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## Development of a lumbar EMG-based coactivation index for the assessment of complex dynamic tasks

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### ABSTRACT

The objective of this study was to develop and test an EMG-based coactivation index and compare it to a coactivation index defined by a biologically assisted lumbar spine model to differentiate between tasks. The purpose was to provide a universal approach to assess coactivation of a multi-muscle system when a computational model is not accessible. The EMG-based index developed utilised anthropometric-defined muscle characteristics driven by torso kinematics and EMG. Muscles were classified as agonists/antagonists based upon 'simulated' moments of the muscles relative to the total 'simulated' moment. Different tasks were used to test the range of the index including lifting, pushing and Valsalva. Results showed that the EMG-based index was comparable to the index defined by a biologically assisted model ( $r^2 = 0.78$ ). Overall, the EMG-based index provides a universal, usable method to assess the neuromuscular effort associated with coactivation for complex dynamic tasks when the benefit of a biomechanical model is not available.

**Practitioner Summary:** A universal coactivation index for the lumbar spine was developed to assess complex dynamic tasks. This method was validated relative to a model-based index for use when a high-end computational model is not available. Its simplicity allows for fewer inputs and usability for assessment of task ergonomics and rehabilitation.

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### KEYWORDS

Co-contraction;  
neuromuscular; trunk  
muscles; co-activation

## 1. Introduction

Complex, dynamic tasks ranging from manual materials handling to sedentary work to activities of daily living impose varying external moments that require internal moments to counterbalance the load. This involves the active coordination between systems of agonist and antagonist activity from the musculature surrounding the trunk. The coactivation between muscle systems serve to increase torso stiffness, stabilise the external load, and adjust to potential perturbations (Thelen, Schultz, and Ashton-Miller 1995). However, a higher neuromuscular effort may be endured due to high levels of coactivity. These efforts exist across all tasks; though when the amount of coactivation exceeds the typical amount necessary to accomplish a task, a higher effort is endured by the neuromuscular system. For example, tasks requiring higher degrees of postural control from higher muscular activations to guard from pain (LBP patients) (Marras et al. 2004) or higher task frequency for inexperienced workers (Marras et al. 2006) would incur a higher neuromuscular effort. Given the complexity in assessing the

neuromuscular effort of an exertion from a multi-muscle system, particularly in a complex dynamic environment, a coactivation index was needed to understand the effort from a systems perspective.

Surface electromyography (EMG) is typically used to measure muscle activity, which is then used to assess coactivity and the index (neuromuscular effort) associated with the coactivity. Current approaches in the literature are highly variable between one another and range from simplistic, a priori defined antagonist/agonist activations to advanced techniques which require the use of computational models to define muscle forces and moments (Le, Best et al. 2017). Given the variability between methods as well as the number of methods in existence, a standard, universally usable index for the comparison of tasks across studies is still lacking in the literature. The challenge of using a priori defined systems is the limited utility to a small number of predefined tasks involving either uniplanar dynamic or multiplanar isometric exertions. Changes in posture or location of the external load may require shifts in agonist or antagonist activity to accommodate the task (Lavender et al. 1992). Depending on the

location of the external load and torso posture, the trunk musculature may change its individual moment contributions. To account for those changes, a method by (Le, Aurand et al. 2017) utilised an extensively validated, biologically assisted dynamic spine model to extract moment data based on the active force components. The moment data from each muscle were then classified as agonists and antagonists using a dot-product method (Andrews and Hay 1983) relative to the total internal moment. A similar technique was utilised by Song, Bok, and Chung (2004) to determine a coactivation index, but was limited to a priori definitions of antagonist activity and isometric testing. The novelty of the method from Le, Aurand et al (2017) is its ability to continuously classify antagonist and agonist activity from calculated moments and provide an understanding of neuromuscular effort of multi-planar, dynamic exertions/tasks. This effort was defined in the form of an index as the proportion of the overall system loading due to coactivity (synergistic activation of agonist and antagonist systems). This method is capable of differentiating levels of coactivation between various complex dynamic tasks involving lifting and pushing and showed that tasks requiring higher levels of postural control would result in a higher coactivation index (ie pushing with obstacles or precision placement during lowering). Although it provides a higher-resolution understanding of coactivity relative to a priori methods, it requires high-end modelling efforts to calculate. To allow for 'universal' application, a method is sought to calculate a coactivation index comparable to the moment-based (operationally defined as 'standard' in this manuscript) approach. Since the basis of the methodology relies on moments derived from the model, it was speculated that a surrogate approach may be developed for cases in which a model is not available.

