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Journal Article

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Publication date:
2024-05

Permanent link:
https://doi.org/10.3929/ethz-b-000670487

Originally published in:
Journal of Biomechanics 168, https://doi.org/10.1016/j.jbiomech.2024.112039

Journal of Biomechanics 168 (2024) 112039

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech

Evaluation of trunk muscle coactivation predictions in multi-body models

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ARTICLE INFO

Keywords:
Spine biomechanics
Muscle coactivation
Musculoskeletal simulation
Multi-body modelling
Computational study

ABSTRACT

Musculoskeletal simulations with muscle optimization aim to minimize muscle effort, hence are considered unable to predict the activation of antagonistic muscles. However, activation of antagonistic muscles might be necessary to satisfy the dynamic equilibrium. This study aims to elucidate under which conditions coactivation can be predicted, to evaluate factors modulating it, and to compare the antagonistic activations predicted by the lumbar spine model with literature data.

Simple 2D and 3D models, comprising of 2 or 3 rigid bodies, with simple or multi-joint muscles, were created to study conditions under which muscle coactivity is predicted. An existing musculoskeletal model of the lumbar spine developed in AnyBody was used to investigate the effects of modeling intra-abdominal pressure (IAP), linear/cubic and load/activity-based muscle recruitment criterion on predicted coactivation during forward flexion and lateral bending. The predicted antagonist activations were compared to reported EMG data.

Muscle coactivity was predicted with simplified models when multi-joint muscles were present or the model was three-dimensional. During forward flexion and lateral bending, the coactivation ratio predicted by the model showed good agreement with experimental values. Predicted coactivation was negligibly influenced by IAP but substantially reduced with a force-based recruitment criterion.

The conditions needed in multi-body models to predict coactivity are: three-dimensionality or multi-joint muscles, unless perfect antagonists. The antagonist activations are required to balance 3D moments but do not reflect other physiological phenomena, which might explain the discrepancies between model predictions and experimental data. Nevertheless, the findings confirm the ability of the multi-body trunk models to predict muscle coactivity and suggest their overall validity.

1. Introduction

Multi-body models of the spine have been used in a wide range of clinically relevant problems to shed light on the biomechanics of the healthy and pathological spine. Several models of the lumbar region of the spine have been described (Bogduk et al., 1992; Cholewicki et al., 1995; de Zee et al., 2007; El-Rich et al., 2004; Huynh et al., 2010; Lambrecht et al., 2009; McGill and Norman, 1987; Shirazi-Adl, 1991) and have become increasingly complex and accurate thanks to the modelling of muscle force–length and force–velocity relationships (Christophy et al., 2012), disc stiffness and ligaments (Abouhossein et al., 2011; Han et al., 2012) and intra-abdominal pressure (Han et al., 2012). The intervertebral joints, initially modelled as spherical, were enhanced by allowing translational displacements (Abouhossein et al., 2011; Ghezelbash et al., 2015), and estimation of load sharing between anterior column and facet joints was suggested (Abouhossein et al., 2011). Bruno et al., (2015) proposed a new method to adjust muscle moment arms and physiological cross-sectional areas based on medical images to achieve a more realistic muscle architecture. In recent years, musculoskeletal spine models have been further developed to model complex 3D deformities and investigate biomechanical loads (Barba et al., 2021; Bassani et al., 2024; Schmid et al., 2020). Although these models have become progressively more detailed, they are based on multiple assumptions and limitations that might affect their predictions and should therefore be critically evaluated.

One of the challenging aspects related to the modelling of musculoskeletal systems is a representation of muscle co-contraction, i.e. the simultaneous activation of muscles controlling movement in opposite directions. Multi-body models with muscle optimisation allow to solve the load-sharing problem and estimate the forces of individual muscles by minimising an objective function subject to dynamic equilibrium equations and additional constraints. Objective functions are defined to mimic the physiological muscle recruitment of the central nervous system, such as the minimization of the muscular effort in the system

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https://doi.org/10.1016/j.jbiomech.2024.112039
Accepted 4 March 2024
Available online 15 March 2024

(Rasmussen et al., 2001). As antagonist activation is disadvantagous from the effort minimization perspective, multi-body simulations based on optimisation are generally not expected to predict muscle coactivation (Dreischarf et al., 2016; Marras, 1988). Yet antagonistic activations were shown in previous studies involving optimisation strategies. For example, mathematical analysis of the general non-linear optimisation problem have demonstrated that any three-dimensional single-joint model can predict the co-contraction of single-joint muscles (Jinha et al., 2006). Musculoskeletal modeling studies of the spine have reported antagonist activation (Ignasiak et al., 2018, Stokes and Gardner-Morse, 1995) and suggested a link with the presence of multi-joint muscles in the model (Stokes and Gardner-Morse, 1995). However, these findings are not widely known or accepted, and prediction of muscle coactivation with optimization-based models remains debated. Some studies consider the prediction of antagonistic activity as a limitation of optimization (El Ouaaid et al., 2013) and linear optimization approaches (Pedersen et al., 1987; Zajac and Gordon, 1989), while others believe that it depends on the three-dimensionality of the model (Pedersen et al., 1987) or the presence of multijoint muscles in the system (Herzog and Binding, 1992).

The prediction of trunk muscles co-contraction is crucial for the accurate modeling of the spine function, relevant for clinical applicability of the models. Experimental data supported the theoretical hypothesis that co-activity of antagonistic muscles is motivated by stability (Granata and Orishimo, 2001) and leads to increased joint stiffness (Cholewicki et al., 2000; Gardner-Morse and Stokes, 2001). This becomes particularly significant in an elderly population characterized by a loss of abdominal strength and thus a threat to stability. Increased antagonist activity is related to increased spinal loads, a risk factor for the development of low back pain, disc degeneration, muscle fatigue and injuries (Chaffin and Andersson, 1991; Greig et al., 2014; Marras et al., 2001). Although to date many of these clinical problems are investigated with musculoskeletal models of the spine, the literature on the simulation of trunk muscle coactivity in musculoskeletal models is controversial and poorly understood.

Therefore, the main aim of this study is to provide a better understanding of the problem of antagonistic muscle predictions in multibody models by performing simulations of simple rigid body systems and a complex musculoskeletal spine model, comparing these predictions with experimental data and investigating factors that may influence them. The specific aims include:

1. Investigation of the conditions needed to predict muscle coactivation, using simple rigid-body models.
2. Evaluation of the influence of factors modulating prediction of antagonistic activations (type of muscle recruitment function and presence of intra-abdominal pressure model), using a more complex realistic lumbar spine model.
3. Qualitative validation of the antagonistic activations predicted with the lumbar spine model against EMG measurements reported in the literature.

2. Methods

2.1. Musculoskeletal modeling

2.1.1. Simple synthetic models

Simple rigid-body systems were constructed and simulated to investigate which conditions allow co-contraction prediction, using a commercial software for three-dimensional multi-body dynamic simulation, AnyBody Modeling System (AnyBody Technology, AnyBody, Aalborg, Denmark). The models consisted of two or three segments, depending on the presence of single- or multi-joint muscles, respectively. The muscles originated in the uppermost segment and inserted in the lowest segment. The mass of each segment was 10 kg. One segment of the model was constrained to the ground. The remaining joints were defined as one degree-of-freedom hinge joints, or as three degrees-of-freedom spherical joints, to represent a planar or a three-dimensional model, respectively. The planar model had therefore one or two degrees-of-freedom, balanced by two single- or multi-joint muscles, respectively, whereas the 3D model had three or six degrees-of-freedom, balanced by three or six muscles with three-dimensional paths, respectively. No elastic properties were implemented. Muscles were modelled as simple active force elements extending over one or more joints to represent mono- and multi-articular muscles, respectively.

