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EMG activity in neck and back muscles during selected static postures in adult males and females

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Surface electromyographic (EMG) amplitudes were gathered from 100 men and 100 women while maintaining the end range of nine motor tasks. Ratios of EMG amplitudes were used to characterise the activation patterns of 14 muscle groups of the back and trunk during 10 motor tasks. Procedures to identify electrode placement sites were developed to ensure reliability of all EMG recordings. Subcutaneous fat was estimated at each muscle site and a correction factor was used to account for signal attenuation due to the impedance attributable to adipose tissue thickness. Logarithmic transformations were performed to obtain a Gaussian distribution of the EMG amplitudes and muscle ratios. The transformed EMG amplitudes and transformed ratios were highly reliable between sessions across nine active motor tasks (Pearson's *r* and intra-class correlations ranged from 0.74 to 0.96). Significant gender differences were observed in the transformed EMG amplitudes and ratios of amplitudes in selected muscles and muscle pairs. It appears that the transformed EMG ratios represent a reliable means of assessing muscle recruitment patterns in a series of well-defined motor tasks in a large population of presumably normal adult male and female subjects. The acquisition of this large database under well-controlled conditions using defined criteria for each motor task provides a template to which individuals with injuries involving the neck and trunk musculature can be compared.

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INTRODUCTION

Absolute differences in EMG amplitudes (Ahern et al, 1988; Arena, Sherman, Bruno and Young, 1990; Cohen, Naliboff, Schandler and McArthur, 1986; Collins, Cohen, Haliboff and Schandler, 1982; Kravitz, Moore and Glaros, 1981), or differences in patterns of EMG activity from bilateral torso sites (Ahern et al, 1988; Cohen et al, 1986; Collins et al, 1982), to discriminate muscle activity between subjects with no back pain and low back pain (LBP) patients during standing or sitting generally have been difficult to identify consistently. Arena, Sherman, Bruno and Young (1989), however, reported that during standing, five types of LBP groups exhibited significantly higher paraspinal EMG levels than normal subjects. Cram and Engström (1986) and Jayasinghe, Harding, Anderson and Sweetman (1978) also reported significantly higher mean paraspinal EMG amplitudes for LBP patients compared with normal subjects during sitting and standing. One study reported significant left–right differences in paraspinal EMG activity of LBP patients compared with normal controls during sitting and standing (Cram and Steger, 1983), while in another no significant differences were found during sitting (Hoyt et al, 1981).

EMG recordings obtained during well-defined kinematic activities have been discriminatory in identifying abnormalities in muscles of the trunk (Ahern et al, 1986, 1988; Basmajian, 1978; Collins et al, 1982; Nouwen, Van Akkerveeken and Versloot, 1987; Price, Clare and Ewerhardt, 1948; Sihvonen, Partanen, Hanninen and Soimakallio, 1991; Wolf and Basmajian, 1978). These studies in general indicate that unique and predictable differences in the recruitment of selected muscles can be detected within and across subjects when the tests are carefully controlled and when a large number of subjects are studied (Edgerton, Wolf, Levendowski and Roy, 1996a,b).

Differences in the methods used, such as electrode location (i.e. muscles studied), electrode spacing, equipment, filtering characteristics and the kind of motor tasks performed, make it difficult to compare results across these studies in attempts to identify abnormalities in EMG associated with LBP (Dolce and Raczynski, 1985; Nouwen and Bush, 1984; Wolf, Segal, Wolf and Nyberg, 1991). In addition, significant subject-to-subject variation exists in the EMG recordings because of varying levels of adipose tissue which can significantly dampen the electrical signal, particularly from deep muscles (Ortengren and Andersson, 1977; Sihvonen, Partanen and Hanninen, 1988; Sihvonen et al, 1991), because the adipose tissue acts as a low-pass filter (Basmajian and DeLuca, 1985) and affects the EMG spectral characteristics (DeLuca, 1993) as well as the EMG amplitude. Inter-subject comparisons also have limited our understanding of trunk muscle function, perhaps because of differences across subjects, muscle morphology and function. Gender appears to have been another factor which has contributed to large inter-subject variability (Nouwen et al, 1987; Wolf, Basmajian, Russ and Kutner, 1979). Thus it appears that the sensitivity of EMG measures to identify unique EMG characteristics in LBP patients in large population studies can be improved by controlling several factors that are known to affect EMG signals (Ahern et al, 1988; Arena et al, 1989, 1990, 1991; Cram and Engstrom, 1986).

The purpose of the present study was to determine the EMG amplitudes and recruitment patterns of control subjects in a large population of men and women while performing a series of precisely identified motor tasks. These data from control subjects can be used as a template to compare patients with musculoskeletal injuries involving the neck or trunk musculature (Kibler, 1990; Roy, Baldwin and Edgerton, 1991).

METHODS

Subjects

One hundred females (mean age 29 ± 9 years; range 18–56) and 100 males (mean age 29 ± 10 years; range 18–61) were studied. The criteria used to ensure that these subjects were 'normal' were as follows: asymptomatic for back pain, leg length discrepancies and scoliosis. Furthermore, these subjects (based on self-report) had never been treated by a physician for back injury or pain. No effort was made to provide a clinical classification to screen subjects that might actually have clinically definable abnormalities (DeLitto, 1994). Even though there was no obvious reason to assume that these subjects were not normal, subsequent analysis of these data based on classification rules to establish 'normal' recruitment patterns indicated that 10% of the males and 9% of the females exhibited abnormal recruitment patterns. The frequency of abnormal and subclinical recruitment patterns observed in a sample of 61 acute back injury patients was significantly greater at 80% (Edgerton et al, 1996a).

A second group of 20 females (mean age 30 ± 6 years; range 20–41) and 20 males (mean age 28 ± 6 years; range 19–43) asymptomatic for back pain and screened as mentioned above were recruited for a between-session reliability study. Two technicians tested each subject once, with a 14 day inter-tester interval.

A third group of seven males (age 26 ± 7 years; range 19–40) and seven females (age 22 ± 3 years; range 18–25) asymptomatic for back pain and screened as mentioned above were recruited for a study to compare the EMG recording collected simultaneously for narrow and wide bandpass settings.

Selection of motor tasks

The EMG activity of the trunk and neck muscles was studied while the subject maintained a fixed position during the execution of nine motor tasks. For each motor task, data from four trials interspersed with a 7 sec standing position were used. For each trial, the subject had 2 sec to proceed into the static end of range position (isometric phase) and hold that position for 5 sec (Fig. 1).

Task 1 consisted of standing in a relaxed position. Motor tasks 2 and 3 were asymmetrical equivalents with one arm raised overhead to 45°, the opposite lower extremity extended posteriorly to 35°, with the pelvis maintained in a stable position while balancing on a narrow base of support. A shoulder shrug, task 4, elevated and upwardly rotated the shoulder girdle, a movement which occurs during lifting or manoeuvring light objects at waist level, such as assembly work. In task 5, both arms were abducted to 90°. This motor task tested the subjects' ability to abduct the shoulder girdle, stabilise the erector spinae and balance the trunk. Task 6 was forward trunk flexion to approximately 45° while the head was maintained in line with the trunk and the lordotic curvature was minimised by maintaining a flat back. This motor task tested the holding capacity of the erector spinae and the trapezius muscles as would be required to retrieve objects while working over a table. In task 7, the arms were flexed to 90° to simulate working with the arms

