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Co-activation alters the linear versus non-linear impression of the EMG–torque relationship of trunk muscles

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Abstract

The use of electromyographic signals in the modeling of muscle forces and joint loads requires an assumption of the relationship between EMG and muscle force. This relationship has been studied for the trunk musculature and been shown to be predominantly non-linear, with more EMG producing less torque output at higher levels of activation. However, agonist–antagonist muscle co-activation is often substantial during trunk exertions, yet has not been adequately accounted for in determining such relationships. The purpose of this study was to revisit the EMG–moment relationship of the trunk recognizing the additional moment requirements necessitated due to antagonist muscle activity. Eight participants generated a series of isometric ramped trunk flexor and extensor moment contractions. EMG was recorded from 14 torso muscles, and the externally resisted moment was calculated. Agonist muscle moments (either flexor or extensor) were estimated from an anatomically detailed biomechanical model of the spine and fit to: the externally calculated moment alone; the externally calculated moment combined with the antagonist muscle moment. When antagonist activity was ignored, the EMG–moment relationship was found to be non-linear, similar to previous work. However, when accounting for the additional muscle torque generated by the antagonist muscle groups, the relationships became, in three of the four conditions, more linear. Therefore, it was concluded that antagonist muscle co-activation must be included when determining the EMG–moment relationship of trunk muscles and that previous impressions of non-linear EMG–force relationships should be revisited.

Keywords: Spine; Abdominal muscle; EMG; Moment; Model; EMG–force; Co-contraction

1. Introduction

Examinations of issues in spine and torso mechanics are often assisted by the use of electromyographic techniques; thus, assumptions must be made regarding the relationship between EMG activation magnitudes and muscular force output. Much of the research concerning EMG–force/torque relationships in the spine literature has focused on that of the extensor musculature. The form of the relationship has been most often identified as non-linear (e.g. Stokes et al., 1987; Thelen et al., 1994; Potvin et al., 1996; Sparto et al., 1998; Staudenmann et al., 2007) although some have determined it to be linear (e.g. Seroussi and Pope, 1987; Dolan and Adams, 1993). Despite the increasing attention paid to the importance of well-coordinated abdominal muscle contraction in ensuring optimal spine health (e.g. van Dieen et al., 2003; Cholewicki et al., 2005; Urquhart et al., 2005; Lee et al., 2006), a very limited amount of work has been done investigating the EMG–torque relationships of the abdominal muscles, yet it too has identified a distinct non-linear form (Stokes et al., 1989, rectus abdominis; Thelen et al., 1994, rectus abdominis and external oblique), with a decline in the rise of the moment as EMG increases.

In determining the nature of the EMG–torque relationship, it appears that there has been a lack of consideration of the additional moment which must be overcome due to antagonist muscle co-activation. Co-activation of muscles acting both agonist and antagonist to a dominant moment is highly prevalent during trunk exertions (Lee et al., 2007; Ross et al., 1993; Thelen et al., 1995; van Dieen et al., 2003). Therefore, it is hypothesized that this activation may

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alter the perceived EMG–torque relationship of trunk muscles, as the torque produced by agonist muscle groups will be continuously underestimated as a function of the comparative amount of antagonist co-activation. The purpose of this paper is thus two-fold: (1) to examine in more detail the EMG–torque relationship of the abdominal musculature and (2) to re-examine the EMG–torque relationship of the extensor musculature with and without accounting for the additional resistive moment that must be overcome due to antagonist muscle co-activation.

2. Methods

Eight healthy males (mean/S.D. age = 24.9/4.7 years, height = 1.79/0.03 m, mass = 82.0/9.1 kg) with no history of back problems, volunteered from the University population. Each read and signed a consent form approved by the University Office of Research Ethics.

2.1. Task

Participants sat with knees supported and pelvis secured in an apparatus designed to foster a neutral spine position (Vera-Garcia et al., 2006). A harness was secured across the chest and attached with a cable to a wall. A force transducer was mounted in-series with the cable.