2.1.2. Lumbar spine model

A previously established lumbar spine model (de Zee et al., 2007) was used with minor modifications to explore antagonistic activations of trunk muscles. The model represents a male anatomy of body height 180 cm and weight 75 kg (Bogduk, 1997; Hansen et al., 2006) and consists of a chain of rigid bodies: pelvis and sacrum, five lumbar vertebrae, a lumped thoracic region, seven cervical vertebrae and a skull segment, and an abdominal sheath. The upper limbs were added to this model (van der Helm, 1994). Intervertebral joints, with joint rotation centres defined according to (Pearcy and Bogduk, 1988), are modelled as three degrees-of-freedom spherical joints. The cervical part is articulated by six spherical joints from T1 to C2 and two hinge joints between the first pair of cervical vertebrae and the skull. No muscles are present in this region, so reaction forces are applied to each joint to support movement. The muscle groups included in this model are: multifidus, erector spinae, psoas major, quadratus lumborum, semispinalis, spinalis, transversus, rectus abdominis, obliquus internus and externus. Each muscle group is divided into a number of fascicles that are modelled as simple force elements spanning over the shortest path between origin and insertion or as a via-node type when accounting for wrapping. The maximum isometric force of each fascicle is calculated as the product of the fascicle's physiological cross-sectional area (PCSA) and 90 N/cm^2. A detailed description of the modelling of the muscles that make up the lumbar spine model can be found in (de Zee et al., 2007). A preliminary model of the intra-abdominal pressure (IAP) is included in the model. The abdominal cavity is simulated by a cylindrical structure bounded by the abdominal muscles and the abdominal sheath. When the transverse muscle is activated, generating a change in the volume of the abdominal cavity (cylinder), it activates the IAP mechanism, which is modelled as an artificial muscle that applies an extension force on the pelvis and thorax segments with a maximum strength equivalent to the maximum abdominal pressure (Han et al., 2012; Liu et al., 2019). To deactivate the mechanism of IAP, the artificial muscle strength was set to zero.

2.2. Simulations

2.2.1. Inverse statics

In order to predict muscle and joint reaction forces, the inverse statics problem is solved, whereby joint forces and torques are estimated from known displacements and rotations by solving a system of Newtonian dynamic equilibrium equations. Due to muscle redundancy, optimization strategies are applied to solve the load sharing problem and estimate individual muscle forces. Muscle recruitment optimization is performed at each stage of the simulation and minimizes an objective function, defined to mimic the physiological neuromuscular recruitment (Damsgaard et al., 2006; Rasmussen et al., 2001).

In this study, inverse-statics simulations were performed (very low velocity) to avoid dynamic effects using AnyBody Modeling System. In the lumbar spine model, the spinal rhythm was based on the measurements of (Wong et al., 2006) for forward flexion and (Rozumalski et al., 2008) for lateral bending. In the reference simulation, the sum of cubed muscle activities was applied as a muscle recruitment objective function, and subsequently the function type was varied (force-based, linear activity-based) to assess its effect on antagonistic activities.

2.2.2. Analyzed antagonistic output

Model-predicted muscle activations were analyzed using both simple rigid-body models and a complex spinal model. In AnyBody, the activity of a muscle fascicle is defined as the ratio of the force it generates to its maximum isometric force ($f^M_{i,max}$). The activity of a muscle group was defined as the average activity of fascicles composing that group.

To assess the degree of trunk muscles coactivation, the coactivation ratio of agonist and antagonist muscles was calculated during the simulated tasks. In forward flexion, the coactivation ratio was defined as a ratio of abdominal to extensor muscles activity and in the right lateral bending as right to left spine muscles activity. The abdominal muscles activity was calculated as the sum of the activities of the external oblique, internal oblique and rectus abdominis muscles; the extensor muscles activity as the sum of the activities of the multifidus and erector spinae muscles; the activity of right (left) spine muscles as the sum of the activities of the muscles belonging to the right (left) side of the spinal column.

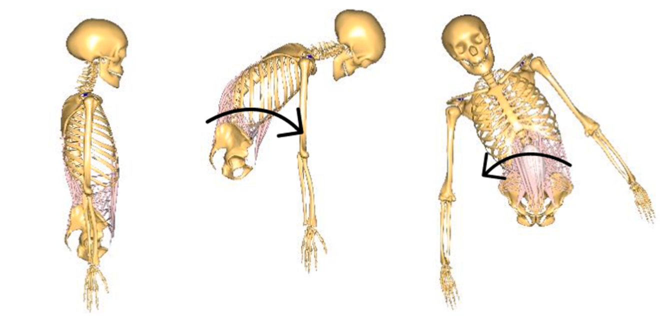
2.2.3. Simulated tasks

2.2.3.1. Simulations of synthetic models. Several dynamic simulations of bending motion of the synthetic models were performed (Table 1), starting from the initial position of vertically aligned segments. The kinematics were prescribed by assigning an angular rotation speed to the joint angles (in degrees per second). In each simulated scenario, also the effect of muscle symmetry was investigated.

2.2.3.2. Simulations of the lumbar spine model. Simulations of the AnyBody lumbar spine model during forward flexion and right lateral bending motion of the spine were performed (Table 2). The pelvis was constrained to the ground and no movement was allowed between pelvis and sacrum. For the forward and lateral bending tasks, an overall trunk inclination angle of 60° and 30° was assigned around the transverse and sagittal axis, respectively. The regional motion was then distributed between the individual vertebrae according to the relationships found in measurements of (Wong et al., 2006) for flexion, of (Rozumalski et al., 2008) for lateral flexion.

Table 2
Illustration of the simulated postures and dynamic tasks.

Upright standing	Forward flexion 60°	Lateral bending 30°



For each task, the influence of factors potentially influencing antagonistic predictions was assessed:

- Model with vs. without IAP
- Linear vs. cubic activity-based muscle recruitment criterion

$$G(f^M) = \sum_i \frac{f_i^M}{f_{i,max}^M} vs. \sum_i \left(\frac{f_i^M}{f_{i,max}^M}\right)^3$$

- Activity- vs. force-based muscle recruitment criterion

$$G(f^M) = \sum_i \left(\frac{f_i^M}{f_{i,max}^M}\right)^3 vs. \sum_i \left(f_i^M\right)^3$$

where G is the objective function of the muscle recruitment problem, f_i^M the muscle force of the muscle i and $f^M_{i,max}$ the maximum isometric force of the muscle i.

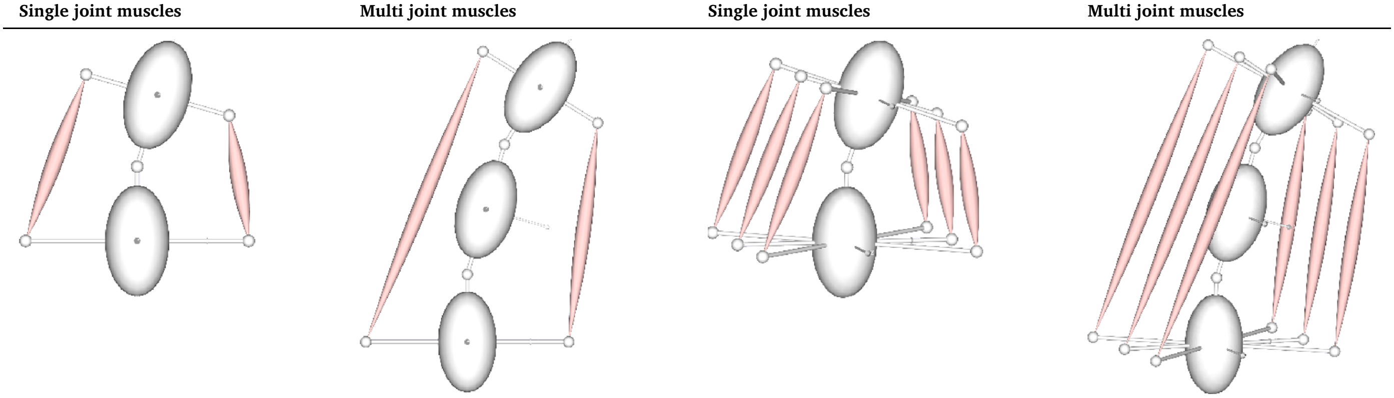
2.2.4. Model validation

To assess the validity of model predictions of muscle coactivity, the state of activation of the antagonistic muscles and the level of coactivation predicted by the model were compared with EMG data reported in the literature.

