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Relationships of EMG to effort in the trunk under isometric conditions: force-increasing and decreasing effects and temporal delays

Ian A.F. Stokes

Department of Orthopaedics and Rehabilitation, University of Vermont, Stafford Hall 434, Burlington, VT 05405-0084, USA

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Abstract

Background. Electromyograms are used in increasingly sophisticated biomechanical analyses to estimate forces within the trunk to prevent and evaluate painful spinal conditions. However, even under nominally isometric conditions the relationship between EMG and effort is complex. This study quantified influences of pulling direction, increasing versus decreasing effort and electromechanical delay on the EMG/effort relationships for principal lower trunk muscle groups in isometric pulling tasks, to determine whether the observed differences between increasing versus decreasing effort relationships were consistent with electromechanical delay or activation differences.

Methods. Twenty-three healthy subjects (15 male, 8 female; mean age 32 years; mean bodymass 74.5 kg) each stood in an apparatus to stabilize the pelvis and performed ramped isometric efforts with a harness around the thorax connected to each of a series of five anchor points on the wall, for angles of pull at each 45° increment from 0° to 180° to the anterior direction. A load cell recorded the generated force for a 5 s timed increase up to a voluntary maximum, a 1 s 'dwell', and a 5 s relaxation back to zero effort. EMG signals were recorded via electrodes (surface, except indwelling for multifidus) from right and left rectus abdominis, internal and external obliques, longissimus, iliocostalis and L2 and L4 level multifidus. EMG signals were rectified with a 250 ms root-mean-square moving average filter. Effort-increasing and effort-decreasing sections of recordings were analyzed separately.

Findings. The EMG/effort relationship had a statistically significantly greater gradient as the effort was increasing than when decreasing for 28 of 70 muscle-angle permutations. This difference in gradient was found to explain a significant part of the apparent lag between effort generated and EMG signal that averaged between 261 and 658 ms before and between 31 and 196 ms for different muscles after the slope difference was taken into account.

Interpretation. The findings were consistent with the notion that the motor unit recruitment differs in increasing versus decreasing isometric efforts, probably because of a small stretching of the muscle as its tension increases. The residual temporal delay was thought to represent electromechanical delay.

Keywords: Electromyography; Trunk-muscles; Electromechanical delay; Muscle activation; Lumbar spine

1. Introduction

The magnitudes of the forces that act on the lumbar spine depend on the degree of muscle activation. The pattern of muscle activation must be compatible with force equilibrium, but since there is a 'redundant' number of muscles compared to the number of degrees of freedom they control, the muscle forces cannot be calculated uniquely. The individual muscle forces may be estimated by analytical models that optimize a 'cost function' that represents the presumed strategy of the

E-mail address: Ian.Stokes@uvm.edu

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central nervous system. To add realism to biomechanical analyses, and to avoid the need to specify the cost function, electromyographic (EMG) measurements are often employed to provide information on the degree of activation of muscles. These models are commonly referred to as being 'EMG-assisted' (Cholewicki and McGill, 1994; Granata and Marras, 1995a,b), or EMG-driven (Sparto et al., 1998).

However, there is a complex relationship between EMG and muscle force (Solomonow et al., 1990; Baratta et al., 1993). In order to estimate muscle force from an EMG signal, the signal may be expressed as a proportion of that recorded at maximum activation (in a maximum voluntary effort). Then the force can be estimated as the product of this normalized value, the muscle cross-sectional area, and the 'specific stress', i.e. the maximum force generated per unit cross section (Cholewicki and McGill, 1994; Sparto et al., 1998; Granata and Marras, 1995a; van Dieën and Visser, 1999). The specific stress varies with the degree of muscle pennation (Kaufman et al., 1989). This represents a linear and time-independent representation of the EMG/force relationship. A more physiological representation of the muscle force includes corrections for the muscle length, shortening velocity, posture (Mouton et al., 1991), and fatigue (Dolan and Adams, 1993; Potvin et al., 1996). It is also known that there is a time lag between the EMG signal and the generated force, often called electromechanical delay (Cavanagh and Komi, 1979; Thelen et al., 1994; Vos et al., 1991; van Dieën et al., 1991). A further practical difficulty in deducing muscle forces from EMG signals is the presence of 'crosstalk' whereby an EMG electrode records a signal from numerous muscles, and is not specific to the activity of the intended muscle over which it is placed.