Participants were instructed to produce controlled isometric (no trunk motion) ramped torque contractions from rest to maximum and back to rest in each of four positions: (1) extensor torque with torso upright (Extensor Upright), (2) extensor torque with torso flexed about the hips to 50% of maximum hip range of motion (Extensor 50), (3) flexor torque with torso upright (Flexor Upright) and (4) flexor torque with torso flexed about the hips to 50% of maximum hip range of motion (Flexor 50). Three trials of each torque contraction were performed in a randomized order.

2.2. Instrumentation and processing

Fourteen channels of EMG were collected from the following muscles bilaterally: rectus abdominis (RA; 2 cm lateral to the midline at the approximate level of the umbilicus), external oblique (EO; approximately 14 cm lateral to the midline oriented infero-medially at 45°), internal oblique (IO; approximately 2 cm medial and inferior to ASIS oriented horizontally), latissimus dorsi (LD; approximately 15 cm lateral to midline at T9 level oriented supero-laterally) and three levels of the erector spinae (T9, L3 and L5; 5, 3, and 1 cm lateral to midline, respectively). Blue Sensor bi-polar Ag–AgCl electrodes (Ambu A/S, Denmark; intra-electrode distance of 2.5 cm) were placed over the muscle belly of each muscle in line with the direction of fibers. Signals were amplified (±2.5 V; AMT-8, Bortec, Calgary, Canada; bandwidth 10–1000 Hz, CMRR = 115 dB at 60 Hz, input impedance = 10 GΩ). Both EMG and force signals were sampled at 2048 Hz.

EMG signals were processed in two ways: in the first method, the raw DC bias was initially removed, followed by low-pass filtering at 500 Hz, rectifying, low-pass filtering at 2.5 Hz (both Butterworth 2nd order), and normalizing to the maximum processed voltage obtained in maximum voluntary isometric contractions. In the second method, each step in the first method was repeated with one additional step: the raw signal was high-pass (HP) filtered at 300 Hz, as suggested by Staudenmann et al. (2007).

An active marker system (Optotrak, Northern Digital Inc., Waterloo, Canada) was used to monitor the position of the upper body throughout each of the contractions. Markers were placed on the following locations on the right side of the body: (1) head (zygomatic process), (2) shoulder (greater tubercle of humerus), (3) elbow (lateral epicondyle), (4) wrist (ulnar styloid), (5) hand (third metacarpal–phalangeal joint). Fins, each with two co-linear markers, were placed at the spinal levels of C7, T12 as well as the sacrum. These fins were used to determine the relative angle of the lumbar spine as well as the projection into the body to determine the approximate locations of the C7/T1 and L4/L5 joint centers. Finally, two markers were placed on the cable attached to the upper body to determine the line of pull of the generated force. Marker data were sampled at 64 Hz.

A two-dimensional top down linked-segment model was used to determine the L4/L5 moment produced by the weight of the upper body (anthropometrics from Winter (2005)). This was summed with the moment determined from the product of the force applied to the cable and its moment arm to the L4/L5 joint to obtain the net external L4/L5 moment.

The normalized EMG signals were entered along with the lumbar flexion angle into an anatomically detailed model of the lumbar spine (McGill and Norman, 1986; Cholewicki and McGill, 1996). A Distribution–Moment approach was utilized to determine individual muscle forces based on normalized activation, instantaneous muscle length, cross-sectional area and an assumed muscle stress of 35 N/cm^2. The net moment produced by each of the extensor and abdominal muscle groups were determined as follows:

$$M_{\text{extensor}} = \sum_{m=1}^{78} r_{m\,\text{extensor}} \times F_{m\,\text{extensor}},$$

$$M_{\text{flexor}} = \sum_{m=1}^{10} r_{m\,\text{flexor}} \times F_{m\,\text{flexor}}, \tag{1}$$

where M_{extensor} and M_{flexor} are the moments produced by the extensor musculature (78 muscle fascicles representing the lumbar and thoracic longissimus and iliocostalis, multifidus, latissimus dorsi and quadratus lumborum muscle groups) and flexor musculature (10 muscle fascicles representing the rectus abdominis, external oblique and internal oblique muscle groups), about the L4/L5 joint, respectively; $r_{m\,\text{extensor}}$ and $r_{m\,\text{flexor}}$ are the extensor and flexor muscle moment arms, about the L4/L5 joint, respectively, and $F_{m\,\text{extensor}}$ and $F_{m\,\text{flexor}}$ are the individual muscle fascicle forces in each of the extensor and flexor muscle groups.