2.2.4.1. Muscle state. Data on the active or inactive state of muscle groups were obtained from EMG measurements based on the studies of (Arjmand et al., 2010) and (McGill et al., 1999) for forward flexion, (Huang et al., 2001) and (Peach et al., 1998) for lateral bending. The

Table 1
Illustration of simulated scenarios of simple multi-body models. The planar model had one or two degrees-of-freedom, balanced by two single- or multi-joint muscles, respectively. The 3D model had three or six degrees-of-freedom, balanced by three or six muscles with three-dimensional paths, respectively.

2D model Single joint muscles	2D model Multi joint muscles	3D model Single joint muscles	3D model Multi joint muscles



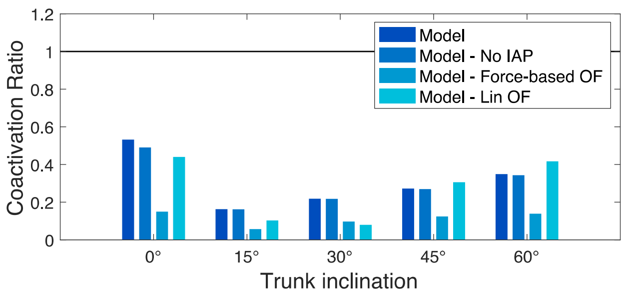


Fig. 1. Predicted coactivation ratios (abdominal muscles activity/extensor muscles activity) for various positions of forward flexion applying different conditions. Model: cubed activity-based objective function with IAP, Model - No IAP: cubed activity-based objective function without IAP, Model - Force-based OF: cubed force-based objective function with IAP, Model - Lin OF: linear activity-based objective function with IAP.

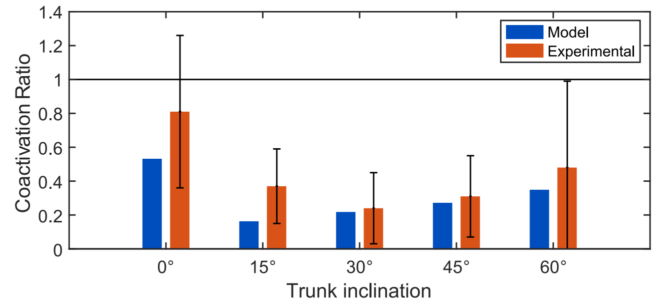


Fig. 2. Comparison of predicted coactivation ratios with experimental data reported by (McGill et al., 1999) for different positions of forward flexion.

experimentally recorded muscle activation state was compared with the state predicted by the model.

2.2.4.2. Coactivation ratio. The level of trunk muscle coactivity during the flexion–extension task was determined from the literature data by calculating the ratio of the sum of the abdominal muscles EMG signals to the sum of the extensor muscles EMG signals. For lateral bending, the ratio of ipsi- (with respect to movement) to contra-lateral EMG was calculated so the coactivation for each muscle group could be quantified. The coactivation ratio obtained from the experimental data was compared with the coactivation ratio calculated from the model predictions.

3. Results

3.1. Simulations of synthetic models

Optimization-based simulations of simplified models showed that the only case when antagonistic activities are not predicted is when a planar model with single-joint muscles is used. On the contrary, simulations of 3D models or models with multi joint muscles predict antagonist activation. The exception is, in absence of single-joint agonist muscles and when agonists and antagonists muscles are symmetrical (perfect antagonist), then the system is indeterminate (see Appendix A).

3.2. Simulations of lumbar spine model

3.2.1. Influence of factors on predicted muscle coactivation

The coactivation ratio of abdominal and extensor muscles at different stages of forward flexion task was almost unaffected by disregarding the IAP model. When a force-based objective function was used, instead of activity-based, the coactivation ratio was reduced by more than 50 % throughout the task (Fig. 1). Simulations with linear (activity-based) muscle recruitment criterion predicted lower coactivity up to 30° of flexion, and higher for larger degrees of flexion, compared to cubic recruitment function.

The effect of IAP on coactivation ratios during lateral bending showed similar trends. Substantial differences were seen by changing the muscle recruitment criterion (see Appendix B).

3.2.2. Model validation

The activation state of the muscles antagonistic to the simulated movements, as predicted by the lumbar spine model, was generally in agreement with the EMG measurements for all muscle groups considered, with the exception of the rectus abdominis muscle. It was not recruited by the model during the flexion forward, in contrast to the literature reporting it to be activated throughout the task (see Appendix C). The model-predicted coactivation ratio varied from min. 0.2 (at 15° flexion) to max. 0.54 (upright), whereas the ratio obtained from the literature was in the range of 0.24 (at 30° flexion) to 0.81 (upright) (Fig. 2). Model-predicted coactivation ratio was found within the standard deviation range of the experimental data during both forward and lateral trunk flexion (Figs. 2 and 3), yet at 15° of lateral inclination the model underpredicted the level of coactivation of the internal oblique muscle.

4. Discussion

Musculoskeletal models provide invaluable insights into the biomechanics of the spine, but their ability to predict coactivation of antagonist muscles remains questioned. Multi-body simulations are considered unable of predicting antagonist activities, as it seems to contradict the muscle optimization concept they are based on. The literature on the topic is controversial, with some works considering the inability to predict co-contraction as one of the major shortcomings of optimisation approaches (El Ouaaid et al., 2013), while others believe it depends on the definition of the optimization algorithm (Pedersen et al., 1987; Zajac and Gordon, 1989). The lack of consensus also concerns the definition of the model and the presence of multi-joint muscles. (Pedersen et al., 1987) suggested that three-dimensionality enables antagonism, which was further supported by (Kaufman et al., 1991), while contradicted by (Herzog and Binding, 1992). The latter study also affirmed that the antagonist predictions depend on a system description involving multi-joint muscles, however, as demonstrated by (Jinha et al., 2006), a planar system with single-joint muscles can also predict antagonist muscle co-activation. Because some of the widely used optimization-based models of the spine, such as AnyBody lumbar spine model, predict muscle co-contraction, this paper aimed to investigate this controversy in more detail.

The main aims were to perform simulations of simplified (synthetic) and complex (realistic) musculoskeletal systems in order to highlight and summarize the conditions that enable predictions of antagonistic activity, validate these predictions in a complex realistic musculoskeletal system, such as a lumbar spine model, and investigate the factors that might influence them.

Simulations of synthetic models showed that predictions of antagonistic activity can occur when multi-joint muscles are present in the model, for both planar and three-dimensional models, unless the

