

Dynamic Stability of Spine Using Stability-Based Optimization and Muscle Spindle Reflex

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Abstract—A computational method for simulation of 3-D movement of the trunk under the control of 48 anatomically oriented muscle actions was developed. Neural excitation of muscles was set based on inverse dynamics approach along with the stability-based optimization. The effect of muscle spindle reflex response on the trunk movement stability was evaluated upon the application of a perturbation moment. The method was used to simulate the trunk movement from the upright standing to 60° of flexion. Incorporation of the stability condition as an additional constraint in the optimization resulted in an increase in antagonistic activities demonstrating that the antagonistic co-activation acts to increase the trunk stability in response to self-induced postural internal perturbation. In presence of a 30 Nm flexion perturbation moment, muscle spindles decreased the induced deviation of the position and velocity profiles from the desired ones. The stability-generated co-activation decreased the reflexive response of muscle spindles to the perturbation demonstrating that the rise in muscle co-activation can ameliorate the corruption of afferent neural sensory system at the expense of higher loading of the spine.

Index Terms—Dynamic simulation, muscle spindle, spine, stability, stretch reflex.

NOMENCLATURE

Θ	Angular position vector in the inertial coordinate system.
W	Angular velocity vector in the body coordinate system.
$G(\Theta)$	Moment vector arising from gravity.
J_1	Moment of inertia matrix about the center of rotation.
N_{input}	Net muscular torque around L5-S1 in body coordinate system.
WW	Skew symmetric matrix corresponding to W .
F	Vector of muscle forces (48×1).

K	Stiffness matrix of the linearized system (3×3).
V	Viscosity matrix of the linearized system (3×3).
A	Linearized system state matrix (6×6).
N_p	Perturbation moment.
B	Transformation matrix between the body and inertial coordinate systems.
$(\partial L/\partial \Theta)^T$	Matrix of moment arm of muscles (3×48).
l	Muscle length.
\dot{l}	Muscle velocity.
σ_{max}	Maximum muscle stress.
r	Spindle discharge rate (firing rate).
g	Muscle spindle gain factor.
l_d	Desired muscle length.
PCSA	Muscle physiologic cross sectional area.
a_m	Muscle activation level of each muscle ($m=1$ to 48).
f_{max}	Maximum muscle force.
$f_p(l)$	Passive force relationship of a muscle.
$f_l(l)$	Force-length relationship of a muscle.
$f_v(\dot{l})$	Force-velocity relationship of a muscle.

Abbreviations of the simulated muscle groups:

RA	Rectus Abdominus.
EO	External oblique.
IO	Internal oblique.
LT	Longissimus Thoracis.
IL	Iliocostalis Lumborum.
LD	Latissimus Dorsi.
PL	Pars Lumborum.
PS	Psoas.
QL	Quadratus Lumborum.

I. INTRODUCTION

DYNAMIC simulation of spinal movement is a powerful approach for investigating how the neuromuscular elements interact to produce and stabilize spinal movement. Dynamic trunk motion has recently been identified as a potential risk factor for low back disorders [1], particularly when the motion occurs in multiple planes simultaneously [2], [3].

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The imposed trunk moments have been observed to increase by 15%–170% during dynamic exertions. Greater trunk moments occur along with increased agonistic and antagonistic muscle activities resulting in greater spinal loads [1]. McGill and Norman [4] reported an averaged moment increase of 19% in their dynamic model from that of the static one. The stability analysis of a dynamic model of spinal movement is helpful to more accurately estimate forces and moments as well as antagonistic muscle activities during a fully dynamic task. Recently, Shirazi-Adl and colleagues [5], [6] developed a kinematic-based finite element model of trunk to evaluate the contribution of active and passive tissues to the spine loading and the stability in different postures. However, their analyses were static and did not account for the acceleration or velocity of the trunk movement. The stability of spine during various dynamic activities has previously been investigated in a model study taking the determinant of the Hessian matrix as the index of stability [7]. In a more recent study, the nonlinear analysis as well as linear eigen-value and perturbation analyses at loaded-deformed configurations were used to estimate the relative stability of the spine in squat and stoop dynamic lifts [8].