The total resistive moment required to be overcome by the agonist muscle group was determined as either (a) the externally calculated moment alone or (b) the summation of the externally calculated moment and the antagonist muscle moment:

(a) $M_{\text{resistive}} = M_{\text{external}},$

(b) $M_{\text{resistive}} = M_{\text{external}} + M_{\text{antagonist}},$ (2)

where $M_{\text{resisitive}}$ is the moment that must be produced by the agonist muscle group, M_{external} is the moment measured externally and $M_{\text{antagonist}}$ is the moment produced by the antagonist muscle group (flexor muscles in the extensor moment trials and extensor muscles in the flexor moment trials).

All moment data were visually windowed over the period from the start of external moment generation until the end of external moment generation. For further analysis, subsequent windows were made of the force increasing and force decreasing portions of the contraction, which will be referred to as concentric and eccentric portions of the contraction

Table 1
Best-fit non-linear coefficients (determined for Eq. (3)) and root mean square difference (% MVC) for both the linear and best non-linear fits, between EMG moments and the externally determined moments alone (in the absence of antagonist muscle moments)

	Full ramp			
	Extensor Upright	Extensor 50%	Flexor Upright	Flexor 50%
Co-efficient	1	6	9	14
RMS	10.32	13.09	13.06	13.50
RMS linear	10.39	14.16	14.88	16.97

(assuming compliant tendinous attachments allowing the musculature to shorten and lengthen in the absence of gross spine movement).

In each of these cases, data from all trials of each condition were pooled, and the linearity between the agonist muscle moment and the resistive moment was tested with the following equation (Potvin et al., 1996):

$$M_{\text{agonist}\,N} = \frac{e^{(-M_{\text{agonist}\,L} * \delta * 0.001)} - 1}{e^{(-0.1 * \delta)} - 1}, \tag{3}$$

where $M_{\text{agonist}\,N}$ is the agonist muscle moment non-linearly normalized to 100% maximum, $M_{\text{agonist}\,L}$ is the agonist muscle moment linearly normalized to 100% maximum and δ is a constant to define exponential curvature (ranging from -50 to 50).

The root mean square difference was calculated between each of the linearly and non-linearly normalized muscle moments ($\delta = -50$ to 50; total 101) and the resistive moment (both with and without accounting for the antagonist moment). For each of the four conditions (Extensor Upright, Extensor 50, Flexor Upright, Flexor 50), the minimum RMS difference indicated the curvature resulting in the best fit between the muscle and resistive moments:

$$\text{RMS}_{\text{difference}} = \sqrt{\frac{1}{T}\sum_{t=1}^{T}(M_{\text{agonist}\,N_t} - M_{\text{resisitve}\,t})^2}, \tag{4}$$

where T is the total number of time instances analyzed across all trials and participants per condition.