It has been noted that the EMG/effort relationship differs, depending on whether the effort is increasing or decreasing (Stokes et al., 1987). This has implications for estimating forces from EMG, but the origin of this effect is not clear, but there are at least three plausible explanations:

1. Differing recruitment of motor units as is observed in lengthening and shortening activations (Joyce et al., 1969). Even in nominally isometric efforts, it is likely that there is shortening of the muscle as the effort increases and vice versa, as a result of series elasticity in the muscle, tendon and other structures.
2. Recruitment at the whole muscle level—the CNS might transfer force generation between different parallel muscles when the task changes from increasing to decreasing force generation. (However, if this were the case, one would expect to see some muscles having increased, and others decreased activation after maximum effort were achieved.)
3. Electromechanical delay (Cavanagh and Komi, 1979; Thelen et al., 1994; Vos et al., 1991; van Dieën et al., 1991). The time lag between the EMG signal and the force generated gives the appearance of hysteresis when EMG is plotted graphically against effort.

This study empirically determined the relationship between the effort (external resisted force) and the EMG signal from seven pairs of trunk muscles in nominally isometric conditions. The recordings were examined for nonlinearities, synergies of muscular recruitment for the different pulling directions, differences in EMG–effort relationships during increasing versus decreasing effort, and temporal lags between EMG and effort. The purpose of this study was to quantify influences of pulling direction, increasing versus decreasing effort and electromechanical delay on the EMG/effort relationships for principal lower trunk muscle groups in isometric pulling tasks in which subjects generated an external force (effort) acting horizontally at different angles, and to determine whether the observed differences between increasing versus decreasing effort relationships were consistent with electromechanical delay or activation differences.

2. Methods

Twenty-three subjects (Table 1) who reported no recent (prior year) back pain were tested while each subject stood in an apparatus with the pelvis immobilized (Fig. 1). They were asked to performed 'ramped-effort' tests with a 5 s timed increase up to a voluntary maximum effort, a 1 s 'dwell', and a 5 s relaxation back to zero effort. Resistance was provided by a horizontal cable from a harness around the thorax to one of five anchorage points on a wall track to the subject's right side at angles of 0°, 45°, 90°, 135° and 180° to the forward direction (Fig. 1b). The sequence of angles was randomly selected. The cable was aligned approximately horizontally and at the level of the T-12 vertebra. Three trials were performed at each angle. A computer screen in front of the subject displayed a vertical bar whose height was proportional to the effort generated, and with a mark to indicate the prior maximum effort.

EMG signals from seven right and left pairs of trunk muscles were recorded, using bipolar EMG electrodes (Delsys Inc. Type DE-02.3, Boston, MA USA). These

Table 1
Details of subjects studied

	Age (years)	Height (m)	Body mass (kg)
Female ($n = 8$)	33.5 (13.2)	1.63 (0.1)	60.7 (10.6)
Male ($n = 15$)	30.3 (9.1)	1.68 (0.5)	81.8 (14.2)

Mean values (with standard deviation in parentheses) are presented.

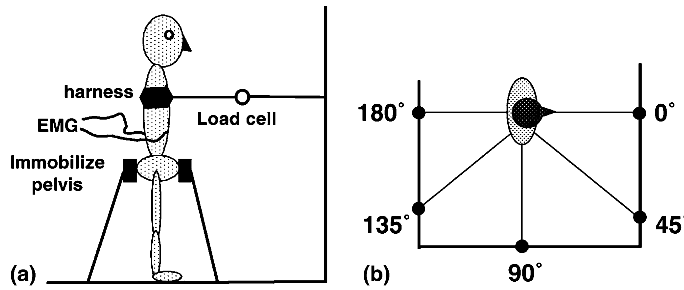

Fig. 1. Diagram of subject and testing configuration (a) side view; (b) plan view.

muscles were: rectus abdominis, internal and external obliques, longissimus, iliocostalis (all surface electrodes), and multifidus at the levels L2 and L4 (indwelling electrodes). The Delsys electrodes have 10×1 mm silver bar electrodes with 10 mm spacing; their single differential amplifiers have a gain of 1000, bandwidth 20–450 Hz, 92 dB (typical) common mode rejection ratio, and $10^{12}\,\Omega$ input impedance. A ground electrode was placed over the lateral aspect of the elbow. EMG and load cell signals were recorded digitally at 2048 Hz.