The contribution of neuromuscular elements during dynamic activities has been estimated by static optimization and inverse dynamics approach which are computationally efficient methods, especially when many muscles are included in the model [9], [10]. However, traditional optimization models of the trunk poorly predict antagonistic muscle co-activity essential to enhance stability of the trunk during dynamic tasks [11], [12]. In addition, few studies have quantitatively examined the association between the dynamic spinal stability and the antagonistic co-activation. Gardner–Morse and Stokes [13] investigated the effect of abdominal muscle co-activation on the lumbar spine stability by raising the lower bound on the antagonistic muscle forces. They found that the stability is increased by the abdominal muscle co-activation. Similar method during dynamic tasks has been implemented by other research studies [14], [15]. The direct incorporation of the stability criterion, as an additional constraint in the optimization model, seems to improve the prediction of antagonistic activities as well as the spine loading [16]–[19].

In the spine neuromuscular system, two mechanisms are assumed to provide the required stiffness and stability.

1) *Intrinsic Mechanism*: Provides appropriate stiffness to the trunk by preactivating muscles in anticipation of any perturbation. In other words, it is assumed that the central nervous system assigns muscular activation patterns in a forward manner so that the system can overcome any perturbation [20]. Stability-based optimization models are supposed to provide intrinsic stiffness to the trunk and partially contribute to its stability in advance of any disturbance.

2) *Reflexive Mechanism*: Stiffens the trunk and contribute to its stability with proper feedback in response to the perturbation. It has been shown that intrinsic stiffness alone is insufficient to stabilize trunk activities in presence of disturbances [21]. Muscle spindles via “stretch reflex” are responsible for providing muscle’s reflexive responses. In reality, reflex response changes the muscle activation and hence its intrinsic stiffness

and viscosity [22]. More importantly, the proportional and differential reflexive gain of feedback can enhance the muscle stiffness and joint stability without the metabolic cost associated with intrinsic mechanism described above. The viscoelastic response of passive tissues such as ligaments, joint capsules, and discs can also be viewed as positional and velocity feedback [23], [24]. But in this study, we have not considered them to keep the equations of motion as simple as possible without loss of generality.

The goal of this study was to develop an efficient computational method for simulation of 3-D dynamic movements of the trunk and its dynamic stability using both the intrinsic mechanism provided by the stability-based optimization and the reflex mechanism provided by the muscle spindle’s action. The hypothesis that co-activation can reduce the need for reflex contributions was also investigated.

II. METHODS

A. Trunk Model

In this study, the trunk was treated as an inverted pendulum constrained at its base (L5-S1 joint) with a ball and socket joint which can revolute in 3-D space under the control of 48 muscle fascicles. The effect of gravity was considered as a concentrated load (350 N) at the center of mass with a distance of 35 cm from the pendulum center of rotation [25]. The anatomical and geometric data of muscle fascicles were based on the literature [7]. The list of muscle abbreviations used in our simulations is provided in Nomenclature. Only muscle fascicles that can create moment about the L5-S1 joint were included in this study. The dynamic equations of motion of this inverted pendulum were derived in a compact form [14], [16], [26]

$$J_1 \dot{W} = -WWJ_1W + N_{\text{input}} - G(\Theta) \quad (1)$$

in which W is the angular velocity vector, Θ is the angular position vector, $G(\Theta)$ is the moment vector arising from gravity, J_1 is the matrix of moment of inertia, and N_{input} is the net muscular torque about the L5-S1 in body coordinate system. WWJ_1W represents the torque due to coriolis forces.