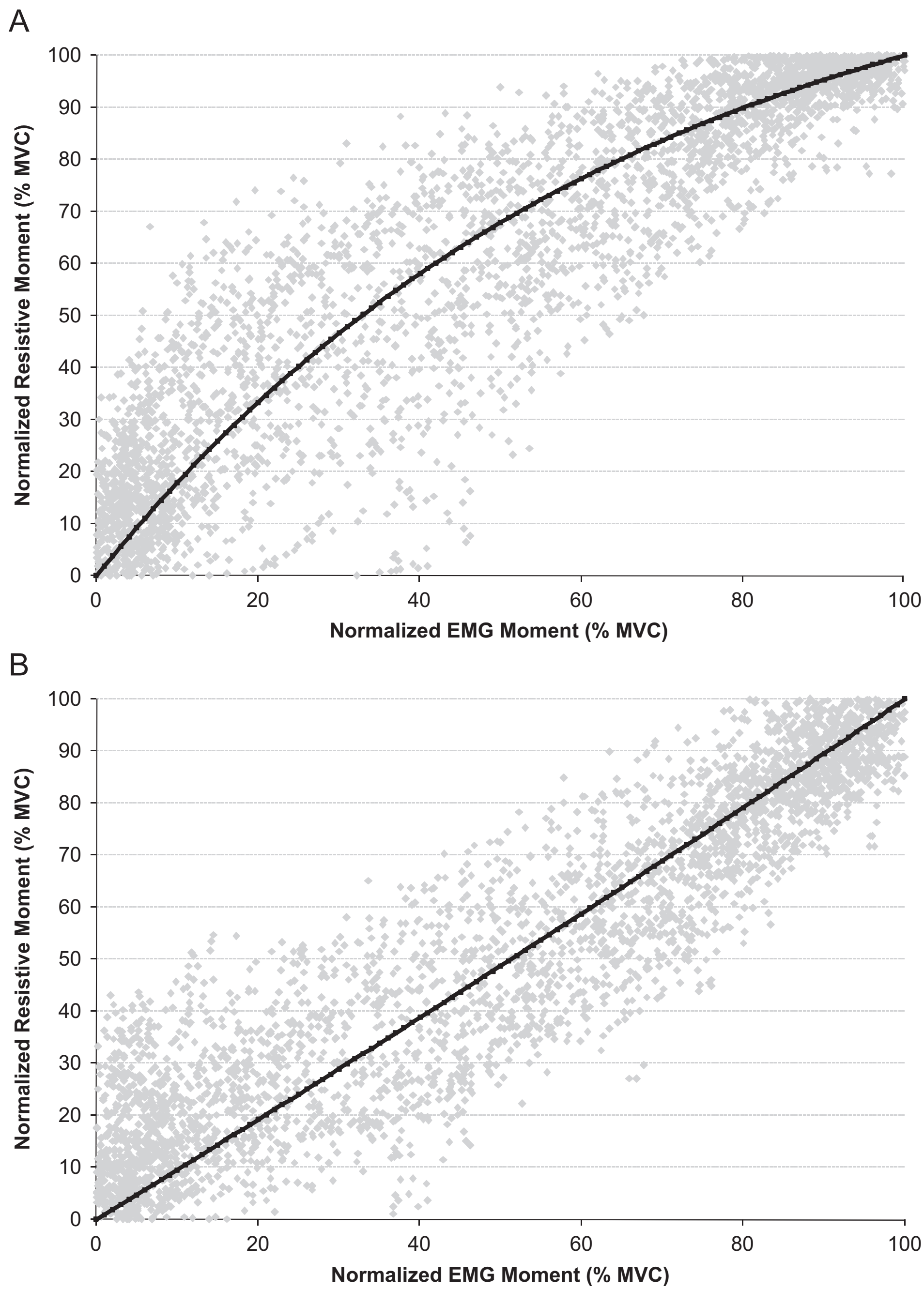

Fig. 1. Scatterplots (all participants and trials) for the Flexor Moment 50% flexed condition displaying the Agonist EMG Moment normalized to 100% of maximum versus the Resistive Moment normalized to 100% of maximum. (A) Resistive Moment is the externally calculated moment alone and (B) Resistive Moment is the combined externally applied moment and antagonist muscle moment.

Finally, an index of trunk muscle co-activation was calculated as the percent ratio of antagonist moment to agonist moment at each instant throughout the contraction.

3. Results

3.1. Effect of antagonist muscle activity

When determining the linearity in the EMG–torque relationship without consideration of antagonist muscle activity, relationships ranged from nearly linear (Extensor Upright) to varying degrees of the non-linear form reported previously in literature, with a declining increase in moment as EMG increased across its spectrum from zero to 100% of maximum (Extensor 50, Flexor Upright, Flexor 50) (Table 1).

Accounting for the additional resistive moment generated by the antagonist muscle groups altered the EMG–torque relationship in all cases, making it more linear in each of the Extensor 50, Flexor Upright and Flexor 50 conditions (Fig. 1, Table 2). The relationship became slightly more non-linear in the Extensor Upright condition; however, the non-linearity was opposite to that found previously in the experimental literature, with a rise in the increasing moment as EMG increased across its spectrum (Fig. 2, Table 2). This same slight non-linear form was also detected in each of the Flexor Upright and Flexor 50 conditions.