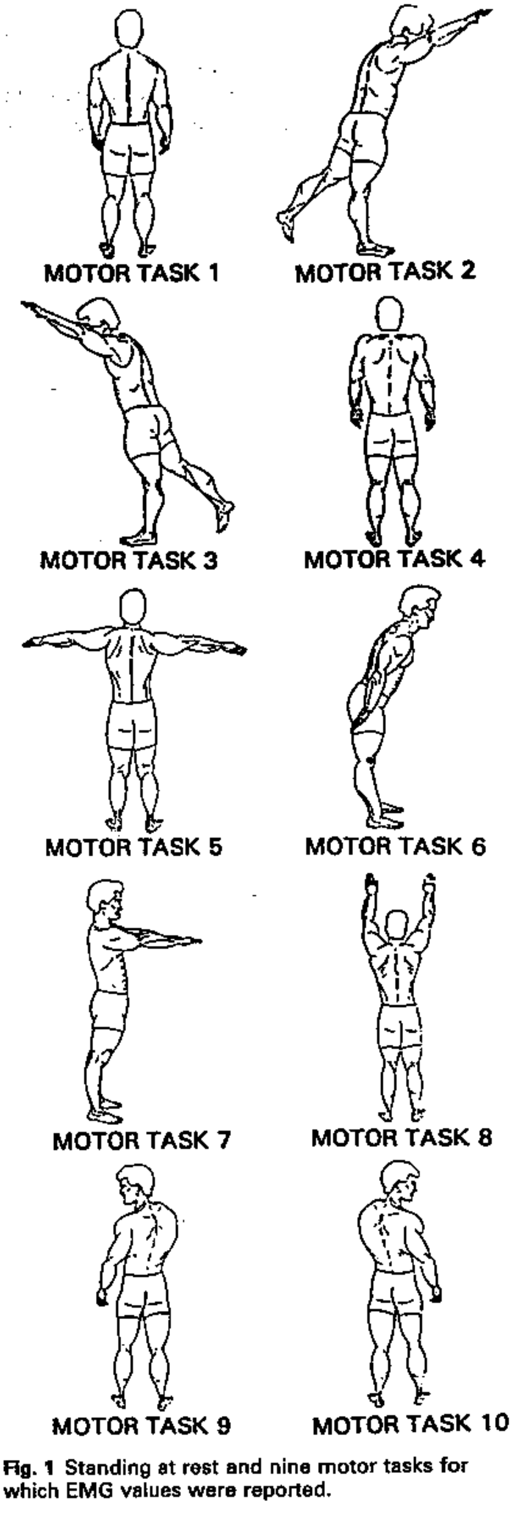


Fig. 1 Standing at rest and nine motor tasks for which EMG values were reported.

in front of the body while maintaining an erect posture. In task 8, the subjects were asked to maintain a symmetrical position while reaching overhead with both arms. Motor tasks 9 and 10 consisted of rotating the trunk to the left and right with the pelvis stabilised and the trunk extended.

Electrode sites

Fourteen electrode sites that include most of the major muscle groups which control and stabilise the cervical, thoracic and lumbar spine were selected for study. To ensure electrode placement reliability, procedures were developed which relied on three landmarks (spinous process of C7, acromioclavicular joint and iliac crest) and 13 anthropometric measurements to define the 14 electrode sites on both sides of the torso. To improve electrode placement reliability, physical measurements were taken from radiographic images of 84 females and 83 males. Regression equations were derived to predict the distance from cervical vertebra C5 to the location of the spinous processes at T5, T10 and L1 based on gender, height and distance from C7 to the iliac crest. The mean overall difference between the predicted and the actual measurement was 0.24±0.04 (SEM) inches across all subjects. Examples of the regression equations for predicting the distance from C5 to T5 are as follows:

Females:

T5 = 0.0127(height) + 0.2575(C5 to iliac crest distance)

Males:

T5 = −0.0022(height) + 0.3016(C5 to iliac crest distance)

Correlations between the radiological measurements and the predicted measurements across sites were highly reliable ($r = 0.99$).

EMG measurements were obtained from seven bilateral muscle sites of the back (Fig. 2). The superior upper trapezius/cervical paraspinal (1) sites (superior upper trapezius) were placed 2.5 cm lateral to the cervical spine at C5 near the origin of the superior trapezius, and over the semispinalis and splenius capitis. The upper trapezius (2) sites were located half the distance between the acromioclavicular joint and the lateral edge of the sternomastoid muscle at the coronal plane. The middle trapezius (3) sites were centred between the spine and the medial edge of the scapula at T5. The lower trapezius/thoracic paraspinal (4) sites (lower trapezius) were placed 2.5 cm lateral to the thoracic spine at T10 near the origin of the lower trapezius and over the spinalis thoracis and longissimus thoracis. The latissimus dorsi (5) sites were located lateral to L1 but centred vertically below the upper trapezius location (site 2). The abdominal oblique (6) sites (oblique) were placed 2 cm above the iliac crest and vertically aligned with the upper trapezius location (site 2). The lumbar paraspinal (7) sites

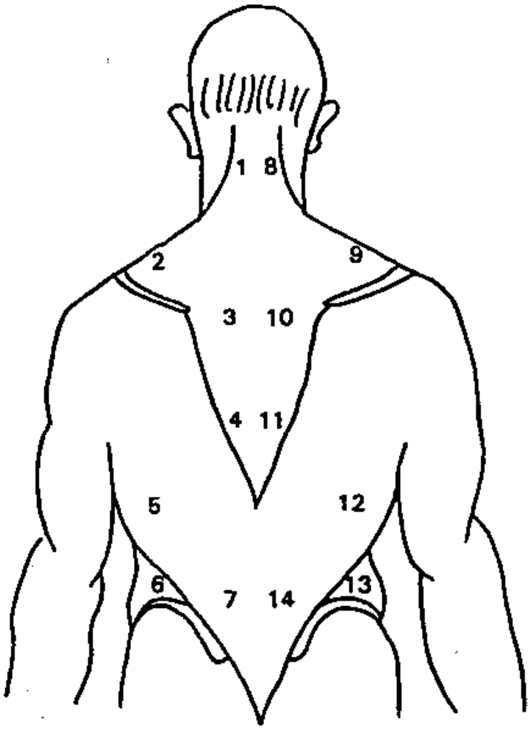


Fig. 2 Location of seven bilateral muscle sites (left/right) for electrode placement: superior upper trapezius/cervical paraspinal (1/8), upper trapezius (2/9), middle trapezius (3/10), lower trapezius/thoracic paraspinal (4/11), latissimus dorsi (5/12), oblique (6/13) and lumbar paraspinal (7/14).

were located 2.5 cm lateral to the lumbar spine at the iliac crest (L4) over the erector spinae.

Skinfold measurements were taken at each muscle site as described previously (Edgerton et al, 1996a) using plastic calipers (Fat-o-meter; Rockton, IL) to estimate the distribution of subcutaneous fat (mm) between the electrode and the muscle at each site.

The skin was prepared with rubbing alcohol and an abrasive cream (Omni Prep, Aurora, CO), and a single electrode set with two silver/silver-chloride 1.0 cm diameter discs spaced 2.5 cm apart (Multi-Biosensors, El Paso, TX) were affixed parallel to the direction of the underlying superficial muscle fibres.

Instrumentation used for normative data

EMG data were collected using 14 channels of a 16 channel system with $10^{12}\ \Omega$ input impedance differential amplifiers and a common ground. Analog signals were first bandpass-filtered, rectified, smoothed using a RC hardware filter with a 145 msec time constant and then underwent A/D conversion. The bandpass filter settings were 66 and 215 Hz with a 12 dB per octave roll-off at both the high and low ends. The 66 Hz setting may be considered low by some investigators because of signal attenuation. However, this setting eliminates 50–60 Hz interference and minimises the effects of ECG artifact (Basmajian and DeLuca, 1985, p. 51). The 215 Hz setting was used because the signal-to-noise ratio can be improved while eliminating less than 2% of the power spectrum (Solomonow et al, 1990). These bandpass settings were chosen to maximise the signal-to-noise ratio because any low-frequency interference would tend to corrupt the electromyographic signal under our measurement conditions. Subsequent ratiometric analysis of the EMG data indicated that highly reliable information capable of discriminating back injury patients from controls can be collected using these bandpass settings (Edgerton et al, 1996a). An audio signal was used to prompt patient response. A visual display of the EMG signal was used for monitoring the general characteristic of the EMG signal.

Instrumentation used for bandpass study

EMG data were collected using 7 channels of a 16 channel system with $10^{12}\ \Omega$ input impedance differential amplifiers, a common ground, fibre optics isolation, a common mode rejection ratio of 130 dB at the electrode site and $\pm 0.1\ \mu V$ zero input noise level (Flexcomp™, Thought Technology Ltd, Montreal, Canada). Analog signals were sampled at 1984 samples per second, underwent 13 bit A/D conversion, and optically transmitted to a digital signal processing board. The EMG data were then bandpass-filtered at two different settings. The wide bandpass specification was 20–500 Hz with 36 dB per octave roll-off with a 60 Hz notch filter. The narrow bandpass was 66–215 Hz with 12 dB per octave roll-off. The digital signals were then rectified, smoothed with a RC filter with a 145 msec time constant, and decimated to 15.5 data points per second.