Dynamic moment data are dependent on the derivation of the forces and the change in moment arms with respect to the vector lines of action. Force calculations necessitate modelling efforts to account for the effects of cross-sectional area (CSA), force-length and force-velocity contributions to the muscle. Changes in the effective moment arms require an understanding of the lines of action and how they move relative to the geometry of the intervertebral bodies and dynamics of the torso. However, in the absence of a model, many of these factors may not be accessible which presents computational complexity in defining coactivation through the standard approach. Since the coactivation index is based upon a system of muscle forces acting relative to one another with dimensionless units, it was postulated that a surrogate approach with proportionate trends may be able to replicate similar index magnitudes to the standard index. This requires the understanding of how each of the individual

components influence the force equation before developing a surrogate.

The objectives of this study were twofold: (1) reduce the components of the force equation to understand how it may affect the coactivation index and (2) develop an EMG-based coactivation index and compare it to the biomechanical model-dependent index developed from Le, Aurand et al (2017). This method would serve to assess coactivation for complex dynamic tasks in multiple-muscle system without a computationally intensive model. Its utility resides in ergonomic task assessment as well as testing of rehabilitation effectiveness.

## 2. Coactivation index structure and component reduction

### 2.1. Biologically assisted model data extraction

The underlying logic of the coactivation index utilises the active force components from the 10 power-producing muscles of the trunk relative to their moment arms ( $r_i$ ) and the total active moment at L5/S1 (Equations (1)–(3)). These muscles include bilaterally the: latissimus dorsi, erector spinae, rectus abdominis, external obliques and internal obliques (represented as  $i = 1:10$ ). Individual active muscle force components were determined by a biologically assisted model as the product of the Gain Ratio (GR, N/cm<sup>2</sup> V), EMG activity (V), cross-sectional area (CSA, cm<sup>2</sup>), force-length ( $f[L]$ , dimensionless) and force-velocity ( $f[V]$ , dimensionless) modulation factors (Hwang et al. 2016) (Equation (1)). The Gain Ratio (GR) is defined as the Gain (maximum muscle force per unit area) divided by the maximum voluntary contraction (MVC) and provides a method to modulate the force component without the need for measuring MVCs (Dufour, Marras, and Knapik 2013). Maximum CSAs allowed for the estimate of the maximum force generation capabilities for each of the 10 muscles. The force-length modulation factor ( $f[L]$ ) accounts for changes in force production from changes in muscle length of the individual muscles relative to the normalised length of the muscle (Cadova, Vilimek, and Daniel 2014). The force-velocity factor ( $f[V]$ ) modulates the force output based upon changes in muscle velocity during concentric and eccentric activity (Close 1964; Hill 1938). Collectively, these components represent the dynamic properties of the active force component of the musculature (Equation (1)).

Active moments were calculated as the cross-product of the L5/S1 moment arm ( $r_i$ ) and the active muscle force component ( $F_i$ ) (Equation (2)). Summation of the active moments provided the total active moment ( $M_i$ ) of the system (Equation (3)). The dot product of the individual muscle's moment ( $m_i$ ) relative to the total active moment ( $M_i$ ) normalised to the resultant total internal moment

results in a scalar projection ( $\text{Proj}_i$ ) defining a muscle's contribution as either an agonist (Nm, positive) or antagonist (Nm, negative) (Equations (4)–(6)). The overall coactivation index was then defined as the product between the ratio (opposition) of the antagonist/agonist systems and the normalised magnitude of the contribution (Equation (7)). The maximum summation of agonist and antagonist systems for the normalised contribution from the entire data-set collected (all subjects and trials). Overall, the idea is that the index would accommodate occupational-related activities with most indices occurring between 0 (lowest – no coactivation) and 1 (high coactivity). However, it is possible for the index to exceed 1 under extreme loading conditions. From the empirical data-set used in this study (further explained in Section 3), the maximum system activation (peak summation of agonist and antagonist activity) at L5/S1 was calculated as 545 Nm. Further details of the logic behind the equation can be found in Le, Aurand et al. (2017).