Electrodes were placed as follows: rectus abdominis-30 mm lateral to the midline at the level of the umbilicus, aligned vertically; external oblique-halfway between the iliac crest and the 12th rib along the mid-axillary line, aligned at an 80° angle to the horizontal; internal oblique 20 mm medial and superior to the anterior superior iliac spine, aligned vertically; longissimus-30 mm lateral to the midpoint of the spinous process of L3, aligned vertically; iliocostalis-60 mm lateral to the midpoint of the spinous process of L3, aligned vertically. Suspect data resulting from technical problems such as loose electrodes, or electrocardiogram artifacts were excluded from further statistical analyses. These records were eliminated by a dual process of visual inspection, and identification of outliers. Multifidus electrodes were custom made fine wire electrodes inserted with a 24-gauge hypodermic needle, after identifying an insertion point and insertion depth from ultrasound images as described in Stokes et al. (2003). In subsequent processing of the data from each of the ramped-effort tests, the EMG signals were passed through a root-mean-square (RMS) filter with a moving window having a width of 250 ms.

Each recording was divided into two segments corresponding to the effort-increasing and the effort-decreasing segment of the recording. These segments were defined by threshold values corresponding to 10% of the generated force range above the minimum and 10% of the generated force range below the maximum. A regression analysis between the RMS-EMG signal and the effort generated was performed separately for the effort-increasing and effort-decreasing segments. The gradient of each EMG/effort relationship provided a measure of each muscle's activation during each test. These gradients were termed 'upslope' and 'downslope' for the two segments. Each gradient was expressed non-dimensionally by multiplying by the maximum effort for that testing angle, and dividing by the maximum EMG value for that muscle obtained from all the ramped effort tests of the corresponding subject.

Based on simplified biomechanics, the activation magnitude of any muscle was expected to vary sinusoidally with the angle of effort. This expected sinusoidal relationship between muscle activation and effort direction was examined by plotting the 'upslope' averaged over the 23 subjects and three trials versus the external effort angle (Θ). Then a sine wave function of the form

$$\text{upslope} = A + B\sin(\Theta + \varphi)$$

was fitted by least-squares to these data, with sine wave amplitude (B) and constant (A) and phase angle (φ) as parameters. In these analyses, the data from right and left muscle pairs were combined, based on presumed symmetry, to provide angle data from 0° to 360° in 45° increments.

The temporal relationship between the EMG and effort signals was examined initially by identifying the maximum values of the effort and the filtered EMG. The time difference between these two events was recorded and termed 'Time delay'. The 'Time shift' attributed to electromechanical delay was determined by cross-correlation analysis of the EMG and the effort. This analysis determined the time shift that maximized the correlation coefficient between the effort and EMG signals (Fig. 2). The cross-correlation was performed alone, and after correcting the EMG/effort relationship by an empirically determined slope factor (Fig. 2b and c). This was done because the 'hysteresis' in the EMG–force recordings was thought to result from a combination of both electromechanical delay, and differing recruitment for force-increasing and force-decreasing segments of the recordings. The slope factor was obtained by identifying the value of a factor (in the range 0.5–1.5) that when multiplied by the EMG values recorded before the maximum effort maximized the correlation coefficient between the effort and EMG signals (Fig. 2b).