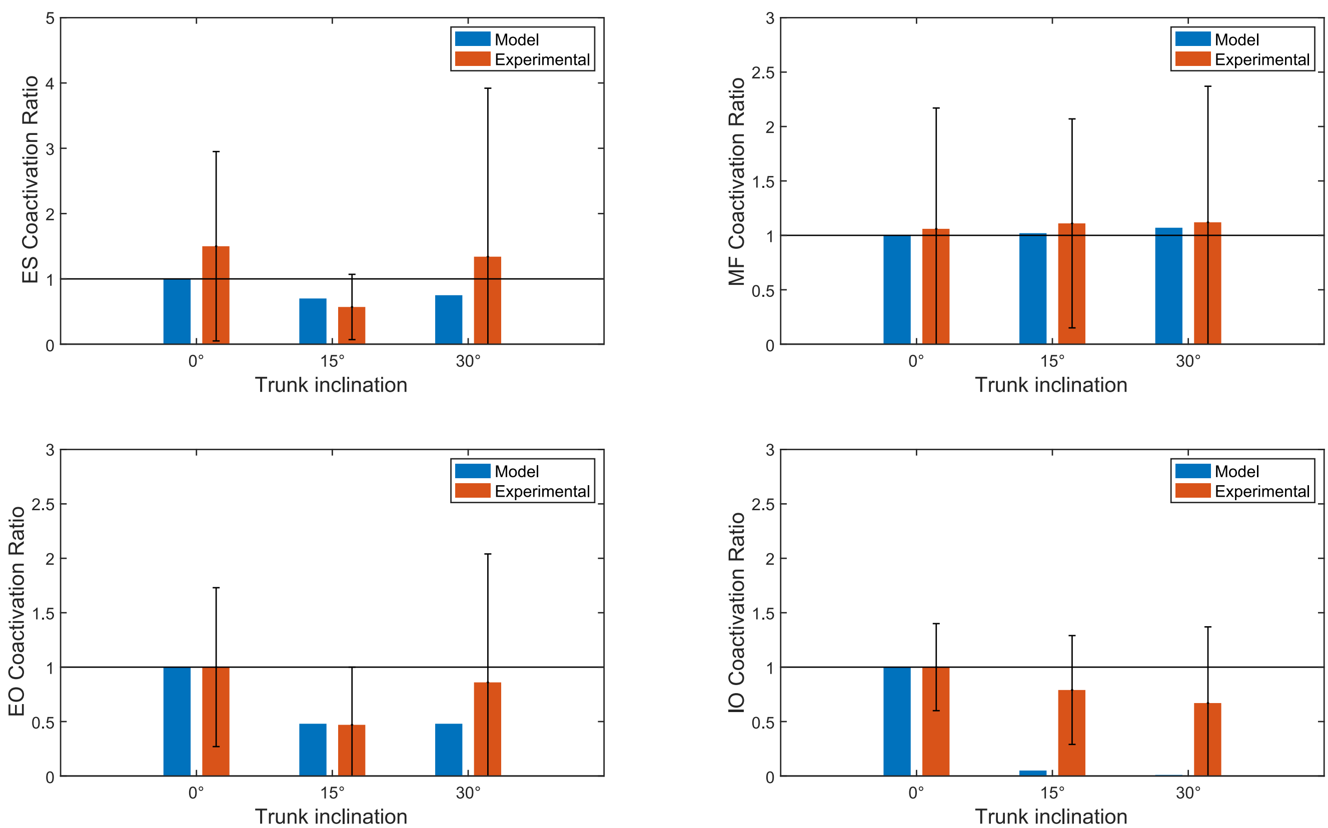


Fig. 3. Comparison of predicted coactivation ratios with experimental data reported by (Peach et al., 1998) for different positions of lateral bending. The coactivation ratio of the rectus abdominis muscle is not shown in the figure, as the model predicts it to be inactive during the task. Muscles: ES - erector spinae, MF - lumbar multifidus, OE - obliquus externus, OI - obliquus internus.

muscles are symmetrical (perfect antagonist). In three-dimensional models, this also applies to single-joint muscles when they are asymmetrical with respect to each other to satisfy the equilibrium around the joint in all three directions. In contrast, no antagonist activity is predicted by a planar model with single-joint muscles only. If one of these conditions between three-dimensionality and the presence of multi-joint muscles is fulfilled, the antagonistic muscle activation is favored by the optimization. This explains the previously observed co-contraction predictions by the spine models such as AnyBody, which are 3D models with multiple muscles spanning over several joints.

The influence of individual factors (IAP, the definition of the objective function and the degree of muscle recruitment criterion) on model-predicted antagonist activations has been explored in this study. The effect of deactivating IAP model on the antagonistic activities was almost negligible. It is expected, as in AnyBody the pressure mechanism is controlled by the transversus muscle, which due to its line of action is not a flexion antagonist. This result is in agreement with the trend observed by (Macintosh et al., 1987; McGill and Norman, 1987), where the effect of IAP had a relatively small effect on trunk biomechanics. On the other hand, a previous study (Cholewicki et al., 2002) reported that high level of IAP is coupled with increased abdominal muscles activity. While the overall effects of the unloading role of the IAP are represented in the AnyBody model (Arshad et al., 2016; Liu et al., 2019), the possible relationship between intra-abdominal pressure and muscle coactivation seems to be neglected. The force-based definition of the objective function led to predictions of non-physiological muscle activities (large errors compared to experimental coactivity ratios), suggesting that the formulation of the optimization problem should consider muscle strength for more realistic antagonistic predictions. In this way - as already generally agreed in the literature (Prilutsky and Gregor, 2000) - larger muscles are recruited to generate greater forces than smaller muscles. The degree of the objective function also influenced the level of predicted co-contraction (expectedly, as it modulates muscle synergies) yet did not determine the presence/absence of antagonist activation. In fact, also simulations of a simplified model with multi-joint muscles predicted antagonist activation independently of a linear or cubic optimization function (results not presented). These findings imply that the model-predicted co-contraction does not depend on the nature of the muscle recruitment problem, but rather on the presence of multi-joint muscles in the model, as previously suggested by (Stokes and Gardner-Morse, 1995).

The qualitative validation of antagonist activities was performed by comparing the muscle activation state and coactivation ratio predicted by the model with EMG measurements reported in the literature, for the main muscle groups. Good agreement was found between model-predicted and experimentally observed muscle coactivation levels and state of muscle recruitment during both forward and lateral trunk flexion, suggesting overall validity of the lumbar spine model in predicting trunk muscle coactivation. Similar patterns of muscle activations were predicted for lateral bending. However, discrepancies observed between model predictions and experimental data should be noted, highlighting the AnyBody – or any multi-body – model limitations. The current models do not reflect all physiological mechanisms of muscle activation that are regulated by the central nervous system. For example, in response to unexpected external perturbations as well as during anticipatory postural adjustments, antagonistic activity tends to increase to make the spine stiffer and thus more stable (Hodges et al., 2000; Van Dieën et al., 2003). This co-contraction results from the activity of the central nervous system and does not respect an energy-preserving muscle recruitment. It cannot therefore be predicted by the model as it only predicts the antagonistic activations that are mechanically needed to solve the dynamic equilibrium equations and balance the three-dimensional joint moments. Some studies have attempted to incorporate abdominal coactivity into the model based on the literature by imposing a priori assumptions. For example, activation of antagonist muscles was prescribed in the model by forcing antagonist muscles to be recruited above certain level (Gardner-Morse and Stokes, 1998; Hughes et al., 1995), or by encouraging antagonist recruitment through a custom optimisation function (El Ouaaid et al., 2013) to promote trunk stability.

Several limitations of the proposed work need to be addressed. Despite the complexity of the AnyBody lumbar spine model, the shortcomings of the applied model and simulations may limit the results. No force–length and force–velocity muscle relationships were taken into account, possibly resulting in overestimated levels of muscle activation, especially at a large degree of flexion. Also, changes in the position of joint centers of rotation up to 60° of flexion were found to affect spinal loads by a maximum of about 13% (Ghezelbash et al., 2015). However, in this study, for sake of computational simplicity, the joint rotation centers were assumed to be fixed. Nevertheless, this is an established and validated model, and our aim was to provide insights into its abilities to predict muscle coactivation, without substantial modifications. Another important limitation arises from the validation of model predictions with reported EMG data. Due to the multilayer architecture of the trunk musculature, EMG recordings may contain muscle signals from different muscle groups, making it difficult to compare with the activities of fascicles in the model. Also, the description of the tasks reported in the EMG publications is rather generic, so despite the attempt to mimic the same exercise with the model, differences in kinematic patterns might influence muscle activations. Therefore, in this study, the comparison with EMG data was not in terms of signal amplitude, but muscle activation state and overall coactivation ratio.

Despite the limitations, this study provides better understanding to the problem of predicting the activity of antagonist muscles in optimisation-based multi-body spine models. It shows that model three-dimensionality or multi-joint muscles are conditions leading to antagonistic predictions. It provides new understanding that activity- vs. force-based optimisation function is a major modulating factor for predicting antagonistic activity, whereas the optimization function type or inclusion of intra-abdominal pressure play lesser role. These findings, supported by a qualitative validation against available EMG data, contribute to expanding the knowledge of musculoskeletal models of the spine and building confidence in their use for future clinical and ergonomic studies.