$$F_i(t) = F_{\text{Active}_i}(t) = \text{EMG}_i(t) \cdot (\text{GR}_i) \cdot \text{CSA}_i \cdot f[L_i(t)] \cdot f[V_i(t)] \quad (1)$$

$$\vec{m}'_i = \vec{r}_i \times \vec{F}'_i \quad (2)$$

$$\vec{M}'_i(t) = \sum_{i=1}^t \vec{m}'_i \quad (3)$$

$$\text{Proj}_i = \frac{\vec{m}'_i \cdot \vec{M}'_i}{||\vec{M}'_i||} \quad (4)$$

$$\text{antagonist}_i = \begin{cases} 0, & \text{Proj}_i > 0 \\ |\text{Proj}_i|, & \text{Proj}_i \leq 0 \end{cases} \quad (5)$$

$$\text{agonist}_i = \begin{cases} \text{Proj}_i, & \text{Proj}_i > 0 \\ 0, & \text{Proj}_i \leq 0 \end{cases} \quad (6)$$

$$CI(t) = \left( \frac{\sum_{i=1}^{10} \text{antagonist}_i(t)}{\sum_{i=1}^{10} \text{agonist}_i(t)} \right) \times \left( \frac{\sum_{i=1}^{10} \text{antagonist}_i(t) + \sum_{i=1}^{10} \text{agonist}_i(t)}{\max(\text{antagonist}_i + \text{agonist}_i)} \right) \quad (7)$$

## 2.2. Component reduction

To achieve the first objective, each component of the active force equation (Equation (1)) was reduced stepwise one by one to understand how it affects the coactivation index

(Equation (7)) by comparing the results to the index from the full equation (standard index) using the coefficient of determination ( $r^2$ ). The reduced component and model-independent calculated force (Section 2.3) is operationally defined as a 'simulated' force (sForce) since the force value changes its magnitude and units after the CSA is extracted. The 'simulated' force units for the reduced equations of  $\text{EMG} \cdot \text{GR}$ ,  $\text{EMG} \cdot \text{GR} \cdot \text{CSA}$  and  $\text{EMG} \cdot \text{GR} \cdot \text{CSA} \cdot f[L]$  are  $\text{N}/\text{cm}^2$ , N and N, respectively. After each reduction in the force equation, the agonist and antagonist systems are then reassessed for the maximum system activation constant ( $\max(\text{agonist} + \text{antagonist})$ ) used to normalise the contribution (Equation (7)). This allows for the index magnitudes to be dimensionless and comparable to the standard index.

## 2.3. Model independent approach

The second objective was achieved by replacing the force component (Equation (1)) with a sForce component driven by normalised EMG (nEMG), anthropometric regression defined muscle lines of action, and scaled by normalised CSA (dimensionless). The purpose of this methodology was to provide a surrogate method to calculate a coactivation index without accessibility to a biologically assisted model.

The nEMG approaches were independent of the model and required inputs of EMG and anthropometric-defined moment arms, CSAs and lines of action for each of the 10 power-producing muscles of the trunk (Jorgensen, Marras, and Gupta 2003; Jorgensen et al. 2001; Marras et al. 2001). Torso kinematics were extracted from motion capture data represented in the form of quaternions to drive the dynamic lines of action. Each of the muscle lines of action were rotated using the quaternion rotation matrix ( $q_{\text{Torso}}$ ) through the rotation/translation of the insertion points (Equation (8)) relative to L5/S1 which was operationally defined as 0.1 m superior to the centroid of the motion capture-defined pelvis. Lines of action were converted into unit vectors (Equation (9)) with a magnitude defined by nEMG which was normalised by the peak of a series of reference exertions ( $\max \text{EMG}_i$ ) (Equation (10)). The nEMG was further scaled as the product with the normalisation of the CSAs (nCSA) relative to the muscle with the largest CSA in the system of muscles analysed, with output defined as sForce (Equation (11)). The sForce data were then used as inputs into Equations (2)–(7).

$$\vec{V}'_i(t) = q_{\text{Torso}}(t)^{-1} \times V_i(t) \quad (8)$$

$$\hat{V}'_i(t) = \vec{V}'_i(t) / ||\vec{V}'_i(t)|| \quad (9)$$

$$nEMG_i(t) = \frac{EMG_i(t)}{\max(EMG_i)} \quad (10)$$

$$\overline{sForce}_i(t) = \hat{V}_i'(t) \times nEMG_i(t) \times \left( \frac{CSA_i}{\max(CSA_{j=1:10})} \right)^j \quad (11)$$

### 3. Methods

#### 3.1. Experimental approach

A study was conducted to test the reduced component and nEMG-based coactivation indices and compare them to the model-driven, standard coactivation index during complex dynamic tasks.