ECG artifact was readily apparent in the wide bandpass tracing when amplitudes were below 10 μV. To eliminate this potential bias between the two bandpass settings, algorithms were derived empirically to recognise the beginning and end of each ECG spike. The data points that corresponded to each heart beat were removed from all seven channels for both the narrow and wide bandpass settings.

Data reduction and statistical analysis

EMG amplitudes for each of four 5 sec 'static end of range position' periods were averaged for each muscle and for each motor task. To account for signal attenuation resulting from adipose tissue across muscle sites and subjects, procedures were used to derive a correction factor to normalise the EMG data. In the development of the adipose tissue correction factor, certain assumptions were made: (1) the rate of attenuation of the EMG signal is a constant and only depends on the adipose tissue thickness; and (2) the source of variability during any particular motor task is a function of the muscle's morphology, tension and gender. Thus an estimate of the common slope (β) would represent the attenuation rate of the EMG signal and the y-intercept (α) would represent the EMG signal in the absence of adipose tissue (Hemingway, Bidermann and Inglis, 1995). An analysis of covariance (ANCOVA) was used to estimate these parameters and then fitted to the EMG recordings of the 200 subjects for the 14 muscles during the 10 motor tasks which accounted for variability in the EMG signal due to muscle morphology, muscle tension and gender. The correction factor derived from the ANCOVA model was estimated by:

$$\log(\text{EMG}) = \alpha_{\text{muscle, motor task, gender}} - \beta \times \text{skinfold} + e$$

where a base 10 logarithmic transformation was performed on the averaged EMG values, $\alpha_{\text{muscle, motor task, gender}}$ is the y-intercept of the regression line for each muscle, motor task, gender combination, β is the common slope for each muscle, motor task, gender regression line, skinfold is the adipose tissue thickness measurement at the muscle site (mm), and e is the residual error resulting from the fit of the regression line to the data.

The increase in mean transformed EMG amplitudes attributed to the correction factor for adipose attenuation was determined for each bilateral muscle group across the 10 motor tasks. Muscle ratios were computed by taking the base 10 logarithm of the adipose corrected EMG for $\text{Muscle}_{1\ldots14}$ divided by $\text{Muscle}_{2\ldots14}$ (i.e. [Log (EMG Muscle_1/EMG Muscle_2)]. Of the 182 muscle ratios per motor task (14×13 muscle combinations), a half were reciprocals (i.e. [$\text{Muscle}_1/\text{Muscle}_2] = [1/(\text{Muscle}_2/\text{Muscle}_1)]$) and could be eliminated because they contained no unique information. Means and standard deviations of each of the remaining 91 transformed ratios were computed for the men and for the women.

Coefficients for skewness and kurtosis were computed for each of the 910 transformed ratios (91 ratios × 10 motor tasks). Standardised coefficients of a magnitude greater than 2.0 were used to identify muscle ratios that did not conform to a Gaussian distribution (Snedecor and Cochran, 1967). Variables with significant skewness and kurtosis were transformed (e.g. base 10 log function) to attain a Gaussian distribution to accurately represent variability of the data (i.e. standard deviation or standard error) about the mean (Moore, 1995). A repeated measures analysis of variance was performed using the $\log_{10}$ EMG transformed values to determine overall significant gender differences. Further analysis of individual gender differences using the transformed EMG amplitudes and ratios of specific muscle groups used the two-tailed t-test for independent groups. The alpha level for statistical significance was set at 0.05. In those instances where multiple t-test comparisons were made, the Bonferroni adjustment of the alpha level was used.

In certain analyses, such as the analysis of the narrow and wide bandpass settings, in which there was no clear distinction between the independent and dependent variables, the Pearson product–moment correlation coefficient (r) and intra-class correlation (IC) coefficient were computed (Zar, 1984). EMG signals were collected from both narrow and wide bandpass settings from the left latissimus dorsi muscle during the execution of two motor tasks: arm abduction to 90°, which requires low EMG activity, and trunk rotation to the left, which requires higher levels of EMG.

Mean transformed EMG amplitudes for the narrow and wide bandpass settings, the total sample root mean squared (RMS) differences between individuals, and percent differences between bandpass settings (i.e. RMS differences divided by the pooled mean transformed EMG amplitudes for both bandpass settings) were computed for each muscle group and motor task combination.

The Pearson product–moment correlation coefficient was also used to evaluate the linear relationship of the between-session pairs of transformed EMG amplitudes and ratios *as well as the intra-class correlation coefficient.* This analytical strategy was applied within and across all motor tasks.

The intra-class and Pearson's correlation coefficients were also computed between the narrow and wide bandpass mean transformed ratios across the seven muscle groups and 10 motor tasks. Before computing ratios between the 210 muscle pairs between the seven muscle groups for 10 motor tasks, EMG activity below 2.5 µV was truncated to 2.5 µV for both bandpass settings to remove variability attributed to measurement error at low amplitudes.

RESULTS

Correction for signal attenuation attributed to adipose tissue

The adipose correction factor (β) was 0.0123 with a coefficient of determination (R^2) of 0.75 for the ANCOVA model. This correction factor was then applied to the EMG measurements from all muscle sites. The mean increase in transformed EMG amplitudes across seven bilateral muscle groups and 10 motor tasks attributed to the correction factor was 12% (Table 1). The upper trapezius had the lowest mean caliper measurements (5.9 mm), greatest mean transformed EMG amplitudes (log = 1.40 µV) and smallest percentage attributed to the correction factor (5.0%). The oblique had the greatest mean caliper measurements (19.2 mm), lowest log transformed EMG amplitudes (log = 0.95 µV) and greatest percentage attributed to the correction factor (25.3%).

Logarithmic transformation

Of the 910 ratios (91 unique ratios × 10 motor tasks), 507 were positively skewed (mean 2.55, SD 1.59, range 0.41 to 13.24) and 804 of the same 910 ratios were predominantly leptokurtic (mean 12.64, SD 19.30, range −0.24 to 180.22). The skew occurred primarily in motor tasks when minimal EMG activity was recorded. After logarithmic transformation, none of the muscle ratios were significantly skewed (mean 0.03, SD 0.31, range −0.93 to 1.89) and only 17 had kurtosis greater than 2.00 (mean 0.17, SD 0.84, range −0.88 to 10.83).

Transformed EMG data

The means and standard deviations for all muscles for the 10 motor tasks are presented for males and females in Figs 3 and 4 respectively. Several general observations are apparent from Figs 3 and 4. The activity of all muscles is relatively low during standing. There are clear differences among muscles in each of the nine movements; for example, the relative differences among muscles are unique for each of the nine movements. These differences reflect symmetry of the motor tasks as would be expected. Although there are gender differences, the general amplitude profile of the EMG signals is strikingly similar in females and males (Figs 3 and 4).