CRediT authorship contribution statement

Alice Caimi: Writing – original draft, Visualization, Methodology, Investigation, Formal analysis, Data curation, Conceptualization. **Stephen J. Ferguson:** Writing – review & editing, Validation, Software, Resources, Methodology, Funding acquisition. **Dominika Ignasiak:** Writing – review & editing, Visualization, Supervision, Resources,

Project administration, Methodology, Conceptualization.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

Appendix A

Equilibrium equations for planar model with multi-joint muscles

A system consisting of three rigid bodies, connected by hinge joints, fixed to the ground and subject to the force of gravity (G), was used to calculate the muscle forces needed to maintain the system in the position shown in Fig. A1 (a).

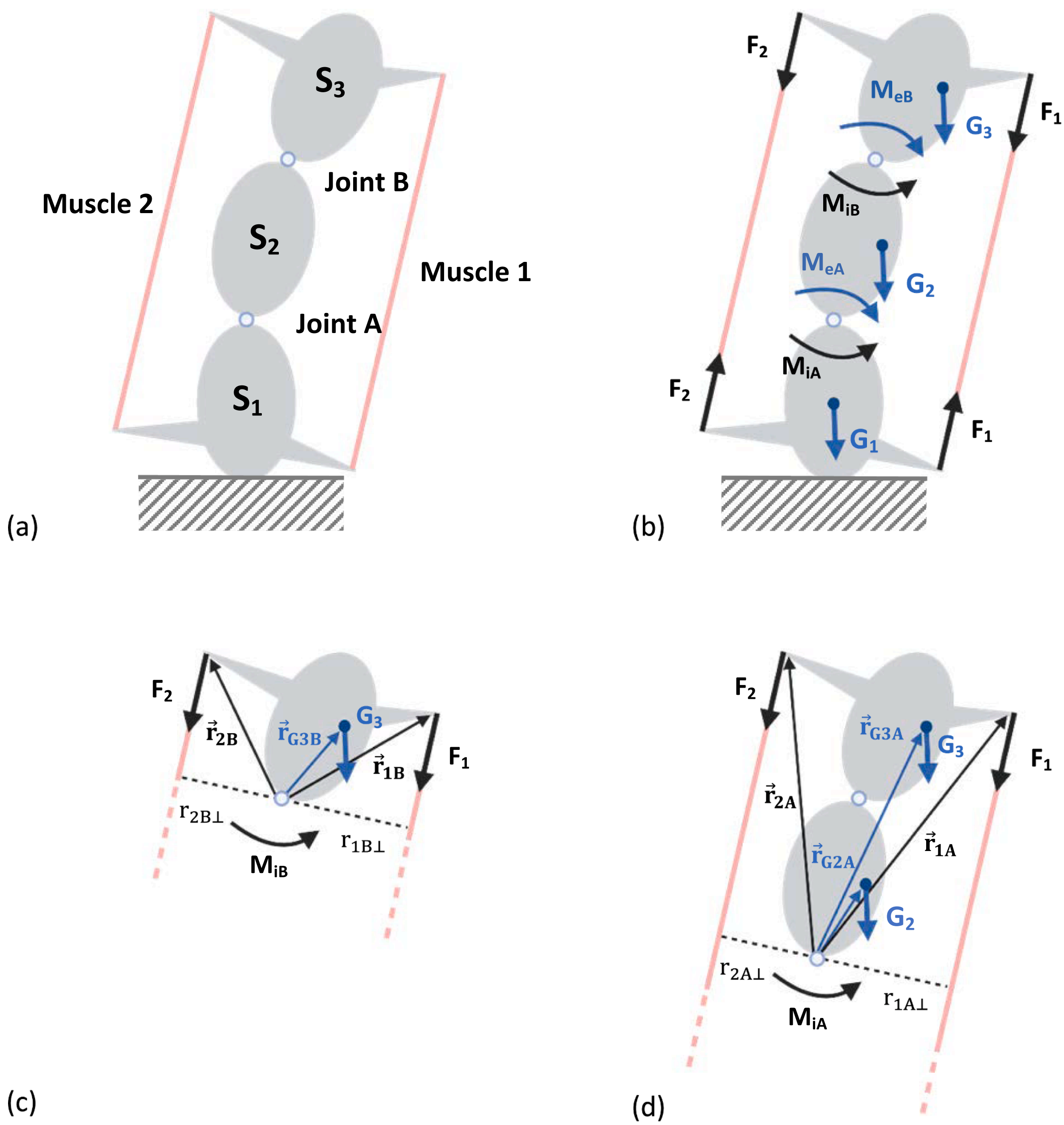


Fig. A1. Force and moments acting on a three-link chain served by two two-joint symmetrical muscles. (a) A three-segment system (S_1, S_2, and S_3 of corresponding masses m_1, m_2, and m_3, respectively) with two biarticular muscles crossing two frictionless revolute joints, A and B. (b) Muscle forces and gravity acting on the system and generating internal and external joint moments, respectively. (c) and (d) Forces involved in the equilibrium equations at joints A and B, respectively, with their respective moment arms.

Assuming the system in static equilibrium, the equilibrium equations at the two joints are:

$$\mathbf{M}_{iA} + \mathbf{M}_{eA} = 0 \tag{1}$$

$$\mathbf{M}_{iB} + \mathbf{M}_{eB} = 0 \tag{2}$$

where the internal muscle moments $\mathbf{M}_i$ counterbalance the external gravity moments $\mathbf{M}_e$.

The internal muscle moments at joint B and A can be calculated as

$$\mathbf{M}_{iB} = \mathbf{r}_{2B} \times \mathbf{F}_2 + \mathbf{r}_{1B} \times \mathbf{F}_1 \tag{3}$$

$$\mathbf{M}_{iA} = \mathbf{r}_{2A} \times \mathbf{F}_2 + \mathbf{r}_{1A} \times \mathbf{F}_1 \tag{4}$$

while the external gravity moments can be calculated as

$$\mathbf{M}_{eA} = \mathbf{r}_{G2A} \times \mathbf{G}_2 + \mathbf{r}_{G3A} \times \mathbf{G}_3 \tag{5}$$

$$\mathbf{M}_{eB} = \mathbf{r}_{G3B} \times \mathbf{G}_3 \tag{6}$$

Considering a planar case, $\mathbf{M} = \mathbf{r} \times \mathbf{F} = |\mathbf{r}||\mathbf{F}|\mathbf{n}\sin\theta$, where θ is the angle between the two vectors and $\mathbf{n}$ is the unit vector perpendicular to the plane formed by $\mathbf{r}$ and $\mathbf{F}$. The magnitude of moment can be then written as: $M = rF\sin\theta = r_{\perp}F$. Therefore, Equation (3) and (4) can be rewritten as

$$M_{iA} = r_{2A\perp} F_2 - r_{1A\perp} F_1 \tag{7}$$

$$Mi_B = r_{2B\perp} F_2 - r_{1B\perp} F_1 \tag{8}$$

Then, multiplying Equation (8) by $r_{2A\perp}$ and Equation (7) by $r_{2B\perp}$ and subtracting member by member yields

$$(M_{iB}\, r_{2A\perp} - M_{iA}\, r_{2B\perp}) = 0 - F_1 (r_{1B\perp}\, r_{2A\perp} - r_{1A\perp}\, r_{2B\perp}) \tag{9}$$

from which

$$F_1 = \frac{(M_{iA}\, r_{2B\perp} - M_{iB}\, r_{2A\perp})}{(r_{1B\perp}\, r_{2A\perp} - r_{1A\perp}\, r_{2B\perp})} \tag{10}$$

With a similar procedure, F_2 can also be obtained and it can be observed that the denominator is the same as F_1

$$F_2 = \frac{(M_{iB}\, r_{1A\perp} - M_{iA}\, r_{1B\perp})}{(r_{1A\perp}\, r_{2B\perp} - r_{1B\perp}\, r_{2A\perp})} \tag{11}$$

Combining Equations (1) and (2) with (5) and (6) it follows that

$$\mathbf{M}_{iA} = -\mathbf{M}_{eA} = -(\mathbf{r}_{G2A} \times \mathbf{G}_2 + \mathbf{r}_{G3A} \times \mathbf{G}_3) \tag{12}$$

$$\mathbf{M}_{iB} = -\mathbf{M}_{eB} = -(\mathbf{r}_{G3B} \times \mathbf{G}_3) \tag{13}$$

Therefore, if the geometry is defined and the gravitational forces are known, the muscle forces F_1 and F_2 can be calculated by Equation (10) and (11). When the geometry of the structure in Fig. A1 causes a null denominator in Equation (10) and (11), i.e. when $r_{1B\perp}\, r_{2A\perp}$ is equal to $r_{1A\perp}\, r_{2B\perp}$, the solution of Equation (10) and (11) does not exist.