B. Muscle Model

Muscle forces were computed by scaling the active force-length, $f_l(l)$, and force-velocity, $f_v(\dot{l})$, relations by the muscle activation level, a , and adding the passive force, $f_p(l)$

$$f = f_{\text{max}} \cdot \{ a \cdot f_l(l) \cdot f_v(\dot{l}) + f_p(l) \} \quad (2)$$

where l is the absolute muscle length, \dot{l} is the muscle contraction or stretch velocity, and f_{max} is the maximum muscle force which is estimated by multiplying the muscle physiologic cross sectional area, PCSA, the with maximum muscle stress, σ_{max} , taken as 0.55 MPa for all muscles [5], [6], [14].

C. Spindle Model

Regulatory behavior of the muscle spindle in response to the muscle stretch has been studied in many research works [27]–[29]. Stretch reflex is one of the most important neuromuscular functions in controlling posture and movement. The

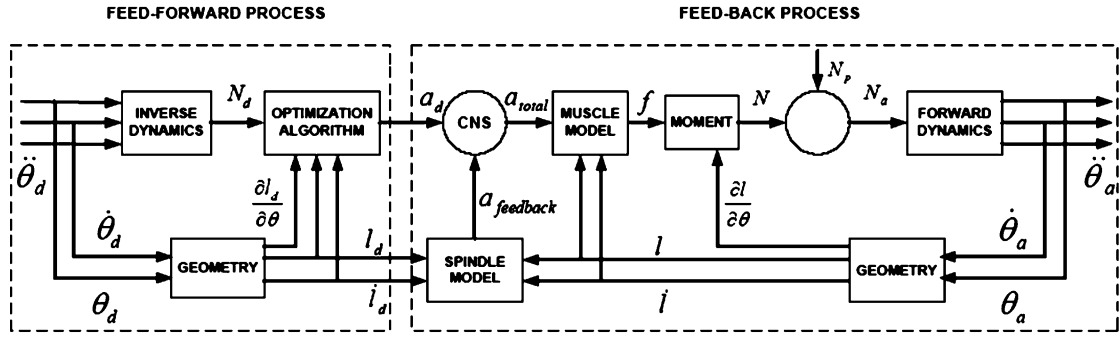


Fig. 1. Computation algorithm. Feed-forward process: muscle neural activations are computed based on the desired kinematics with or without application of the stability constraint in the optimization routine. Feedback process: effects of muscle spindles on total muscle activations and the kinematics profiles of the movement are evaluated with or without application of the perturbation moment. θ_a , $\dot{\theta}_a$, and $\ddot{\theta}_a$ are the actual kinematics, θ_d , $\dot{\theta}_d$, and $\ddot{\theta}_d$ are the desired kinematics, a_d is the desired activation, N_d , N_a , N_p are the desired, actual, and perturbation moments. A delay of 20 ms is applied in the transmission of feedback signals.

model of stretch reflex invoked in our algorithm was derived from earlier works [28], [30]

$$\begin{aligned} r &= g(l - l_d)\dot{l}^{-3} \quad \text{when } l - l_d > 0, \dot{l} > 0 \\ r &= 0.1g(l - l_d) \quad \text{when } l - l_d \geq 0, \dot{l} \leq 0 \end{aligned} \quad (3)$$

where r is the spindle discharge rate (firing rate), l is the muscle length, \dot{l} is the muscle velocity, g is the spindle gain factor, and l_d is set to be the desired muscle length computed from the desired kinematics at each instance of simulation time. Physiologically, the value of gain may vary from one muscle to another and even for a single muscle during different tasks [31]. A linear transformation was adopted from the earlier studies to map the spindle firing rate r to the muscle activation level a_{feedback} [28], [32].