Further analysis determined that the majority of the change in linearity occurred in the concentric (force increasing) portion of the contraction (Table 2); the eccentric portion (force decreasing) of the contraction still displayed a slightly rising increase in moment per unit increase in EMG in all conditions except Extensor Upright.

Table 2
Best-fit non-linear coefficients (determined for Eq. (3)) and root mean square difference (% MVC) for both the linear and best non-linear fits, between EMG moments and resistive moments (calculated as the sum of the antagonist muscle moment and externally determined moment) (also shown are the best-fit coefficients and RMS differences when the raw EMG was high-passed filtered at 300 Hz)

	Extensor upright	Extensor 50%	Flexor upright	Flexor 50%
Full ramp				
Coefficient	−3	2	−3	−1
RMS	9.15	10.92	12.90	12.34
RMS linear	9.49	11.05	13.06	12.38
Coefficient HP	−3	2	−2	−1
RMS HP	9.32	10.92	12.90	12.34
RMS linear HP	9.73	11.02	12.95	12.46
Concentric				
Coefficient	−2	2	−3	−2
RMS	8.52	11.38	13.83	10.61
RMS linear	8.78	11.48	14.13	10.72
Coefficient HP	−3	2	−3	−2
RMS HP	8.36	11.24	13.98	10.66
RMS linear HP	8.67	11.32	14.15	10.78
Eccentric				
Coefficient	−2	3	4	2
RMS	7.52	7.40	9.55	8.64
RMS linear	7.65	7.68	9.98	8.75
Coefficient HP	−2	3	4	2
RMS HP	7.90	7.46	9.57	9.04
RMS linear HP	8.07	7.78	10.01	9.17

3.2. Amount of antagonist muscle activity present

A relatively high level of antagonist muscle activity was present in all the conditions examined in this study (Fig. 3). The greatest amount of antagonist activity occurred in the Flexor 50 condition (ranging from 50% to 298%), and the least in the Extensor 50 condition (ranging from 19% to 27%).

3.3. Effect of high-pass filtering

High-pass filtering of the raw EMG signal had very little effect on the EMG–torque relationship in all cases (Table 2).

4. Discussion

The primary result of this study was that accounting for antagonist muscle activity influences the relationship between trunk EMG and its generated torque. Specifically, antagonist muscle activity creates an additional resistive moment that has to be overcome by the agonist muscle groups; ignoring this gives the impression of a non-linear relationship between the agonist EMG and the externally generated moment. The true nature of the trunk EMG–torque relationship was found to be more linear than has often been previously reported (Fig. 1), and in fact may display a slight opposite non-linearity (Table 2, Fig. 2) to that normally cited in experimental literature, with an increase in the rise in moment as EMG increases through its range of activation. This opposite non-linearity has been predicted theoretically using motor unit based models of EMG (Milner-Brown and Stein, 1975; Fuglevand et al., 1993).

The amount of co-activation that occurred in the isometric flexor and extensor moment tasks studied here was quite high (Fig. 3). Generating the flexor moments, in particular, produced a substantial amount of activation from the trunk extensor musculature. This is not at all surprising in the Flexor 50 condition, where activation of the extensor musculature was required simply to balance the flexor moment created by the mass of the upper body. In the other three conditions, however, the co-activation served very little or no direct purpose in balancing externally produced moments, and therefore acted primarily to provide a level of stiffness and stability sufficient to prevent the spine from buckling under load. The average