A particular case, where this relationship holds ($r_{1B\perp}\, r_{2A\perp} = r_{1A\perp}\, r_{2B\perp}$), is when the lines of action of the antagonist and agonist multi-joint muscles are parallel to each other and parallel to the line connecting joint A and joint B. By approaching this situation, the muscle forces F_1 and F_2 approach an infinite number. This implies that a muscle arrangement in which the antagonist and the agonist muscles are almost aligned with the direction of the two joint centers can lead to very large antagonist and agonist forces to balance the joint moments (for multi-joint muscles). This highlights the necessity of faithful representation of anatomical muscle attachment sites of multi-joint muscles in the multi-body models (such as spine models). It also suggests greater importance of antagonists with line of action markedly different from agonists than “true” (parallel) antagonists, as they can balance the moments with lower force. It needs to be noted that in complex anatomical models, this problem will be complicated by non-linearity of multi-joint muscles path, three-dimensionality, as well as muscles redundancy and recruitment optimization. Nevertheless, this simple analysis demonstrates that the antagonist activity is necessary to balance gravity moments in multi-joint systems, in the absence of agonist single-joint muscles. Static simulations of a three-rigid-body system served by two multi-joint muscles were performed in Matlab to provide a qualitative assessment of the need for antagonism at different degrees of flexion angle (Fig. A2). The muscle forces required to keep the system in static equilibrium were calculated using equations (10) and (11) and are shown in Fig. A3.

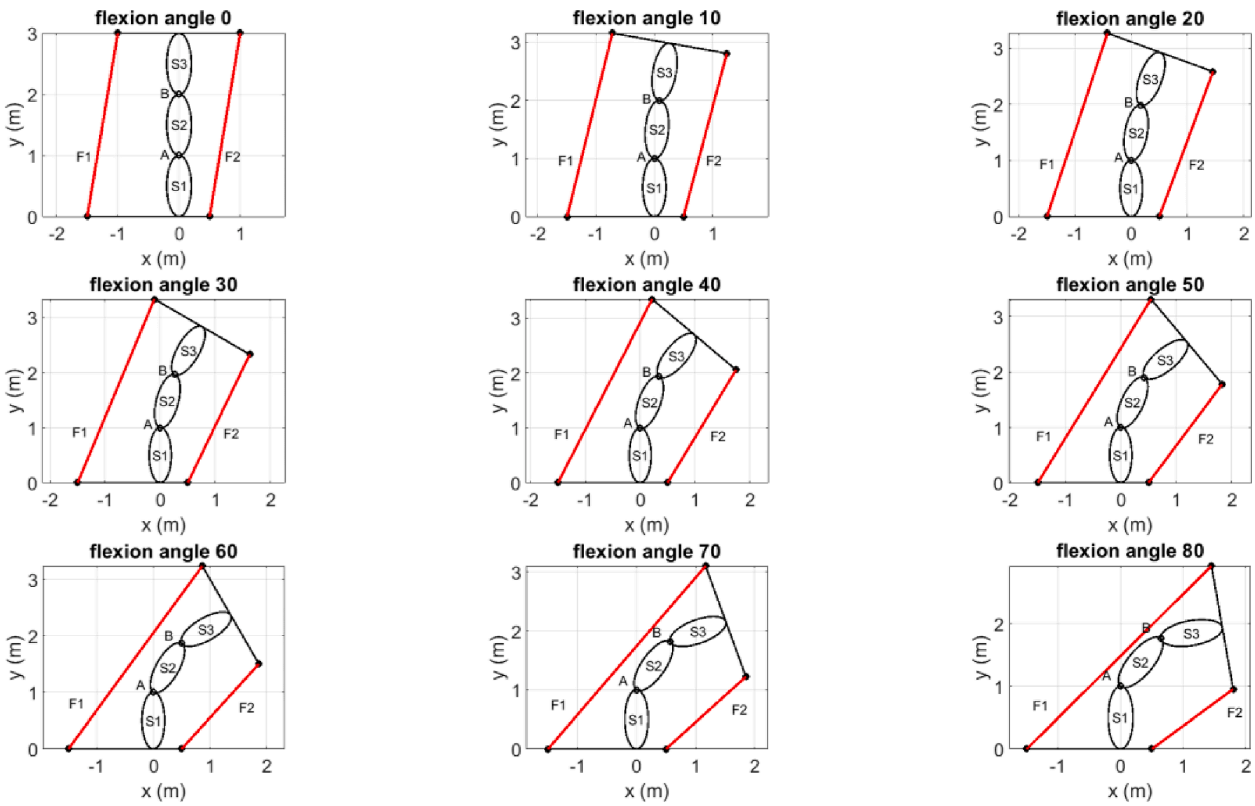


Fig. A2. Illustration of the three-segment system with multi-joint muscles at different flexion angles between 0° and 80°. The flexion angle is evenly distributed between the angles S1-S2 and S2-S3. Segments weight 1 kg, are subject to gravitational force and are connected by two frictionless revolute joints.

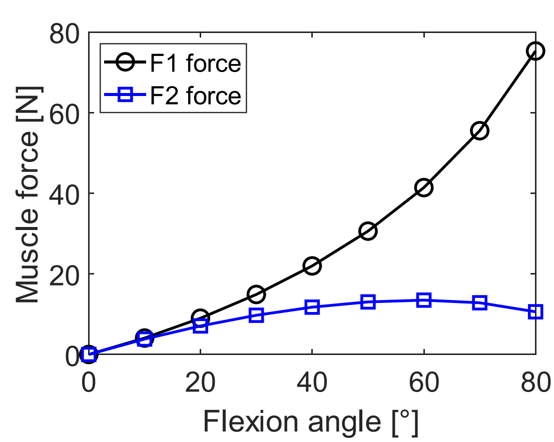


Fig. A3. Forces of muscles F1 (agonist) and F2 (antagonist) at different degrees of flexion angle.