#### 3.2. Subjects

Seventeen subjects (7 males and 10 females) were recruited for this study (age  $26.7 \pm 5.8$  years, mass  $73.6 \pm 17.1$  kg and height  $172.4 \pm 7.1$  cm). All subjects reported no LBP in the past 6 months. Subjects provided informed consent prior to participating and the study was approved by the University Institutional Review Board.

#### 3.3. Experimental design

Three different tasks were tested to assess the nEMG-based coactivation indices relative to the standard indices. Tasks were grouped and counterbalanced as lifting/lowering, pushing and Valsalva manoeuvres. Within the grouped tasks, a separate set of independent variables were tested, except for the Valsalva, which was executed while standing upright and repeated four times. Lifting/lowering tasks involved different levels of handle height (chest and mid-thigh), asymmetry (clockwise (CW) 45, CW90 and sagittal-symmetric), weight (4.5 and 11.3 kg) and precision placement (constrained and none). Pushing tasks involved different combinations of speed (slow, preferred and fast), type of push (straight and turn), weight (54.4 and 145.2 kg) and precision placement (constrained and none). Further details of the experimental design and procedures can be found in Le et al. (2017).

#### 3.4. Apparatus

Electromyographic (EMG) data were collected with a 16-channel MA400-28 EMG system (Motion Lab Systems, Inc., Baton Rouge, LA, USA) and sampled at a rate of 1000 Hz. Signals were high-pass filtered at 30 Hz, low-pass filtered at 450 Hz, and notch filtered at 60 Hz as well as its aliases. Signals were rectified and smoothed using

a zero-phase moving average filter. Kinematic data were collected using the 24 infrared camera OptiTrack Flex 3 motion capture system (NaturalPoint, Corvallis, OR, USA). Kinetic data were collected from a Bertec 4060A force plate (Bertec, Worthington, OH, USA).

#### 3.5. Procedure

Subjects were briefed on the experiment and provided informed consent prior to participation. Afterwards, subject anthropometry was collected and surface EMG electrodes were placed bilaterally on the latissimus dorsi, erector spinae, rectus abdominis, external obliques and internal obliques (Mirka and Marras 1993). Reflective markers were placed on 41 landmarks on the body to collect kinematic data. Subjects were then asked to perform a series of calibration exertions using a 9.07 kg medicine ball for the purposes of EMG normalisation (Dufour, Marras, and Knapik 2013). These motions involved picking up the medicine ball (with handles) from approximately waist height to chest level near the body. While keeping the ball next to the chest, the subjects were instructed to do a side-to-side lateral bend, followed by sagittal flexion to the upright position. Once the subjects completed the calibration lifts, they were instructed on the tasks assigned and allowed to practice the tasks to reduce learning effects.

#### 3.6. Data analysis

The independent variables were extracted (Table 1) at the peak coactivation index for each component reduction (model-dependent) and each model-independent approach for all combinations of tasks. The standard approach refers to the biomechanical model-dependent approach encapsulating all components of the force equation. Model-independent (nEMG) methods were normalised relative to a series of multi-planar reference exertions using a 9.07 kg medicine ball and Valsalva manoeuvres. Peak EMG values of the reference exertions were extracted for each muscle for normalisation. After the calculation of the various indices, the peak of the nEMG-based index was compared relative to the peak of the standard index through the coefficient of determination ( $r^2$ ).

### 4. Results

The nEMG-based index had a comparable resolution to the standard index (Tables 1 and 2). These results describe the relative fit of the different reduced-component and nEMG approaches relative to the standard index. As expected, as components were reduced from the model-dependent approach, the  $r^2$  also declined. Out of the nEMG, model-independent approaches, the best 'fit' was



**Table 1.** Coefficients of determination ( $r^2$ ) of different reduced-component and nEMG coactivation methods relative to the standard index.