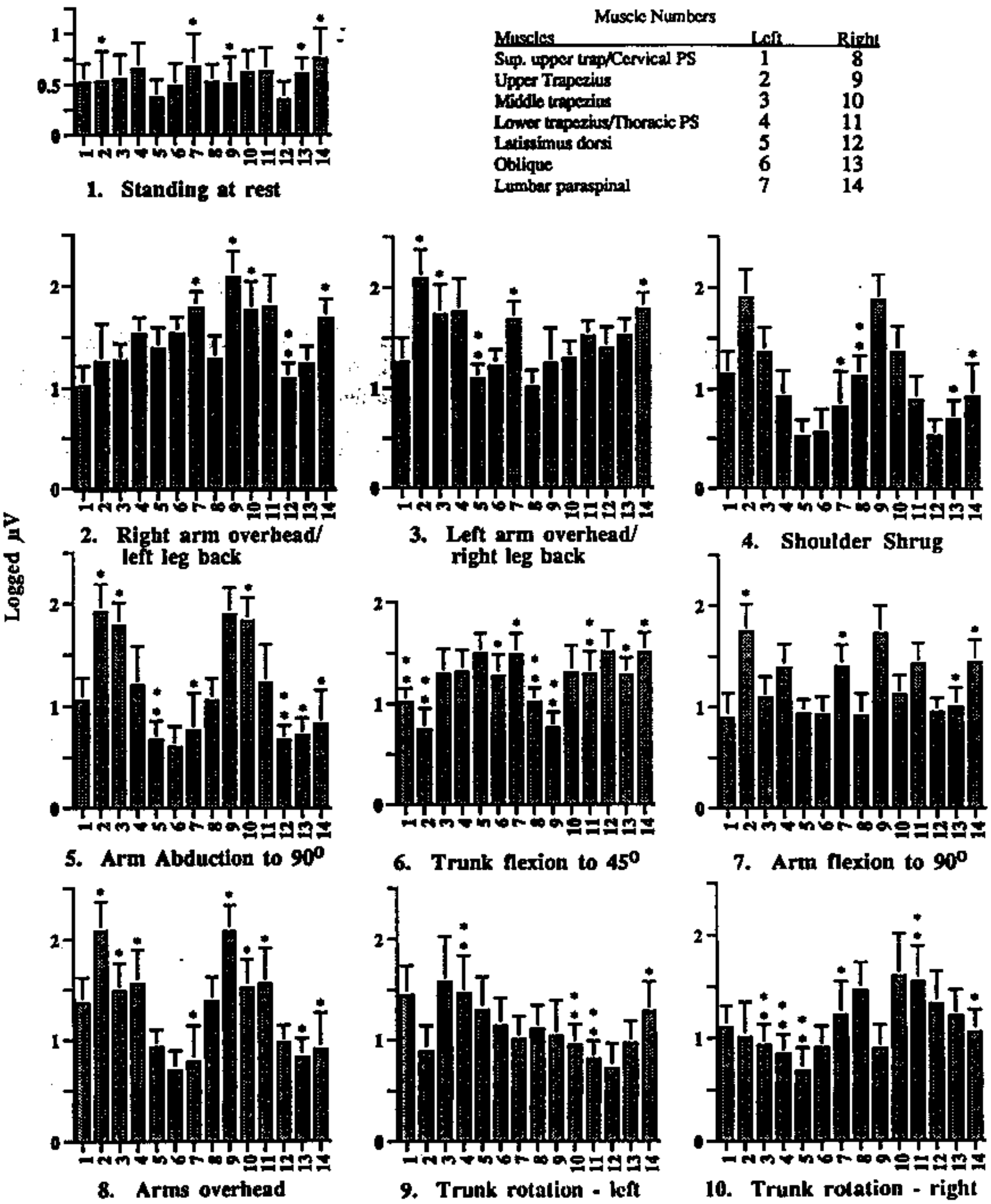
The mean transformed EMG values were very consistent between homologous muscle sites and confirm theoretical kinesiological assumptions pertaining to muscle recruitment. For example, during the end range of a shoulder shrug (Figs 3 and 4, motor task 4) and arm abduction to 90°

Table 1
Skinfold measurements, transformed EMG amplitudes with correction for adipose tissue, transformed EMG and percent attributed to adipose correction across 10 motor tasks for 100 males and 100 females

Muscle group	Caliper measures (mm)	Adipose corrected log EMG (µV)	Log EMG attributed to adipose correction (µV)	Percent attributed to adipose correction
Superior upper trapezius	7.2±2.5	1.09±0.33	0.09±0.03	8.3
Upper trapezius	5.9±2.4	1.40±0.61	0.07±0.03	5.0
Middle trapezius	10.5±3.7	1.32±0.42	0.13±0.05	9.8
Lower trapezius	10.8±4.0	1.27±0.44	0.13±0.05	10.2
Latissimus dorsi	11.8±5.8	0.96±0.41	0.15±0.07	15.6
Oblique	19.2±5.7	0.95±0.38	0.24±0.07	25.3
Lumbar paraspinal	10.7±4.8	1.11±0.47	0.13±0.06	11.7

Values are means±SD.

Logged Mean EMG - Males

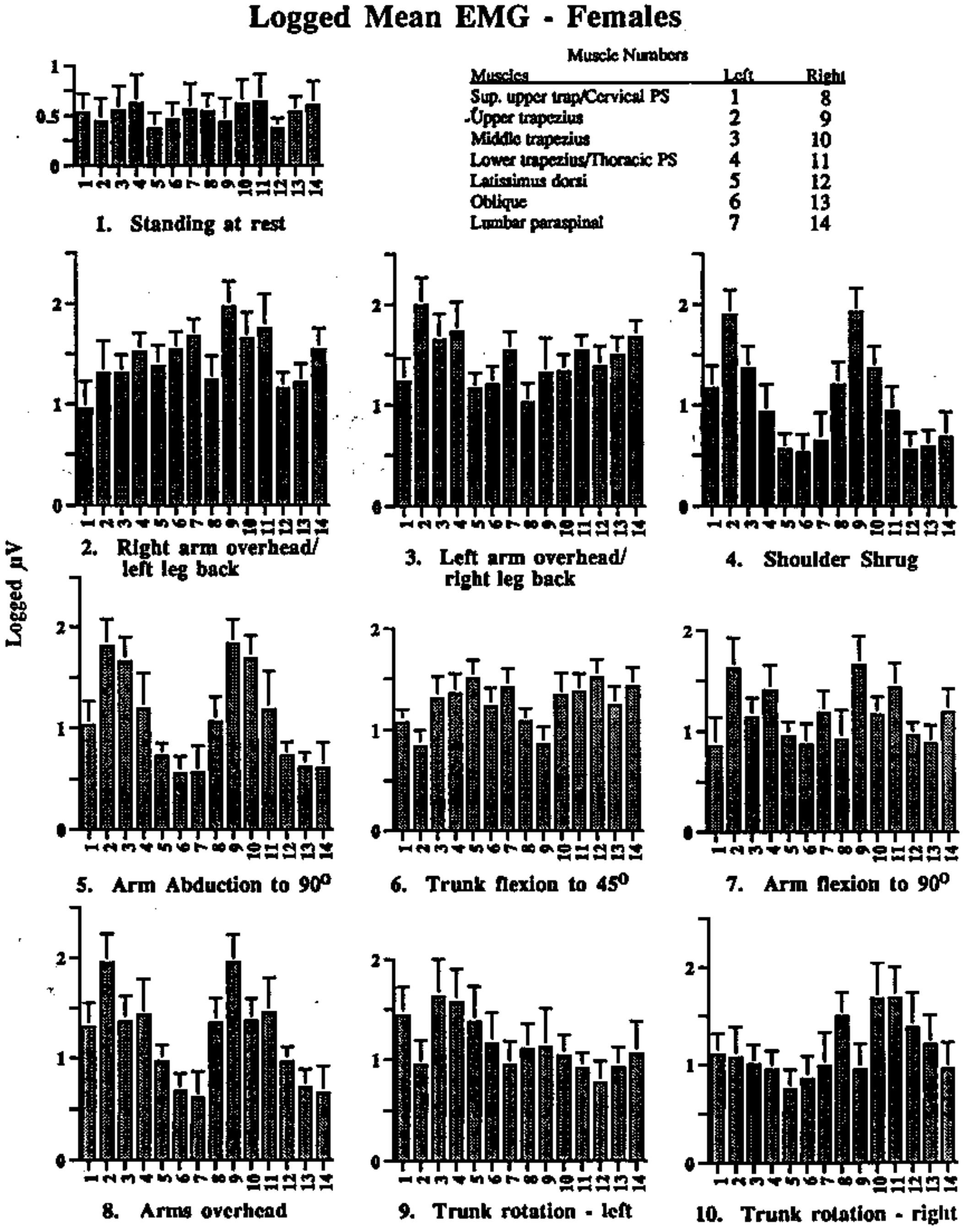


Muscle Numbers

Muscles	Left	Right
Sup. upper trap/Cervical PS	1	8
Upper Trapezius	2	9
Middle trapezius	3	10
Lower trapezius/Thoracic PS	4	11
Latissimus dorsi	5	12
Oblique	6	13
Lumbar paraspinal	7	14

1. Standing at rest
2. Right arm overhead/ left leg back
3. Left arm overhead/ right leg back
4. Shoulder Shrug
5. Arm Abduction to 90°
6. Trunk flexion to 45°
7. Arm flexion to 90°
8. Arms overhead
9. Trunk rotation - left
10. Trunk rotation - right

Fig. 3 Mean and standard deviations of the $\log_{10}$ of the EMG amplitudes of 14 muscles (see Fig. 2) during 10 motor tasks (see Fig. 1) as described in the Methods section. Significant gender differences are noted (*males>females; **males<females). Data derived from 100 male subjects. Super=superior, trap=trapezius, PS=paraspinal. (For reference, 2=$\log_{10}$ of 100 µV.)