D. Stability Criteria

In the equation of motion, the net muscular torque can be substituted in terms of corresponding muscle forces and their moment arms

$$J_1 \dot{W} + WWJ_1W + G(\Theta) = -B^T \frac{\partial L}{\partial \Theta} F \quad (4)$$

where B is the transformation matrix between the body coordinate system and that of the inertial coordinate system, $(\partial L / \partial \Theta)^T$ is the matrix of moment arm of muscles, and F is the vector of muscle forces. Assuming static equilibrium condition about an equilibrium point Θ_e and following linearization of the equation of motion around the equilibrium point

$$\begin{aligned} J_1 \dot{W}|_e + \left(\frac{\partial G(\Theta)}{\partial \Theta} \Big|_e + B^T \frac{\partial L}{\partial \Theta} \Big|_e \frac{\partial F}{\partial L} \Big|_e \frac{\partial L}{\partial \Theta} \Big|_e \right) (\Theta - \Theta_e) \\ + \left(B^T \frac{\partial L}{\partial \Theta} \Big|_e \frac{\partial F}{\partial L} \Big|_e \frac{\partial L}{\partial \Theta} \Big|_e B \right) W = 0. \end{aligned} \quad (5)$$

In the compact form, the linearized equation of motion can be rewritten as

$$J_1 \dot{W}|_e + K(\Theta - \Theta_e) + VW = 0 \quad (6)$$

where $J_1 \dot{W}|_e$ is the inertial torque about the equilibrium point, K is the stiffness matrix that arises from conservative forces (gravitational and muscle stiffness), and V is the viscosity matrix that arises from nonconservative forces due to muscle viscosity. In state space form, the (7) can be written as

$$\begin{bmatrix} \dot{\Theta} \\ \dot{W} \end{bmatrix}_{6 \times 1} = A_{6 \times 6} \begin{bmatrix} \Theta - \Theta_e \\ W \end{bmatrix}_{6 \times 1} \quad (7)$$

in which

$$A = \begin{bmatrix} \text{zeros}_{3 \times 3} & B_{3 \times 3} \\ -J_1^{-1} K_{3 \times 3} & -J_1^{-1} V_{3 \times 3} \end{bmatrix}_{6 \times 6}. \quad (8)$$

To ensure stability of the linearized system, eigen-values of the system state matrix A should have negative real parts [20], [23], [33]

$$\text{REAL}(\text{Eig}(A)) < 0. \quad (9)$$

E. Computation Algorithm

The ‘‘geometry’’ process in the solution algorithm (see Fig. 1) involved the computation of instantaneous muscle lengths (based on the angular position and the origin-insertion coordinates of muscles in the upright posture [34]), muscle velocities (based on the derivative of muscle lengths with respect to time) and the moment arm of muscles (based on the derivative of muscle lengths with respect to each degree-of-freedom). The program ran in MATLAB 12 and Optimization Toolbox was used for the optimization process.

Feed-Forward Process: The first stage involved a feed-forward process, where the static optimization was used to derive a set of neural excitation of muscles under steady-state conditions. Based on the prescribed kinematics, the needed joint torque, which should be provided by trunk muscles, was computed using inverse dynamics. The performance criterion (P) was taken as the sum of the squared muscle activations [9], [10], [35]. The optimization problem was set by the following equations:

$$\text{Min } P = \sum_{m=1}^{48} (a_m)^2$$

subject to

$$\begin{aligned} J_1 \dot{W} + WWJ_1W + G(\Theta) &= -B^T \frac{\partial L}{\partial \Theta} F \\ \left. \begin{aligned} f_{\max, m} &= \sigma_{\max} \times \text{PCSA}_m \\ 0 \leq a_m &\leq 1 \end{aligned} \right\}; \quad m = 1 \text{ to } 48 \\ \text{REAL}(\text{Eig}(A)) &< 0. \end{aligned} \quad (10)$$

Feedback Process: The second stage, as depicted in Fig. 1, involved a feedback process in which kinematics parameters were computed through forward dynamics subject to neural excitation of muscles. In order to evaluate the performance of spindles in providing the reflexive stiffness, a perturbation moment was added to the existing moment. Having the net muscular moment, resulting kinematics parameters were obtained by numerical integration (depicted as forward dynamics in Fig. 1). It has been shown that the equations of motion are ill-posed [36]. Spindles were responsible to compensate any deviation of the predicted kinematics and the desired one which could be due to neural noise, numerical errors, or the perturbation. Due to the neural transmission delay, a delay of 20 ms was applied in the transmission of feedback signals which lies within the range of reported spinal muscle reflex latencies [37], [38].