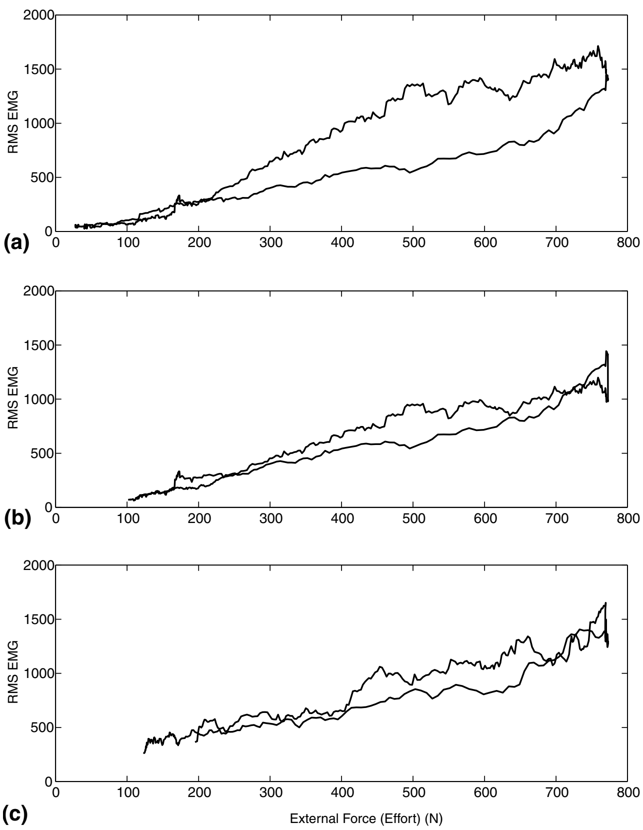

Fig. 2. Sample RMS-EMG/effort relationship (left iliocostalis of Subject 17, at 90° effort direction): (a) Data as recorded; (b) with slope adjusted as determined by cross-correlation analysis and (c) after adjusting for slope factor and for time delay as determined by cross-correlation analysis.

3. Results

The values of the maximum efforts averaged (SD in parentheses) 509 (198) N, 542 (193) N, 513 (201) N, 528 (218) N, and 544 (256) N at angles of 0°, 45°, 90°, 135° and 180°, respectively. The angle of maximum activation of each muscle (as assessed by the maximum 'upslope') followed the pattern expected from the presumed mechanical advantage of each muscle. The maximum upslope was observed at 0° for all dorsal muscles except left longissimus and left iliocostalis (maximum at 45°). It was at 180° for all right abdominal muscles; at 90° for left internal and external obliques and 135° for left rectus abdominis.

The RMS-EMG/effort was nearly linear, as expected. In regression analyses for each muscle at its most active angle, adding a quadratic term to the RMS-EMG/effort relationship increased the average R^2 from 84% to 89%.

The mean upslope was greater than the mean downslope in all muscle-angle permutations (Fig. 3). The differences between upslope and downslope were statistically significant for 28 of 70 muscle-angle permutations.

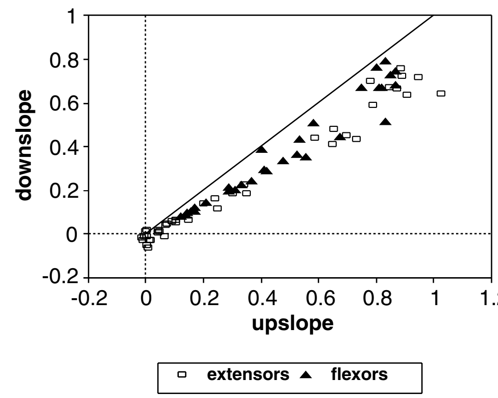

Fig. 3. Upslope versus downslope. Each data point represents the mean for one muscle averaged over 23 subjects and three trials. The upslope was always larger than the downslope. The solid line represents equality between upslope and downslope.

The slope-angle relationship approximated the expected sinusoidal relationship (Fig. 4). However, there was evidence of antagonistic activation at angles where the sinusoidal relationship predicted little or no activation. All abdominal muscles demonstrated significantly ($p < 0.05$) increasing activation with increasing effort at all angles; therefore these muscles were always activated either as agonists or as antagonists.

The temporal delay between the times of maximum EMG and maximum generated effort was observed to average between 261 and 658 ms for different muscles (Table 2). After the EMG/effort was adjusted for the slope difference, the time shift was significantly less in all muscles, and the correlation coefficient between EMG and effort was greater in all cases (Table 3). Slightly greater values of temporal shift were observed for dorsal muscles (mean 93 ms) than abdominal muscles (mean 89 ms) (Table 2).

4. Discussion

Even in nominally isometric conditions, the observed relationship between EMG and external effort depended on whether the effort was increasing or decreasing, and included a time lag. The EMG–effort slope relationship with angle of pull indicated that there was considerable antagonistic coactivation, especially in abdominal muscles, as has been reported by Granata and Marras, 1995b, Potvin and O'Brien (1998) Thelen et al., 1995 and Lavender et al., 1992.