	*	*	*				x	
	F-L							
	CSA	CSA						
	GR	GR	GR		nCSA	nCSA <sup>2</sup>	nCSA <sup>3</sup>	nCSA <sup>4</sup>
	EMG	EMG	EMG	nEMG	nEMG	nEMG	nEMG	nEMG
Overall	0.970 (0.035)	0.954 (0.089)	0.826 (0.174)	0.612 (0.251)	0.630 (0.256)	0.745 (0.213)	0.780 (0.197)	0.780 (0.203)
Lift/Lower	0.963 (0.043)	0.931 (0.120)	0.804 (0.201)	0.596 (0.267)	0.617 (0.273)	0.738 (0.233)	0.756 (0.223)	0.768 (0.214)
Pushing	0.974 (0.024)	0.973 (0.025)	0.836 (0.143)	0.599 (0.222)	0.615 (0.228)	0.738 (0.192)	0.790 (0.168)	0.777 (0.194)
Valsalva	0.996 (0.004)	0.995 (0.007)	0.960 (0.034)	0.935 (0.105)	0.934 (0.111)	0.909 (0.099)	0.944 (0.051)	0.944 (0.056)

Notes: Data presented as mean ( $\pm$  standard deviation).

\*Denotes dependence on biologically assisted model. The sForce approach chosen for further analysis is represented by 'x'. Results presented as mean ( $\pm$  SD).

**Table 2.** Coefficients of determination ( $r^2$ ) for the nEMG\*nCSA<sup>3</sup> coactivation method relative to the standard index.

Lifting/Lowering (Asymmetry*Height)	$r^2$	Pushing (Speed*Type)	$r^2$
Sagittal/Chest	0.874 (0.150)	Slow/Straight	0.700 (0.192)
CW45/Chest	0.791 (0.183)	Preferred/Straight	0.743 (0.190)
CW90/Chest	0.605 (0.279)	Fast/Straight	0.891 (0.103)
Sagittal/Mid-Thigh	0.742 (0.193)	Slow/Turn	0.738 (0.166)
CW45/Mid-Thigh	0.745 (0.226)	Preferred/Turn	0.803 (0.136)
CW90/Mid-Thigh	0.782 (0.195)	Fast/Turn	0.870 (0.091)

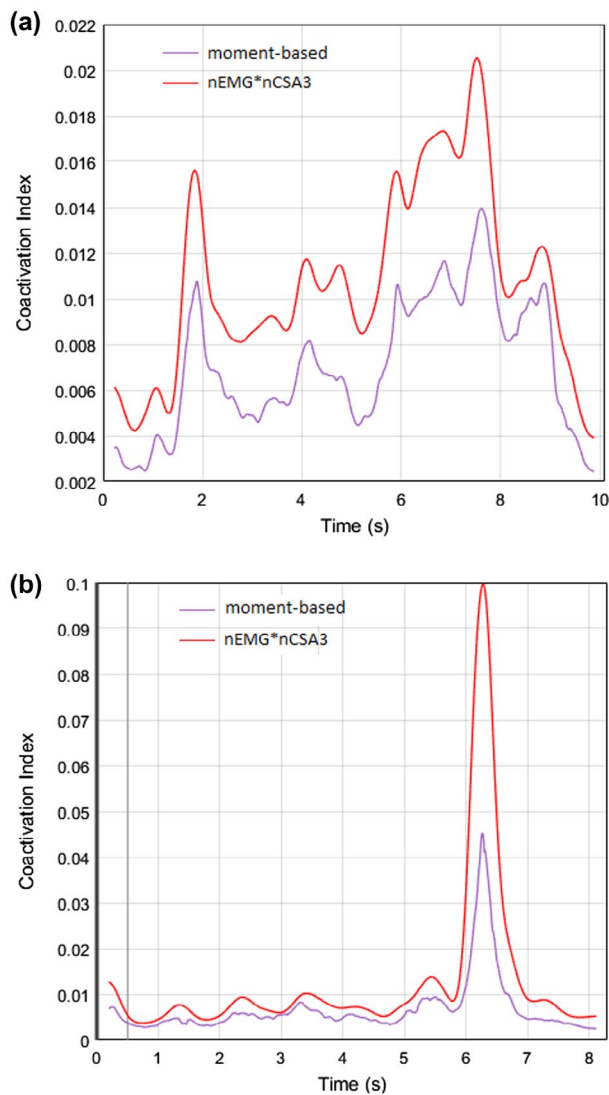
Notes: Weights and precision placement variables were pooled for the tasks. Results presented as mean ( $\pm$  SD).

found to be nEMG\*nCSA<sup>3</sup>. Sample continuous plots comparing the different approaches (reduced component and nEMG) relative to the standard index can be found in Figure 1(a), (b) for lifting and pushing, respectively. The comparison between all the methods across the global tasks also showed that the nEMG\*nCSA<sup>3</sup> method provided the closest match independent of the model (Figure 2). To understand the limitations of coactivation predictions during complex dynamic tasks using the nEMG approach, Table 2 describes the comparability of the fits for lifting and pushing tasks. During lifting, comparability decreased with increasing asymmetry at chest height. During pushing, comparability increased with speed.

