arm positions $\times$ two trials $\times$ four electrode pairs). Thus, contralateral co-activation seemed not to be simply additive to ipsilateral muscle activity. This may in part be ascribed to differences in postural reflex patterns between the unilateral and bilateral situations. Non-additivity at one or more levels in the CNS is another plausible explanation (Brooks 1986).

The considerable extent of contralateral co-activation implies that shoulder-neck muscle activity cannot be predicted with any validity from observations of external loads only. This statement is further supported by our observation that a certain GH torque calls for a 25% greater EMG activity during a ramp contraction than when holding weights. The ramp contraction apparantly elicits a more pronounced proximal fixation of the arm. This is consistent with several authors who have showed postural adjustments to be related to sensory-motor demands in hand-arm work (e.g. Weber et al. 1980; Westgaard and Bjørklund 1987). Hand-arm work operations associated even with crude demands for active motor control seem to cause an increase in shoulder-neck muscle activity.

Conclusions

1. The amplitude of surface EMG recorded above the upper trapezius muscle at a constant glenohumeral torque depends on the vertical and horizontal position of the arm.
2. Changes in arm position or glenohumeral torque cause different changes in EMG amplitude, according to electrode location above upper trapezius. Thus, a multipartite function of upper trapezius is suggested.
3. Shoulder-neck muscle activity is significantly influenced by several factors not biomechanically related to arm position or glenohumeral torque. Contralateral coactivation is one such factor.

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