Appendix B

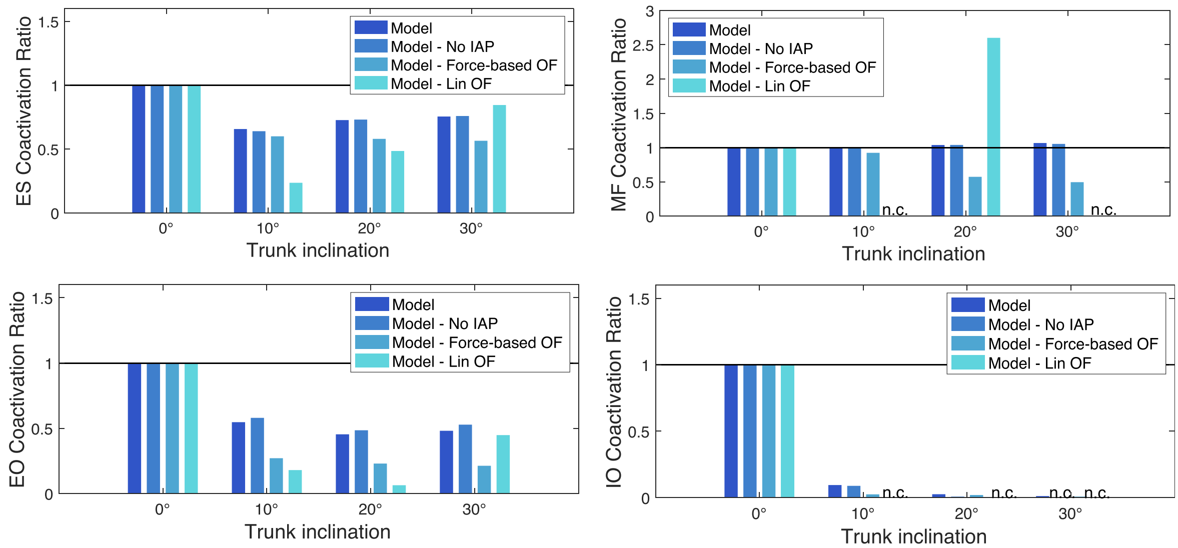


Fig. B1. Predicted coactivation ratios (right spine muscle activity/left spine muscle activity) for various positions of lateral bending applying different conditions. Not calculable (n.c.) if the predicted left or right muscle activity was 0. Model: cubed activity-based objective function with IAP, Model - No IAP: cubed activity-based objective function without IAP, Model - Force-based OF: cubed force-based objective function with IAP, Model - Lin OF: linear activity-based objective function with IAP. Muscles: ES - erector spinae, MF - lumbar multifidus, OE - obliquus externus, OI - obliquus internus.

Appendix C

The activities of various muscle groups predicted by the lumbar spine model are presented side-by-side with EMG measurements during forward flexion (Fig. C1) and lateral bending (Fig. C2) of the trunk. The muscle activity computed by the model (i.e. ratio of generated force to maximum isometric force) does not directly correlate with EMG signal intensity, therefore making direct comparisons is not advised (Lund et al., 2012). Qualitative comparisons should focus only on the state of muscle activation (active / inactive) and – with caution – overall trends, while taking into account the following limitations.

- There is no established relationship between myoelectric activity and muscle force. Various mechanical, physiological, anatomical and electrical changes occur during an anisometric contraction that substantially influence the relationship between signal amplitude and muscle force (De Luca, 1997).
- The multilayer architecture of the trunk musculature causes one of the most common artefacts related to EMG signal recording, the so-called 'crosstalk', which consists of the presence of signals from other muscles superimposed on the signal of the muscle to be recorded. Despite careful selection of electrode placement and size, this phenomenon cannot be eliminated (De Luca et al., 2012).
- The experimental data reported in the literature is electrical potential of EMG signal normalized to that collected during a maximum voluntary contraction (MVC) exercise, while in AnyBody the muscle activity is defined as the muscle force normalized to the assumed maximum isometric force ($f^{M}_{i,max}$) of the muscle.
- The AnyBody model has several limitations. In particular, the muscle model used in the present study does not account for the contractile properties of the muscle sarcomere, the muscle passive properties, or the presence of connective tissues in the muscle arranged in series (tendons) and in parallel (muscle sheaths). These elements determine the force-generating characteristics as a function of the length and speed of change in length of the muscle.

However, the aim of the study was to evaluate the ability of the AnyBody model of the lumbar spine to predict the activation of antagonists' muscles. The attempt of qualitative validation of muscle predictions to literature-reported EMG signals suggests overall model validity, considering the active/inactive state of the muscles, but also highlights its limitations, e.g. flexion-relaxation phenomenon at the larger flexion degree (stabilized by active force muscle in the model). Musculoskeletal simulation models predict co-contraction when the recruitment of antagonist muscles is mechanically favourable, but cannot predict co-contraction resulting from other physiological phenomena such as anticipatory postural adjustments or threatened stability situations.

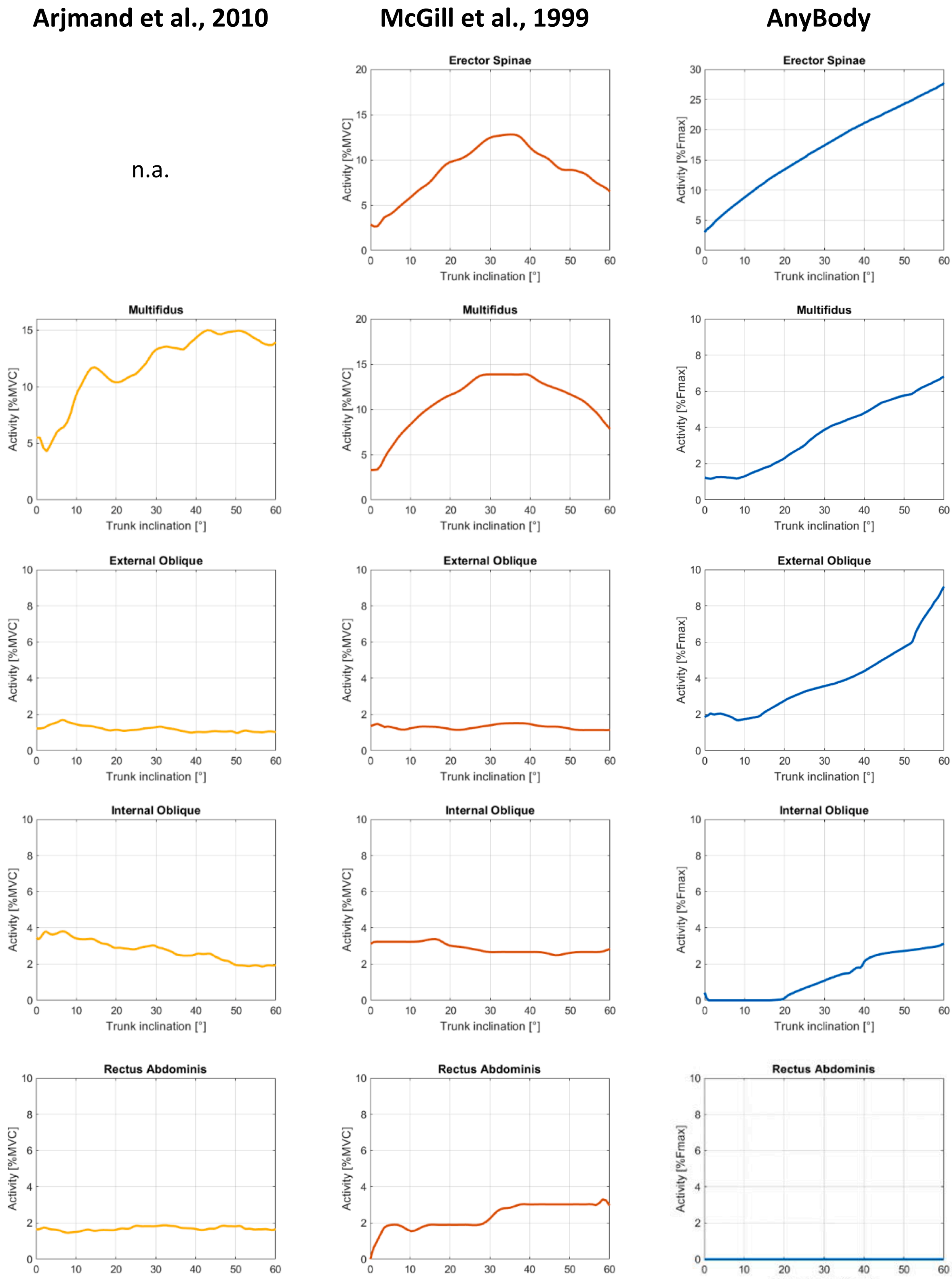
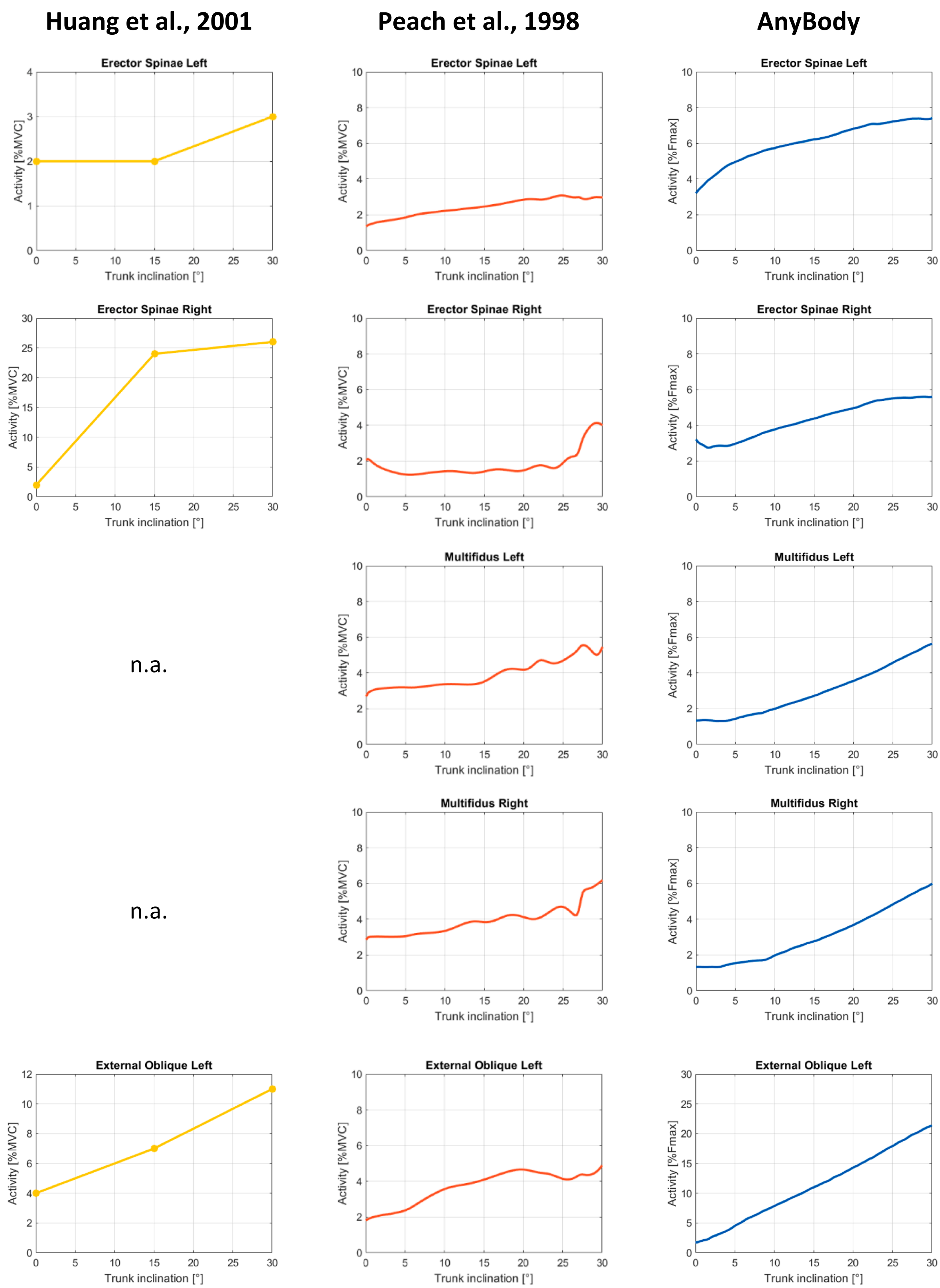