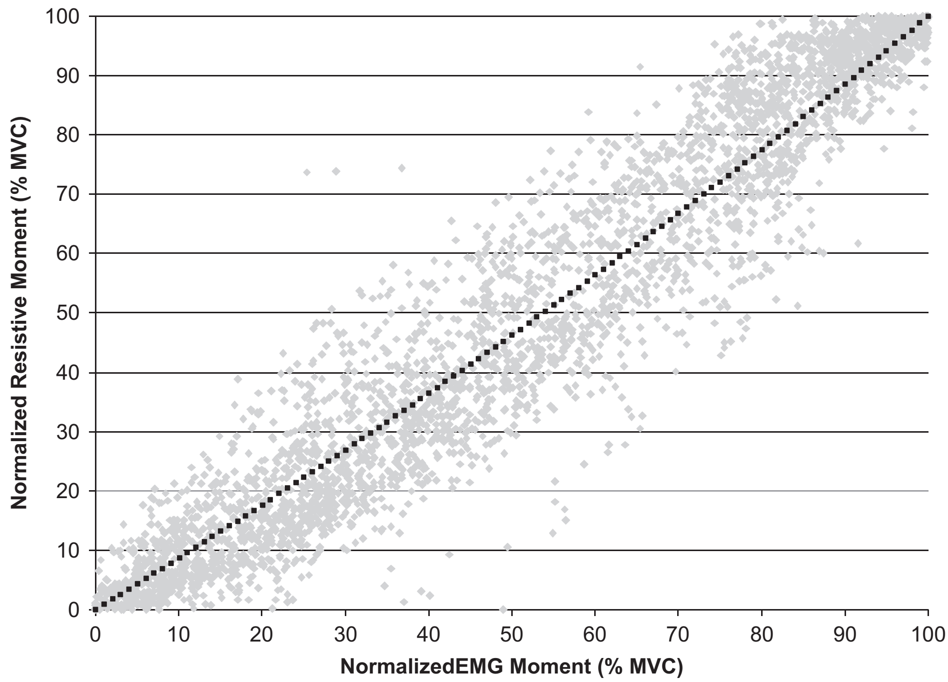

Fig. 2. Scatterplots (all participants and trials) for the Extensor Upright condition displaying the Agonist EMG Moment normalized to 100% of maximum versus the Resistive Moment (accounting for the antagonist muscle moment) normalized to 100% of maximum. Note that the slight non-linearity in the curve fit is opposite to that normally cited in the experimental literature.

level of co-activation, as calculated in this study, never dropped below 18%. This supports previous findings and hypotheses stating that some level of antagonist co-activation is constantly required to maintain the integrity of the spinal column (Cholewicki and McGill, 1996; Brown and Potvin, 2005). Therefore, it is not surprising that consideration of this consequent additional moment is necessary to properly model the torque generated by the agonist muscle group of interest.

Contrary to previous work (Potvin and Brown, 2004; Staudenmann et al., 2007), high-pass filtering the raw EMG signal did not improve the estimation of the generated muscle torque in this study (Table 2). This portion of the study was not intended as a robust analysis of the effect of the high-pass filtering of EMG data; rather it was designed to test whether the single best cut-off value as determined by Staudenmann et al. (2007) would apply to the current data. It should be noted that filtering in this manner did alter the magnitude (i.e. gain) of the relationship (not reported here), as in Staudenmann et al. (2007), but did not alter the form of the relationship, which was the focus of this study. It is also interesting to note that in Staudenmann et al. (2007), high-pass filtering had the effect of increasing the net antagonist moment to a greater extent than the agonist moment, which corresponded to an improved linearity in the EMG–torque relationship.

The participants in the current study were limited to eight healthy males. The goodness-of-fit of the experimental data, combined with the intended purpose of the study to demonstrate the necessity of considering antagonist muscle activation in determining EMG-based estimates of spinal force/torque, indicates that this number of participants has been sufficient to accomplish this goal. Consideration of antagonist activity has been clearly shown to be essential for at least the eight participants studied here; this makes sense both biologically and mechanically, and alone should indicate that this is a consideration that should not be overlooked in these types of analyses.

Finally, the additional purpose of this paper was to test the EMG–torque relationship of the abdominal musculature. Negligible differences were found between trunk extensor and abdominal flexor muscles in terms of the form of the EMG–torque relationship. Thus, the surface EMG signals obtained from these muscles can be treated similarly in the data processing stage; however, the scaling magnitude between the EMG estimated torque and actual torque will be highly dependent upon model assumptions, anatomical fidelity and measurement accuracy, and must be additionally considered in order to model the net muscle force and torque outputs, and corresponding joint forces and measures of stability and/or stiffness. The current findings will improve the modeling and estimation of these joint parameters, which is essential to further the understanding of the muscular relationship to spine injury, rehabilitation and performance.