Muscle Numbers

Muscles	Left	Right
Sup. upper trap/Cervical PS	1	8
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Middle trapezius	3	10
Lower trapezius/Thoracic PS	4	11
Latissimus dorsi	5	12
Oblique	6	13
Lumbar paraspinal	7	14

Fig. 4 Mean and standard deviations of the $\log_{10}$ of the EMG amplitudes of 14 muscles (see Fig. 2) during 10 motor tasks (see Fig. 1) as described in the Methods section. Data derived from 100 female subjects. Super = superior, trap = trapezius, PS = paraspinal. (For reference, 2 = $\log_{10}$ of 100 μV.)

(Figs 3 and 4, motor task 5), the EMG activity confirmed the theoretical assumption that the upper trapezius stabilised the shoulder girdle by elevating and upwardly rotating the scapula. Additional stabilisation was provided by the middle trapezius via abduction and, to a lesser extent, by the superior upper trapezius and lower trapezius muscles via upward rotation. The anticipated relative inactivity of the lower back muscles was reflected in the low mean transformed EMG activity. During arm flexion to 90° (Figs 3 and 4, motor task 7), it was anticipated that the lower trunk would need to maintain an erect posture to offset the weight of the arms forward. The relatively high mean transformed EMG amplitude from the lumbar paraspinal muscles confirm this recruitment pattern.

Differences between narrow and wide bandpass settings

The Pearson's correlation coefficient between transformed EMG amplitudes for both the narrow and wide bandpass settings for the latissimus dorsi of 14 subjects during trunk rotation to the left and arm abduction to 90° were 0.98 and 0.89 respectively (Fig. 5A and B). The intra-class correlations were less for trunk rotation (IC = 0.82) and substantially reduced during arm abduction to 90° (IC = 0.12). The relationship between the narrow and wide bandpass-transformed EMG amplitudes appears to be more predictable in motor tasks requiring relatively high EMG activities (Fig. 5A). When examining the relationship between narrow and wide bandpass settings and the average transformed EMG amplitudes across the seven muscle groups and 10 motor tasks, the Pearson's correlation was 0.98, with an intra-class correlation of 0.89. The wide bandpass EMG amplitudes were consistently greater than the narrow bandpass EMG amplitudes (Fig. 5A, B and C), with the largest differences occurring at the low amplitudes.

The differences in transformed EMG amplitudes between the narrow and wide bandpass settings were less than 18% above log 1 μV, 24% at log 0.75 μV, 37% at log 0.50 μV and 78% at log 0.25 μV (Fig. 5D). These EMG amplitudes correspond to approximately 10, 6, 3 and 2 μV, respectively.

The impact of narrow and wide bandpass settings seems to be much less apparent or minimal when examining the ratios of transformed EMG amplitudes. The EMG ratios presented in Fig. 5E produced a Pearson's correlation coefficient of 0.95, with an intra-class correlation of 0.93. The correlations among EMG ratios for narrow and wide bandpass settings during trunk rotation to the left (elevated EMG activity) for the 14 subjects ranged from 0.99 to 0.97. The range of intra-class correlations for trunk rotation to the left was 0.95–0.98. The range of Pearson correlations for arm abduction to 90° (low EMG activity) was 0.85–0.98, and the range for intra-class correlations was 0.59–0.90. These data demonstrate that the differences in EMG amplitudes observed between the narrow and wide bandpass settings have minimal effects on the $\log_{10}$ transformed ratios of EMG amplitudes.

Gender differences

The repeated measures analysis of variance showed significant gender differences. Two-tailed *t*-tests identified significant gender differences in 58 of the 140 $\log_{10}$ transformed EMG values and in 251 of the 910 transformed ratios. For example, Fig. 3 shows that the mean transformed EMG values for the lumbar paraspinal muscles were consistently greater in males than females. However, the average transformed EMG values for the latissimus dorsi were consistently greater in females than males. When examining the rank order of muscle recruitment for each of the motor tasks, the order of transformed EMG amplitudes was virtually identical. This result indicates that males and females recruit their back muscles similarly during the execution of the motor tasks.

Between-session reliability

The Pearson's correlation coefficients (*r*), as well as the intra-class correlation coefficients (IC), for the between-session data (Table 2) indicate that the transformed EMG amplitudes and trans-