After the EMG/effort was adjusted for the slope difference, the time shift was substantially less, and close to values reported for electromechanical delay in trunk muscles (Thelen et al., 1994; van Dieën et al., 1991), and for leg muscles (Cavanagh and Komi, 1979; Vos et al., 1991). When the differing EMG–effort relationship for increasing or decreasing effort was not taken

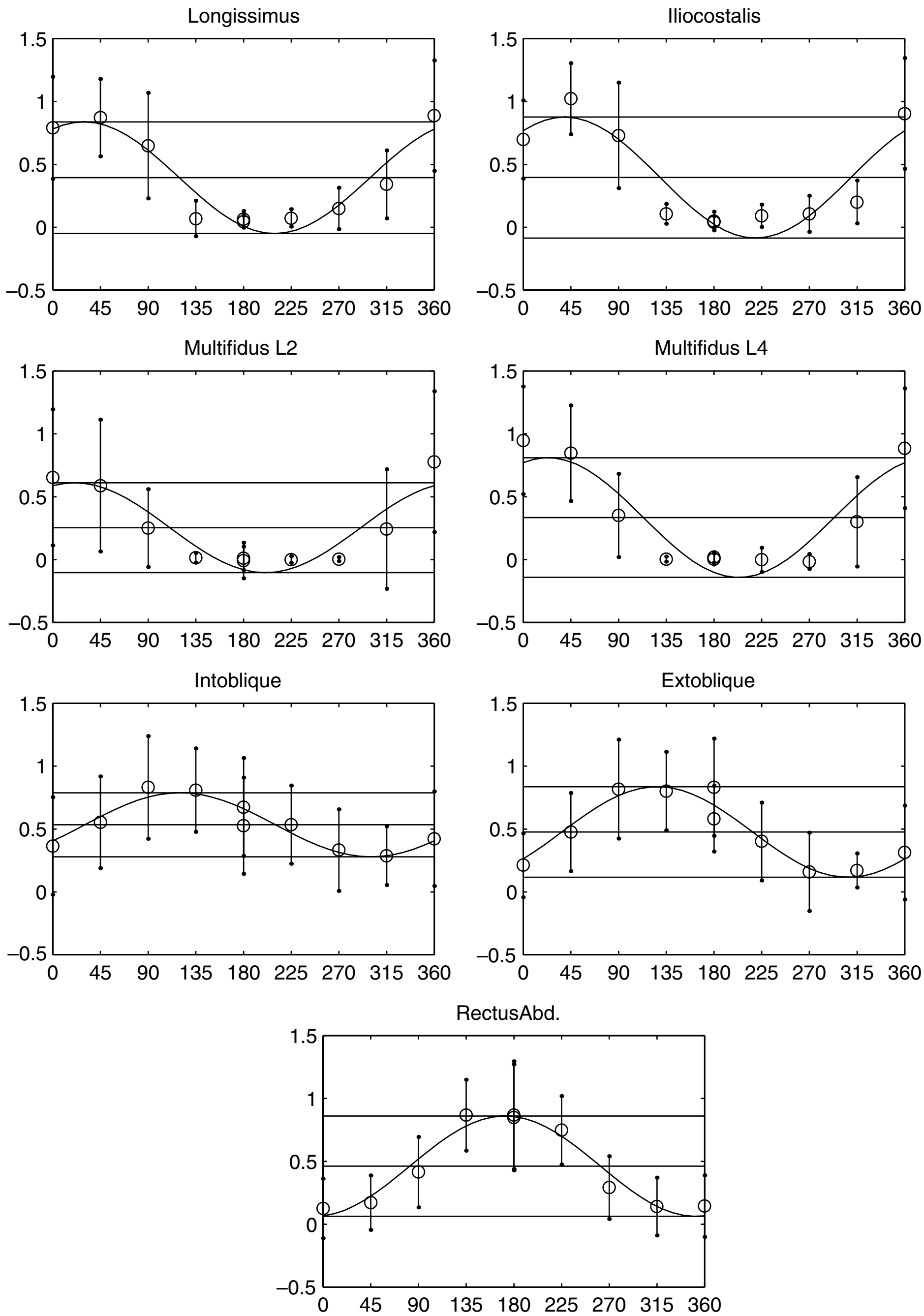

Fig. 4. Mean EMG/effort slope (upslope), plotted against the angle of the external force (effort) with extension = 0° and 360°, and flexion = 180°. The error bars represent the standard deviation. The sine waves were fitted by least-squares to the averaged (23 subjects and three trials) data points with each subplot showing data compiled from both the right and left muscle for each muscle pair. Horizontal lines represent the mean, maximum and minimum of each sine wave. Abdominal muscles were activated at all angles, including when these muscles were considered to be antagonistic (c. 0 and 360°).

into account, the electromechanical delay was evidently over-estimated (Table 2). Electromechanical delay is difficult to measure directly, and in this study the standard deviations in Table 2 are high, probably because estimates were obtained from a single ramped effort trials up to a maximum effort, followed by slow relaxation. Measurements of electromechanical delay have been made with continuously cycling efforts (Thelen et al., 1994; van Dieën et al., 1991). Probably the ideal method to obtain the time and effort-varying EMG–effort relationship is in a test with pseudorandomly varying effort (Kearney and Hunter, 1984)).