The maximum system coactivation (max(antagonist + agonist)) (Equation (7)) of the entire data-set was operationally defined as 545Nm for the standard and 0.2 m for the nEMG approach. The nEMG index coactivation system normalisation factor is 0.2 m because the muscle EMG data are dimensionless multiplied by their respective moment arms. In regards to the biologically assisted model, these constants were associated with high spinal loads of 7940 N of compression and 1310 N of A/P shear at L5/S1 which are beyond the NIOSH permissible limits of 6400 N for compression (Waters et al. 1993) and occasional exposure limit of 1000 N for A/P shear (Gallagher

and Marras 2012). Since the constants represented a high bound for occupational loading, they may be used as universal constants for normalisation outside of this study.

Overall, the nEMG-based index results were as expected in regards to previous studies and comparable to the standard index findings from Le, Aurand et al. (2017). The expected findings involved tasks requiring higher levels of control such as precision placement (Davis et al. 2002), asymmetric lifting (Marras et al. 2004), and pushing speed and level of control during pushing (Marras, Knapik, and Ferguson 2009). Since the focus of this paper is primarily on the comparisons between the model-independent method with the model-dependent method, further discussion of the statistically significant differences between conditions can be found in Le, Aurand et al. (2017). For the purposes of visualising the comparable magnitudes and trends, a sample of the index predictions of the nEMG approach versus the model approach between pushing tasks can be found in Figure 3. Although the findings were expected, the novelty of the approach resided in its ability to provide insight on the neuromuscular efforts for a variety of multi-planar dynamic exertions as a system, which had not been previously explored. The finding that the nEMG-based method gave similar results to the standard approach highlights the strength of this technique to



**Figure 1.** Sample continuous data from (a) lifting/lowering at 90 degrees of asymmetry with precision placement at mid-thigh height and (b) pushing at fast speed and heavy weight. Note the peaks occurring (a) for lifting/lowering during precision placement and (b) when the momentum of the cart needed to be stopped at the end of the task.

approximate a coactivation index for complex dynamic tasks without the benefit of a biomechanical joint model.

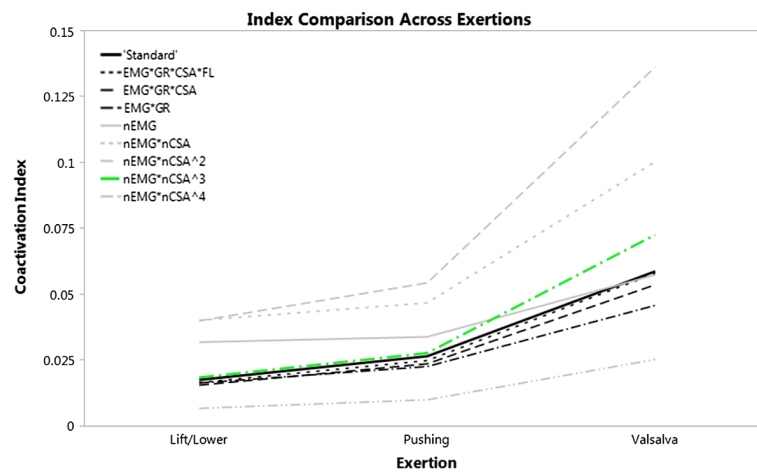
## 5. Discussion

The purpose of this study was to develop a nEMG-based, model-independent coactivation index with the lowest number of components possible to differentiate between various complex dynamic tasks. The intent of this approach was to provide a meaningful and concise way to describe coactivation from a systems perspective (index) when a biomechanical model is not accessible across a variety of tasks within a range from 0 to 1 with extreme cases possibly exceeding 1. This objective was