Fig. C1. Muscle activity levels as a function of forward trunk inclination, adapted from the EMG measurements of (Arjmand et al., 2010) (left) and (McGill et al., 1999) (middle) and calculated with the AnyBody lumbar spine model (right).



Huang et al., 2001
Peach et al., 1998
AnyBody
Erector Spinae Left
Erector Spinae Right
Multifidus Left
Multifidus Right
External Oblique Left
Activity [%MVC]
Activity [%Fmax]
Trunk inclination [°]
n.a.
n.a.

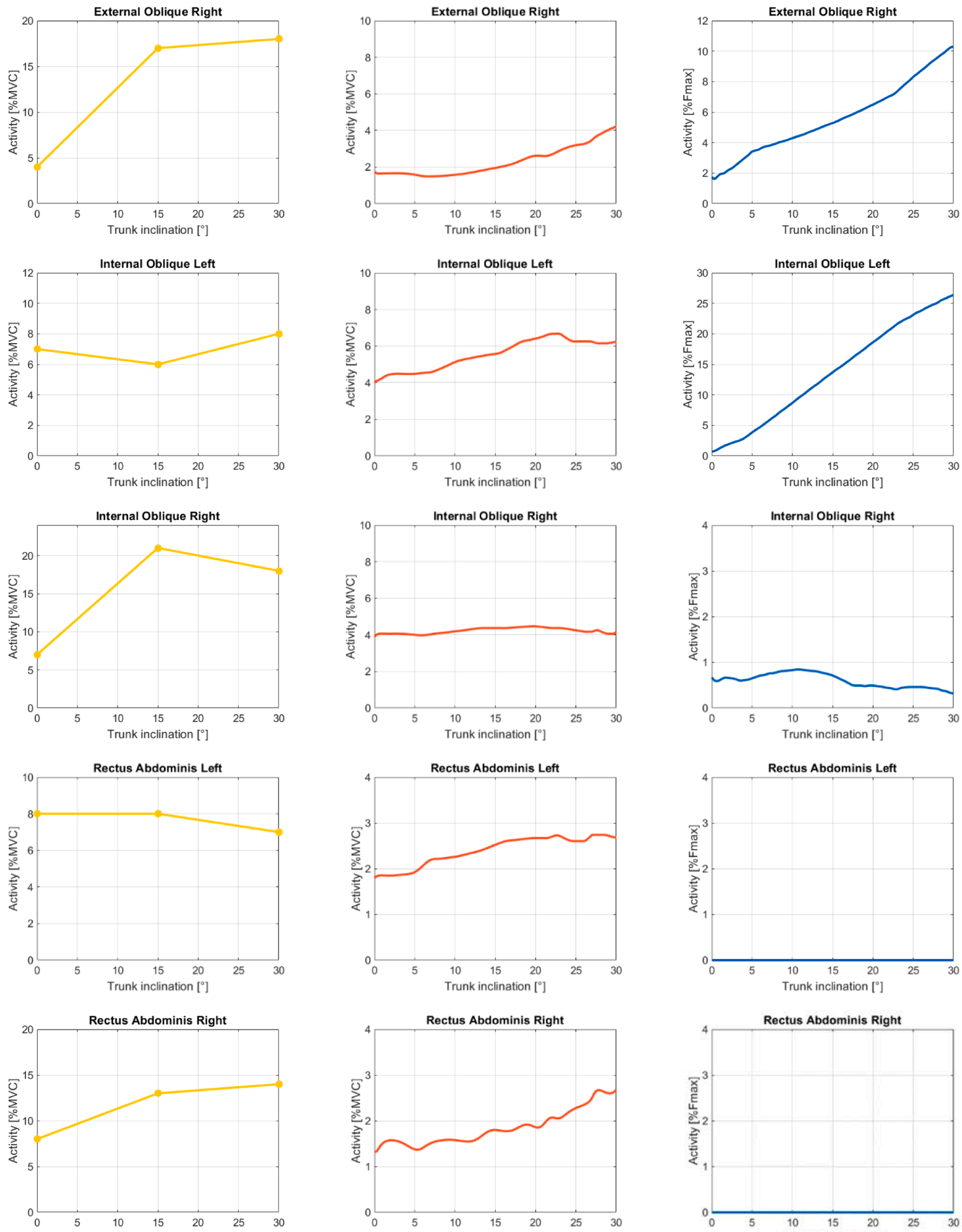


Fig. C2. Muscle activity levels as a function of lateral trunk inclination, adapted from the EMG measurements by (Huang et al., 2001) (left) and (Peach et al., 1998) (middle) and calculated with the AnyBody lumbar spine model (right).

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