These measurements were all made under nominally isometric conditions, without the added complexity of

Table 2
Mean (SD) of temporal factors in the EMG–effort relationships

Muscle	Time delay between max EMG and max. effort (ms)	Time shift that maximized EMG/effort correlation (ms)	Time shift after compensation for slope difference (ms)
Left longissimus	435 (451)	415 (171)	93 (91)
Right longissimus	386 (550)	429 (219)	39 (101)
Left iliocostalis	436 (550)	432 (181)	142 (137)
Right iliocostalis	437 (484)	365 (208)	59 (90)
Left multifidus L2	544 (649)	384 (336)	196 (284)
Right multifidus L2	542 (555)	302 (348)	31 (194)
Left multifidus L4	658 (584)	448 (241)	116 (141)
Right multifidus L4	454 (463)	407 (230)	66 (155)
Left internal oblique	455 (646)	295 (296)	141 (217)
Right internal oblique	385 (618)	245 (305)	141 (257)
Left external oblique	463 (520)	302 (255)	53 (182)
Right external oblique	261 (541)	256 (233)	69 (241)
Left rectus abd.	327 (416)	178 (169)	48 (78)
Right rectus abd.	352 (625)	236 (241)	83 (207)

Averaged over 23 subjects, and 3 trails, for each muscle at the effort angle at which the activation (upslope) was the greatest.

Table 3
Mean correlation coefficients between RMS-EMG and effort, averaged over 23 subjects and three trials, for the testing angle at which each muscle's 'upslope' was a maximum

Muscle	Initial value	After slope adjust only	After time shift only	After slope adjust and time shift
Left longissimus	0.76 (0.30)	0.81 (0.26)	0.86 (0.20)	0.86 (0.15)
Right longissimus	0.71 (0.33)	0.77 (0.28)	0.85 (0.24)	0.83 (0.17)
Left iliocostalis	0.79 (0.27)	0.84 (0.22)	0.80 (0.20)	0.87 (0.15)
Right iliocostalis	0.70 (0.35)	0.76 (0.26)	0.85 (0.26)	0.82 (0.17)
Left multifidus L2	0.45 (0.46)	0.57 (0.38)	0.78 (0.36)	0.71 (0.26)
Right multifidus L2	0.41 (0.47)	0.50 (0.42)	0.59 (0.35)	0.65 (0.29)
Left multifidus L4	0.51 (0.47)	0.62 (0.39)	0.57 (0.35)	0.74 (0.23)
Right multifidus L4	0.42 (0.48)	0.54 (0.40)	0.65 (0.37)	0.70 (0.25)
Left internal oblique	0.79 (0.23)	0.83 (0.20)	0.58 (0.17)	0.86 (0.14)
Right internal oblique	0.74 (0.32)	0.80 (0.23)	0.85 (0.25)	0.84 (0.15)
Left external oblique	0.85 (0.18)	0.88 (0.15)	0.80 (0.16)	0.89 (0.11)
Right external oblique	0.79 (0.24)	0.82 (0.20)	0.88 (0.20)	0.84 (0.15)
Left rectus abd.	0.79 (0.26)	0.82 (0.22)	0.82 (0.22)	0.85 (0.17)
Right rectus abd.	0.76 (0.28)	0.81 (0.22)	0.83 (0.23)	0.84 (0.16)

the length–tension and shortening velocity effects. However, the most plausible explanation for the differing EMG–effort relationships for increasing and decreasing effort conditions was that the series elastic structures, as well as the flexibility of the testing apparatus, resulted in a small lengthening of agonistic muscles as the effort increased and vice versa. Because all the trunk muscles studied here demonstrated antagonistic activation at all testing angles, this effect was present even when a muscle was considered to be an antagonist.

In these experiments the subjects resisted a force that acted horizontally at about the level of T12, thereby generating a shear force and moment that differed at different lumbar levels. The alternative of pure moment generation was not evaluated. Loading that is representative of typical daily activities probably involves a combination of these two idealized cases.

These EMG–effort relationships, especially the greater EMG for effort-increasing tasks than effort-decreasing tasks, should be taken into account in ergonomic and biomechanical studies of muscle activation patterns and inferred spinal loading.

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