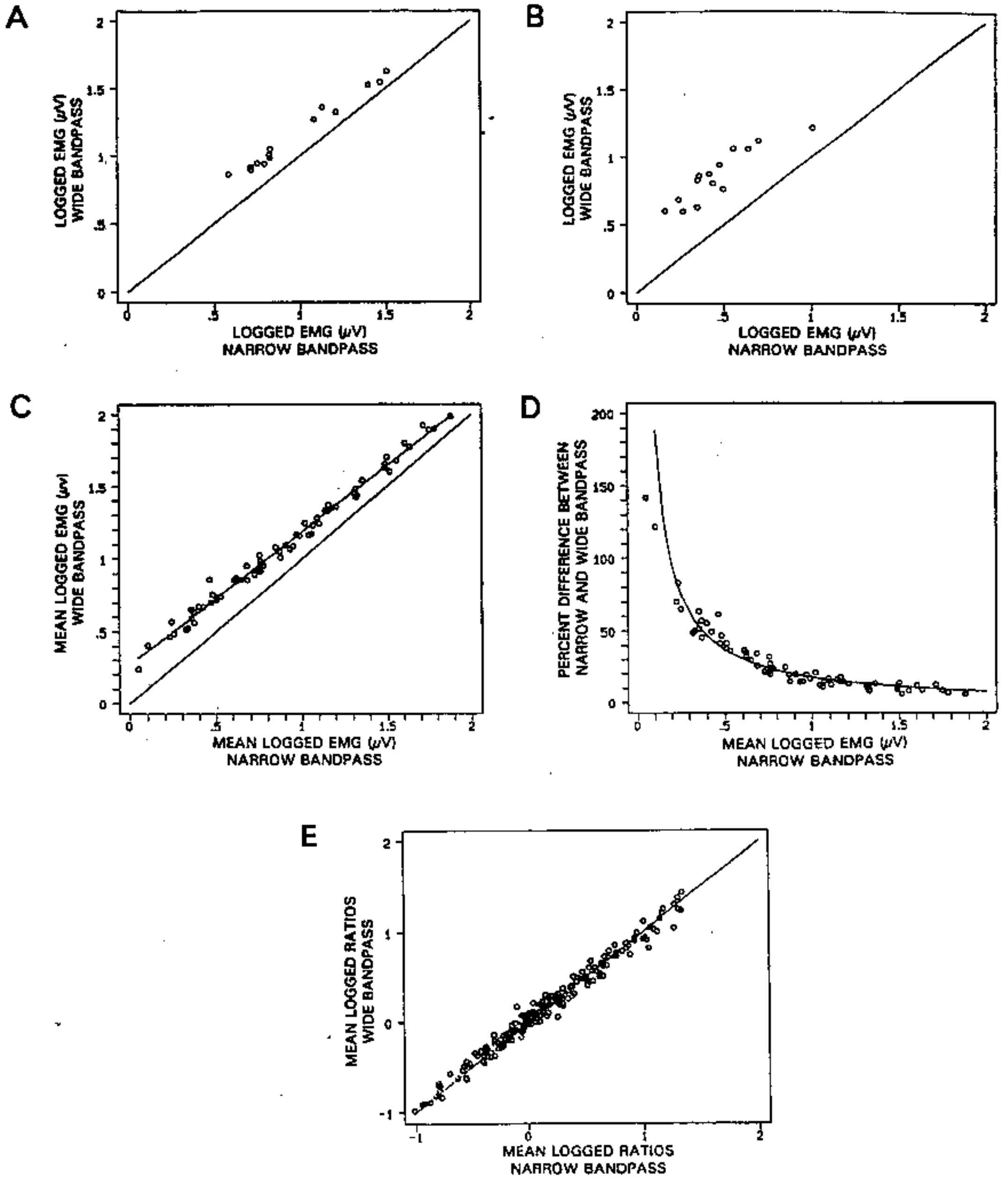

Fig. 5 Comparison of transformed EMG amplitudes between narrow and wide bandpass settings for latissimus dorsi during (A) trunk rotation to the left with greater expected EMG activity and (B) arm abduction to 90° with lower expected EMG activity. (C) Correlation between mean transformed EMG amplitudes for seven muscle groups across 10 motor tasks ($r=0.98$, IC$=0.89$). (D) Relationship between transformed EMG amplitude for the narrow bandpass and the percentage difference between the narrow and wide bandpass settings (best-fit equation: percent difference $= -0.021+0.195/\text{transformed EMG}_{\text{narrow bandpass}}$). (E) Correlation between the mean transformed ratios for narrow and wide bandpass settings ($r=0.95$, IC$=$ 0.93).

Table 2

Between-session correlations for transformed[a] EMG amplitudes and transformed ratios of EMG amplitudes for 14 muscle groups during each of 10 motor tasks for 20 males and 20 females

	Transformed EMG				Transformed ratios			
	Males		Females		Males		Females	
Motor tasks	*r*[b]	IC[c]	*r*	IC	*r*	IC	*r*	IC
1. Standing	0.63	0.61	0.61	0.59	0.66	0.65	0.60	0.60
2. R arm overhead/L leg back	0.94	0.93	0.90	0.90	0.94	0.93	0.92	0.91
3. L arm overhead/R leg back	0.94	0.93	0.89	0.88	0.94	0.94	0.90	0.89
4. Shoulder shrug	0.94	0.94	0.93	0.93	0.92	0.92	0.94	0.93
5. Arm abduction to 90°	0.96	0.95	0.95	0.94	0.95	0.95	0.94	0.94
6. Trunk flexion to 45°	0.92	0.90	0.88	0.86	0.90	0.89	0.88	0.87
7. Arm flexion to 90°	0.94	0.94	0.94	0.93	0.94	0.94	0.94	0.93
8. Arms overhead	0.92	0.92	0.96	0.96	0.90	0.90	0.94	0.94
9. Trunk rotation, left	0.80	0.79	0.83	0.82	0.83	0.82	0.83	0.81
10. Trunk rotation, right	0.83	0.80	0.75	0.74	0.86	0.85	0.81	0.79
Overall average	0.88	0.87	0.86	0.86	0.88	0.88	0.87	0.86

[a] Log$_{10}$ transformation; [b] Pearson product–moment correlation coefficient; [c] intra-class correlation coefficient. R, right; L, left.

formed ratios for each motor task were highly reliable, with the overall average correlations (*r* and IC) across all motor tasks ranging from 0.86–0.88 for both males and females. The most reliable transformed EMG amplitudes and transformed ratios were obtained during arm abduction to 90° and arm flexion to 90° (*r* and IC correlations ranged from 0.93 to 0.96) and the least reliable results were obtained while standing at rest (*r* and IC correlations ranged from 0.59 to 0.66).

EMG amplitudes *vs* ratios

An example of the ratios among selected muscles, including the left superior upper trapezius, upper trapezius, middle trapezius and lower trapezius muscles, during forward arm flexion to 90° (motor task 7) and arms overhead (motor task 8) are presented in Figs 6 and 7, respectively.

During arm flexion to 90° (motor task 7), the mean transformed upper trapezius EMG amplitude was significantly greater for males than for females, but no gender differences were observed in the superior upper trapezius/cervical paraspinal, middle trapezius and lower trapezius/thoracic paraspinal muscles (Fig. 3, motor task 7). The mean transformed ratios between the upper trapezius and all of its synergists exhibited significant gender differences (Fig. 6A), while the recruitment of the superior upper trapezius muscle relative to the middle and lower trapezius muscles was similar (Fig. 6B).

With arms overhead (motor task 8), the mean transformed EMG amplitudes from the upper trapezius, middle trapezius and lower trapezius muscles were significantly greater for males than for females, but no gender differences were observed in the superior upper trapezius muscle (Fig. 3, motor task 8). The mean transformed ratios of the upper trapezius and superior upper trapezius muscles were significantly different for males and females, but no gender differences were noted between the upper trapezius or superior upper trapezius muscle and the middle trapezius and lower trapezius muscles. The gender differences in the mean transformed amplitudes of the middle trapezius and lower trapezius muscles were not observed in the mean transformed ratios (Fig. 7A and B).

DISCUSSION

Since muscle recruitment patterns have been found to be highly variable during dynamic activity (Ahern et al, 1988; Nouwen et al, 1987; Wolf and Basmajian, 1978), we assessed static responses at the end range of well-defined motor

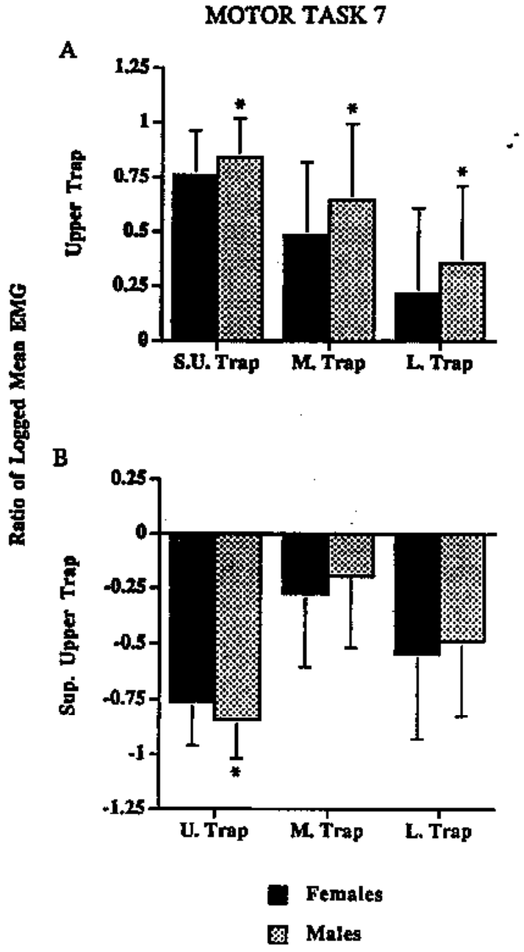

Fig. 6 Selected log ratios of EMG amplitudes from 100 female and 100 male subjects with arms flexed to 90° (motor task 7). (A) Mean and standard deviation of left upper trapezius with superior upper trapezius (S.U. Trap), middle trapezius (M. Trap) and lower trapezius (L. Trap). (B) Mean and standard deviation of left superior upper trapezius with upper trapezius, middle trapezius and lower trapezius. *Significant gender differences.