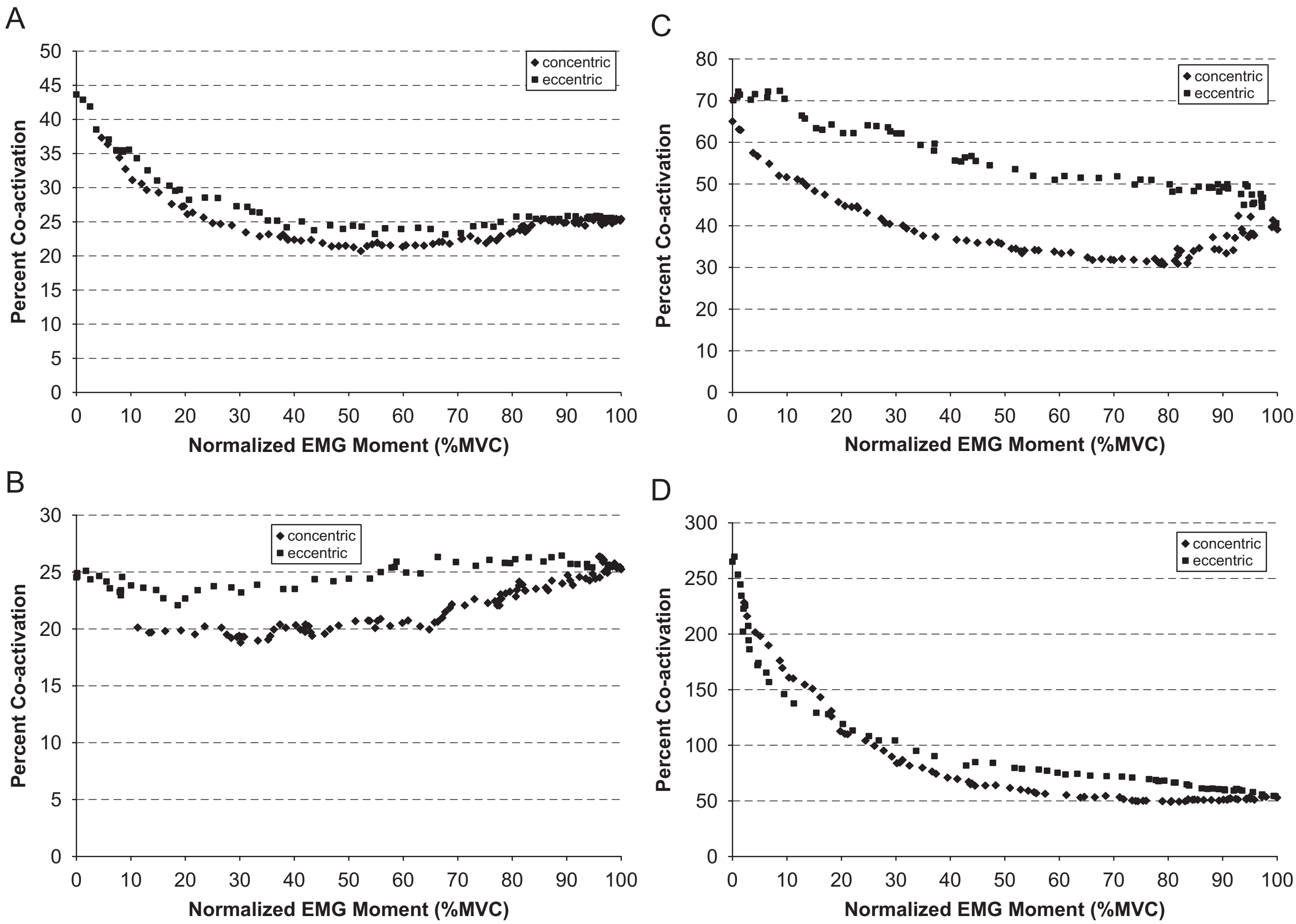

Fig. 3. Average co-activation index (percent ratio of antagonist-to-agonist muscle moment) normalized with respect to the dominant EMG moment across all participants for each condition: (A) Extensor Upright, (B) Extensor 50%, (C) Flexor Upright and (D) Flexor 50%. Relationship is shown for each of the concentric and eccentric portions of the ramped contraction.

Conflict of interest statement

Neither author has any affiliations that have influenced the content of this work.

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