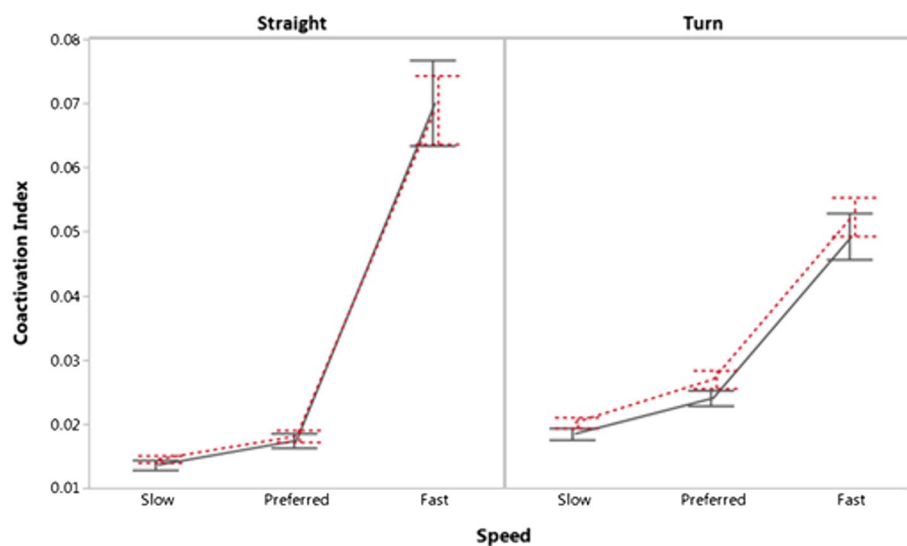
achieved through two stages: (1) by stepwise reduction of the modulating components of the force equation (Equations (1) and (2)) developing a nEMG-based approach independent of the biologically assisted model and then comparing the continuous output to the standard index extracted from the model using the coefficient of determination ( $r^2$ ). In general, the nEMG approach chosen can differentiate between complex dynamic tasks while exhibiting similar magnitudes and trends to the standard index. The advantages of having a model-independent approach are to provide a method with reduced computational complexity and allow for 'universal' application in the assessment of coactivity during complex dynamic tasks.

### 5.1. Component reduction (model dependent)

The reduction of components from the active force equation (Equation (1)) resulted in reduced comparability relative to the standard approach. Removal of the force-velocity and force-length modulation factors resulted in small reductions to the coefficients of determination with lifting/lowering most affected. This was likely due to the range of motion required to move the weight at wider asymmetries or deeper torso flexion. Muscle lengths and velocities adapt to the complex postures to accomplish the task. These changes are highly interdependent with muscle activity due to their relation with the frequency of excitation of active muscle fibres (Bigland and Lippold 1954). Therefore, force-length and force-velocity relationships are necessary to fine tune muscle force output through the understanding of length changes and muscle velocity during eccentric/concentric loading, respectively. When muscle CSA was also extracted from the force equation, a 17% decrease in overall comparability was seen relative to the standard index. CSA is a necessary component due to its relation to maximal force output (Gungor et al. 2015; Maughan, Watson, and Weir 1983) which is typically described as the product of gain and CSA and occur between the physiological boundaries of 30 and 100 N/cm<sup>2</sup> (McGill and Norman 1987). The extraction of the gain ratio contribution reduced the comparability by another 26%. The gain ratio is a factor in the equation represented as the gain divided by the MVC (Dufour, Marras, and Knapik 2013). This empirically validated approach eliminated the need for the measurement of MVCs. Although there was a reduction in comparability, the index derived from the EMG\*GR method was still capable of differentiating between complex dynamic tasks (Figure 2). Overall, the active force equation was found to be highly dependent on the gain ratio and CSA. An approach with a similar scaling factor was needed to modulate muscle activity independent of the biologically assisted model.



**Figure 2.** Mean comparisons between the different reduced component and nEMG indices. The solid black line represents the standard index and green dash-dotted line represents the model independent approach chosen. Note the similar trends for lifting/lowering and pushing tasks for the nEMG\*nCSA<sup>3</sup> method compared to the standard index.



**Figure 3.** Coactivation index calculation comparison between the nEMG\*nCSA<sup>3</sup> model-independent method (dotted line) and 'standard' model-dependent approach (solid line) for the pushing task interaction of speed and push type (mean  $\pm$  SE). Note the similarities in trends and magnitudes.