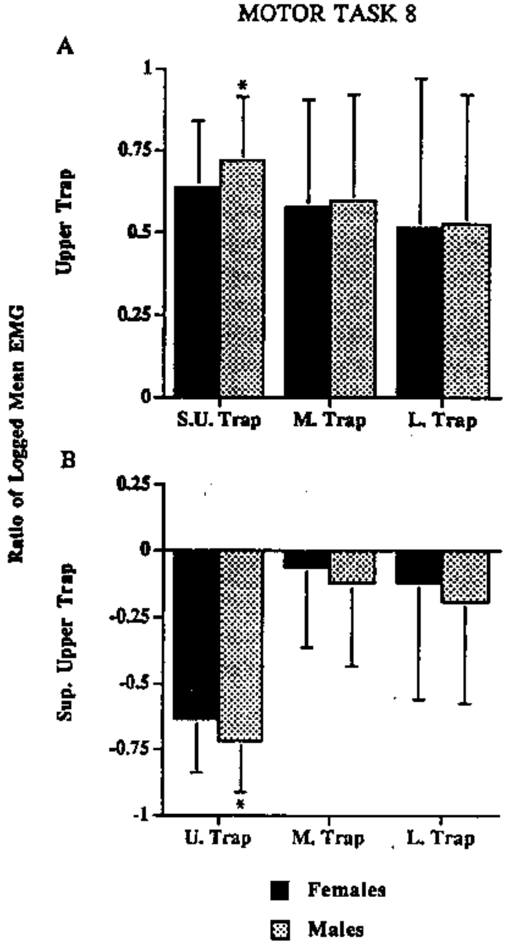

Fig. 7 Selected log ratios of EMG amplitudes from 100 female and 100 male subjects with arms raised overhead (motor task 8). (A) Mean and standard deviation of left upper trapezius with superior upper trapezius (S.U. Trap), middle trapezius (M. Trap) and lower trapezius (L. Trap). (B) Mean and standard deviation of left superior upper trapezius with upper trapezius, middle trapezius and lower trapezius. *Significant gender differences.

tasks in the present study. That EMG recordings were made from muscles with a constant length and shape (i.e. same joint position for a given motor task) probably improved reliability and stability of recruitment and recording condition during the performance of each motor task. In addition, by repeating each motor task four times, the potential variability for a given motor task, muscle and subject was reduced even more.

Maintaining the static end ranges of the motor tasks used in the present study requires a modest effort. This presumably makes these tests more practical and reliable than recordings derived from maximum voluntary contractions. For example, the influence of motivation will be less of a factor when the subject is not required to exert maximum effort. The tests used in this study only require the subject to offset the forces of gravity.

These tasks also minimise the possibility of exacerbating muscle injury.

The mean transformed EMG amplitudes presented in Figs 3 and 4 were corrected for signal attenuation attributed to adipose tissue, since this procedure has been suggested to reduce between-subject variability for EMG amplitudes (Hemingway et al, 1995). The correction resulted in a mean increase in transformed EMG amplitudes across seven bilateral muscle sites and 10 motor tasks of 12%, ranging from 5% for the upper trapezius to 25% for the obliques.

The between-session transformed EMG amplitudes and ratios were highly reliable across the nine active motor tasks (*r* and IC correlations ranged from 0.74 to 0.96; Table 2). However, the transformed EMG amplitudes and ratios were substantially less reliable during standing (*r* and IC correlations ranged from 0.59 to 0.66; Table 2) when low EMG activities were recorded (see Figs 3 and 4).

Similar conclusions were arrived at for datasets which had been processed using a wide bandpass setting or a narrow bandpass setting. The latter was used to remove the low-level ECG signal which could otherwise contribute to inaccuracies in the measurement of very low-level EMG activity. The selections chosen for the total bandwidth of the EMG signal clearly did not alter our basic observations and greatly facilitated analysis of the data. Our data suggest that although frequency bands below 66 Hz add significant levels of information to the raw EMG signal, there is sufficient information in the remaining bands to perform our analyses. Other authors have noted the significance of low-frequency bands in the discrimination of changes in the activity of spinal muscles (Dolan, Mannion and Adams, 1995), but the median frequencies in spinal muscles are in the range 80–130 Hz, suggesting that a majority of the EMG signal power is well above 66 Hz (Thompson and Biederman, 1993; Tsuboi et al, 1994). Furthermore, the impact of the bandpass settings on the ratios of EMG values appears to be minimal.

Crosstalk, the contribution of activity in neighbouring muscles to the EMG activity recorded in the muscle of interest, can be problematic in some circumstances (Hof and Van der Berg, 1981; Loeb and Gans, 1986; DeLuca and Merletti, 1988; Merletti, Knaflitz and DeLuca 1992), although proper electrode placement and appropriate recording electronics can help to minimise the problem (Basmajian and DeLuca, 1985, pp. 42–45; Koh and Grabiner, 1993). In our experiments, electrode placement overlay muscle groups and we made no assumption about an individual muscle within a group, other than that they share a similar kinesiological function during the execution of the motor tasks. Nevertheless, our results show that our choice of electrode sites provides information which differentiates recruitment among muscles and across motor tasks. Furthermore, we have shown in previous publications that this recording protocol effectively identifies the presence or absence of low back pain (Edgerton et al, 1996a,b).

Significant gender differences were reported for all 10 motor tasks (Fig. 3). Males exhibited greater EMG activity than females in the lumbar paraspinal muscles during all motor tasks except the left lumbar paraspinal muscle in trunk rotation to the left (motor task 9). These findings are consistent with Nouwen et al (1987), who reported lower EMG levels for females than males in the paraspinal muscles during flexion/extension, lateral bending and rotation, and with Wolf et al (1979), who reported significantly greater activity for males than females during rotation, stooping and squatting. The significantly higher EMG activity observed in both the upper and middle trapezius of males during motor tasks with the arms flexed beyond 135° (i.e. right arm in motor task 2, left arm in motor task 3 and both arms in motor task 8) may be attributable to increased upper torso muscle development. The significant differences in EMG activity of the superior upper and upper trapezius during forward trunk flexion (Fig. 3, motor task 6) suggest that females retract their shoulders slightly more than males to offset the weight of the flexed upper torso.

To maintain arms overhead, the mean transformed EMG amplitudes of the upper, middle and lower trapezius muscles were greater for males than females (Fig. 3, motor task 8). However, no gender differences were observed in the transformed ratios between these muscles, suggesting the relative contribution of these muscles to the task were similar for males and females (Fig. 7A). Thus it appears that some of the morphological and gender differences that may be reflected in EMG amplitudes may be minimised by using ratios of EMG amplitudes.

An example of abnormal patterns of activation in response to a back injury may be observed during trunk rotation with the pelvis stabilised (i.e. motor tasks 9 and 10). The theoretical assumption that the ipsilateral erector spinae, ipsilateral external oblique and contralateral internal oblique should be active during trunk rotation based on kinesiological considerations are reflected in the transformed EMG values (Figs 3 and 4). The mean transformed paraspinal EMG activity of the lumbar paraspinal muscles contralateral to the rotational motor task were greater than the ipsilateral side, confirming previous findings (Ahern et al, 1988; Basmajian, 1978; Kravitz et al, 1981; Wolf et al, 1991; Wolf et al, 1979) that the contralateral paraspinal muscles stabilise the vertebral column during rotation. However, the standard deviations of the transformed EMG amplitudes suggest that the manner by which normal subjects recruit the obliques and lumbar paraspinal muscles relative to each other during trunk rotation is variable, and may be attributed to the difficulty in maintaining the pelvis in a stable position while the trunk rotates. However, these muscles were reliably characterised by transformed ratios ($r = 0.83$, $IC = 0.81$).

Within our database, the age range was 18–61 years. This age distribution seems to be appropriate to establish normalcy of back muscle recruitment patterns in adults. Prior to the age of 60 there is no clear evidence that the EMG signal is significantly affected or that the distribution of fibre type along with neuromuscular function are altered (Ayoagi and Sheppard, 1992). Direct analysis of our data comparing young (<28 years, $n = 17$) and old (>49 years, $n = 17$) groups matched for sex and body mass index yielded no significant differences between ratios of muscle group EMG activity. There was a tendency for increased EMG activity in the lumbar paraspinal muscles for the older group. This tendency was translated into those ratios involving the lumbar paraspinal muscle group (i.e. ratios with the lumbar paraspinal muscle group in the numerator tended to be higher). However, none of the muscle ratios examined for the older group approached abnormality based on a 95% confidence interval for our population data.

CONCLUSION

Between-session EMG amplitudes and ratios of EMG amplitudes for seven different bilateral muscle groups during 10 different motor tasks were found to be highly reliable. Significant differences in EMG amplitudes between 100 males and 100 females were observed. Ratios of EMG amplitudes from a battery of muscles were found to provide unique and consistent information pertaining to relative recruitment of muscle groups in a large population of normal males and females. These data demonstrate the feasibility of using a normative database as an objective reference for identifying abnormal muscle patterns in injured patients. Although further studies are needed to define more clearly the population characteristics of EMG patterns associated with subjects of different ages, particularly the elderly, a variety of well-defined aetiologies involving back pain and muscle dysfunction can be compared with the present normative database.

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References

Ahern DK, Follick MJ, Council JR, Laser-Wolston N 1986 Reliability of lumbar paravertebral EMG assessment in chronic low back pain. Archives of Physical Medicine and Rehabilitation 67: 762–765

Ahern DK, Follick MJ, Council JR, Laser-Wolston N, Litchman H 1988 Comparison of lumbar paravertebral EMG patterns in chronic low back pain patients and non-patient controls. Pain 34: 153–160

Arena JG, Sherman RA, Bruno GM, Young TR 1989 Electromyographic recordings of 5 types of low back pain subjects and non-pain controls in different positions. Pain 37: 57–65

Arena JG, Sherman RA, Bruno GM, Young TR 1990 Temporal stability of paraspinal electromyographic recordings in low back pain and non-pain subjects. International Journal of Psychophysiology 9: 31–37

Arena JG, Sherman RA, Bruno GM, Young TR 1991 Electromyographic recordings of low back pain subjects and non-pain controls in six different positions: Effects of pain levels. Pain 45: 23–28

Ayoagi Y, Shephard RJ 1992 Aging and muscle function. Sports Medicine 14: 376–396

Basmajian JV 1978 Cyclobenzaprine hydrochloride effect on skeletal muscle spasm in the lumbar region and neck: Two double-blind controlled clinical and laboratory studies. Archives of Physical Medicine and Rehabilitation 59: 58–63

Basmajian JV, DeLuca CJ 1985 Muscles alive, 5th edn. Williams and Wilkins, Baltimore, MD

Cohen MJ, Naliboff BD, Schandler SL, McArthur C 1986 Comparison of electromyographic response patterns during posture and stress tasks in chronic low back pain patterns and controls. Journal of Psychosomatic Research 30: 131–141

Collins GA, Cohen MJ, Haliboff BD, Schandler SL 1982 Comparative analysis of paraspinal and frontalis EMG, heart rate and skin conductance in chronic low back pain patients and normals to various postures and stress. Scandinavian Journal of Rehabilitation Medicine 14: 39–46

Cram JR, Engstrom D 1986 Patterns of neuromuscular activity in pain and non-pain patients. Clinical Biofeedback and Health 9: 106–115

Cram JR, Steger JC 1983 EMG scanning in the diagnosis of chronic pain. Biofeedback and Self Regulation 8: 229–241

De la Berra EJ, Milner TE 1994 The effects of skinfold thickness on the selectivity of surface EMG. Electroencephalography and Clinical Neurophysiology 93: 91–99

DeLitto A 1994 Are measures of function and disability important in low back care. Physical Therapy 74: 452–462

DeLuca CJ, Merletti R 1988 Surface myoelectric signal and cross-talk among muscles of the leg. Electroencephalography and Clinical Neurophysiology 69: 568–576

DeLuca CJ 1993 Use of the surface EMG signal for performance evaluation of back muscles. Muscle and Nerve 16: 210–216

Dolan P, Mannion AF, Adams MA 1995 Fatigue of the errector spinae muscles: A quantitative assessment using 'frequency banding' of the surface electromyography signal. Spine 20: 149–159

Dolce JJ, Raczynski JM 1985 Neuromuscular activity and electromyography in painful backs, psychological and biomechanical models in assessment and treatment. Psychological Bulletin 97: 502–520

Edgerton VR, Wolf SL, Levendowski DJ, Roy RR 1996a Evaluating patterns of EMG amplitudes for trunk and neck muscles of patients and controls. International Journal of Rehabilitation and Health 2: 1–8

Edgerton VR, Wolf SL, Levendowski DJ, Roy RR 1996b Theoretical basis for patterning EMG amplitudes to assess muscle dysfunction. Medicine and Science in Sports and Exercise 28: 744–751

Hemingway MA, Bidermann HJ, Inglis J 1995 Electromyographic recordings of paraspinal muscles: Variations related to subcutaneous tissue thickness. Biofeedback and Self-Regulation 20: 39–49

Hof AL, Van der Berg JW 1981 EMG force processing; I–IV. Journal of Biomechanics 14: 747–792

Hoyt WH, Hunt HH, DePauw MA, Bard D, Shaffer F, Pasias JN, Robbins DH, Runyon DG, Semrad SE, Symonds JT, Watt KC 1981 Electromyographic assessment of chronic low-back pain syndrome. Journal of the American Osteopathic Association 80: 57–59

Jayasinghe WJ, Harding RH, Anderson JAD, Sweetman BJ 1978 An electromyographic investigation of postural fatigue in low back pain – a preliminary study. Electromyography and Clinical Neurophysiology 18: 191–198

Kibler WB 1990 Clinical aspects of muscle injury. Medicine and Science in Sports and Exercise 22: 450–452

Koh TJ, Grabiner MD 1993 Evaluation of methods to minimize cross-talk in surface electromyography. Journal of Biomechanics 26 (suppl 1): 151–157

Kravitz E, Moore ME, Glaros A 1981 Paralumbar muscle activity in chronic low back pain. Archives of Physical Medicine and Rehabilitation 62: 172–176

Loeb GE, Gans G 1986 Electromyography for experimentalists. University of Chicago Press, Chicago, IL

Mertetti R, Knaflitz M, DeLuca CJ 1992 Electrically evoked myoelectric signals. Critical Reviews in Biomechanics and Engineering 19: 293–340

Moore DS 1995 The basic practice of statistics. Freeman, New York

Nouwen A, Bush C 1984 The relationship between paraspinal EMG and chronic low back pain. Pain 20: 109–123

Nouwen A, Van Akkerveeken PF, Versloot JM 1987 Patterns of muscular activity during movement in patients with chronic low-back pain. Spine 12: 777–782

Ortengren R, Andersson GB 1977 Electromyographic studies of trunk muscles, with special reference to the functional anatomy of the lumbar spine. Spine 2: 44–52

Price JP, Clare MH, Ewerhardt RH 1948 Studies of low backache with persistent muscle spasm. Archives of Physical Medicine and Rehabilitation 29: 703–709

Roy RR, Baldwin KM, Edgerton VR 1991 The plasticity of skeletal muscle: Effects of neuromuscular activity. Exercise and Sports Science Reviews 19: 269–302

Sihvonen T, Partanen J, Hanninen O 1988 Averaged surface EMG in testing back function. Electromyography and Clinical Neurophysiology 28: 335–339

Sihvonen T, Partanen J, Hanninen O, Soimakallio S 1991 Electric behavior of low back muscles during lumbar pelvic rhythm in low back pain patients and healthy controls. Archives of Physical Medicine and Rehabilitation 72: 1080–1087

Snedecor GW, Cochran WG 1967 Statistical methods, 6th edn. Iowa State University Press, Ames, IA

Solomonow M, Baten C, Smit J, Baratta R, Hermens H, D'Ambrosia R, Shoji R 1990 Electromyogram power spectra frequencies associated with motor unit recruitment strategies. Journal of Applied Physiology 68: 1177–1185

Thompson DA, Biederman HJ 1993 Electromyographic power spectrum analysis of the paraspinal muscles. Spine 18: 2310–2313

Tsuboi T, Satou T, Egawa K, Izumi Y, Miyazaki M 1994 Spectral analysis of the electromyogram in lumbar muscle fatigue induced endurance contraction. European Journal of Applied Physiology 69: 361–366

Wolf LB, Segal RL, Wolf SL, Nyberg R 1991 Quantitative analysis of surface and percutaneous electromyographic activity in lumbar erector spinae of normal young women. Spine 16: 155–161

Wolf SL, Basmajian JV 1978 Assessment of paraspinal electromyographic activity in normal subjects and in chronic back pain patients using a muscular biofeedback device. International Series in Biomechanics IVB. University Park Press, Baltimore, MD

Wolf SL, Basmajian JV, Russ TC, Kutner M 1979 Normative data on low back mobility and activity levels. American Journal of Physical Medicine 58: 217–229

Zar JH 1984 Biostatistical analysis, 2nd edn. Prentice-Hall, Englewood Cliffs, NJ