## 5.2. EMG-based approach (model independent)

Using torso kinematics, normalised EMG (nEMG) and anthropometric-defined moment arms, CSAs and muscle lines of action, an nEMG-based coactivation index was developed. This approach allowed for the calculation of the coactivation index independent of the model. The muscle lines of action for the nEMG-driven approach were based upon a straight-line assumption driven by the kinematics of the torso and the magnitude of the nEMG relative to its respective moment arm. The standard approach was model-dependent and utilised a muscle wrapping method, which allowed for a wider assessment of complex, asymmetric postures (Hwang et al. 2016). Considering the difference between methodologies, the nEMG-based

approach chosen can produce similar outcomes as the model-based EMG\*GR\*CSA approach ( $r = 0.88$ ). The key to the nEMG method was its scaling by the CSA. As previously discussed, CSA is associated with force output. Hence, it was postulated that an understanding of trend of maximal force output relative to the CSA in the spinal muscles would assist in scaling the nEMG and may be determined as the product between gain and CSA. Since the latest version of the biologically driven model utilised a GR, it was not possible to empirically derive the maximum muscle forces for comparison because the equation encapsulated an optimised MVC. Therefore, data from a previous model incorporating MVCs were utilised to compare max force relative to CSA (Le et al. 2012). Based upon



the independent data-set, a general polynomial trend was observed, thereby prompting the idea of incrementing the exponential value (Equation (11)) to examine the fits relative to the 'standard' index. The normalisation of the CSA was utilised to test the approach on other multiple-muscle systems. Scaling of the nEMG was necessary to dampen the magnitude of the simulated force and its effect on the defined moment and relative agonist/antagonist classification. The consequence of not scaling the nEMG in the equation was misclassification of smaller muscles (ie rectus abdominis) in relation to their moment arm. For example, a normalised erector spinae (ES) EMG value may be 0.5 V/V and rectus abdominis (RA) 0.35 V/V during a lift, but because of the longer moment arm of the RA, the simulated moment calculation would have a higher result than the ES. Realistically, due to the size of the ES and its smaller moment arm, the force/moment output would be much larger than the RA. Hence, the influence of smaller muscle size needs to be adjusted accordingly. The findings showed that the resolution of the EMG-based index was best when using  $nEMG \cdot nCSA^3$  to represent the simulated force. Overall, this method could differentiate between the different complex dynamic tasks comparable to the standard with a limitation involving lifts requiring axial twisting at chest height. Hence, the straight-line approach as used for the nEMG-method may be limited to tasks within a smaller range of motion. This limitation is due to muscle line of action assumptions during highly asymmetric postures when using the straight-line approach (Marras et al. 1993). Based upon the nEMG index findings relative to the wrapping model standard index, pushing tasks appeared to match well and lifting/lowering was comparable for sagittal chest height lifts and all asymmetries at mid-thigh height.

### 5.3. Limitations

A few limitations need to be considered for the application and interpretation of the index. First, the combinations of tasks were limited in scope. Lifting was limited to one-sided asymmetry and two heights, thus assuming symmetry between the sides. Pushing was limited to one type of cart and handle height. Although it is understood that many other factors during pushing and lifting could play a role in changing coactivation patterns, the main purpose of the study was to control tasks for external validity to evaluate the sensitivity of the indices for separating tasks. Second, the nEMG-based index was dependent on the moment arms and CSAs derived from a set of anthropometrically dependent regressions. The use of a different set of moment arms may affect the findings and misclassify muscle contribution. Third, although there was ample time to practice the tasks, variability was found due

to anthropometric differences, strength, and fear of failure during precision tasks. Despite these limitations, the nEMG-index was still sensitive to the conditions tested and supported by previous findings while being independent of the model.

## 6. Conclusions

This study has presented a method to assess coactivation of a multiple-muscle system for the assessment of complex dynamic tasks independent of a biologically assisted model. The nEMG-based index developed provides an understanding of the neuromuscular effort of various tasks comparable to the index calculated from high-end modelling. The approach is based upon anthropometric, regression-defined moment arms and continuous classifications of antagonist/agonist activity. Experimental testing of the nEMG-based index was compared to a model-dependent standard index and demonstrated its effectiveness in differentiating between tasks. Future studies would test the application of the nEMG-based methodology on other multiple-muscle systems such as the cervical spine and the upper and lower extremities.

## Disclosure statement

No potential conflict of interest was reported by the authors.

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