F. Simulated Cases

We used the developed algorithm to simulate a point to point flexion movement from the upright posture to the flexion angle of 60° during 1 s, thereby producing a symmetric muscular activity. According to the experimental studies, voluntary point to point trunk movement is performed in two phases: the acceleration phase followed by the deceleration phase [39], [40]. The acceleration profile of the point to point trunk movement was approximated by a sinusoidal function. Velocity and position profiles were computed by numerical integration of the acceleration profile [Fig. 2(a)]. These kinematics profiles were treated as the desired kinematics profiles of the trunk movement which were to be produced via the neuromuscular elements. The net joint moment profile was subsequently computed by solving the dynamic equation of motion [Fig. 2(b)] using the feed-forward process.

Table I summarizes the simulations performed in this study. In the first simulation (Case 1, Table I), the performance of the feed-forward process was evaluated both with and without the stability constraint while the perturbation and spindle feedback were absent. In the second simulation, the performance of the feedback process was evaluated with the application of a 30 Nm perturbation flexion moment at the time 0.15 s for the duration of 0.07 s without inclusion of the stability constraint in the optimization (Case 2). In the third simulation, to examine how the feed-forward and the feedback mechanisms interact to provide stability, the same perturbation moment was applied when the stability constraint was considered in the optimization through the feed-forward process (Case 3).

III. RESULTS

The activation profiles of the main spine flexors and extensors in the simulation Case 1 are shown in Fig. 3. As expected, for

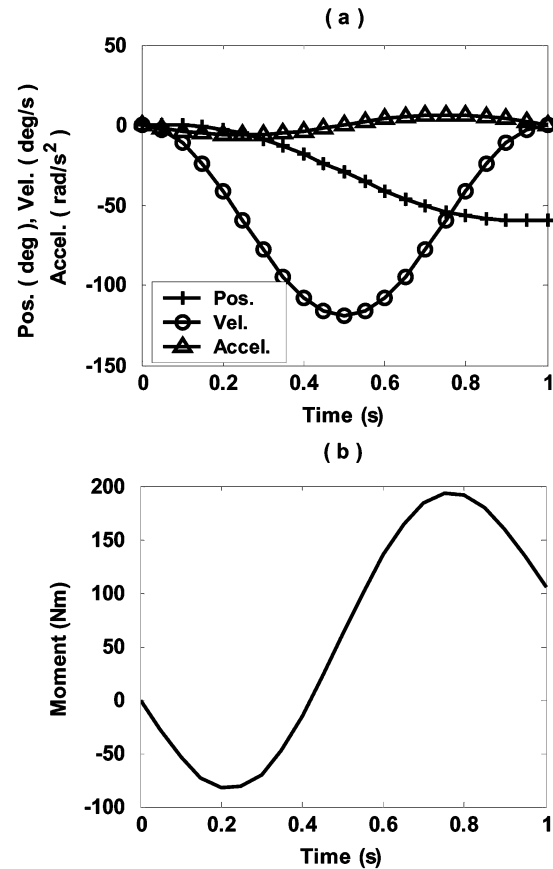


Fig. 2. Desired position, velocity, and acceleration profiles of the movement (a) and the net joint moment (b) considered for all simulation cases.

TABLE I
THREE SIMULATION CASES FOR TRUNK MOVEMENT FROM NEUTRAL 0° TO 60° FLEXION IN PRESENCE (+) OR ABSENCE (-) OF PERTURBATION, STABILITY CONSTRAINTS, AND SPINDLE FEEDBACK

	Case 1	Case 2	Case 3
Perturbation	-	+	+
Stability Constraint	+/-	-	+
Spindle Feed-back	-	+/-	+

the trunk flexion from the upright position, trunk flexors were initially activated to initiate trunk flexion movement. The contribution of RA to movement acceleration was noticeable which is consistent with the results of the literature [40]. Other flexors such as EO and IO were also activated in the first phase of the movement except for one portion of the IO which is regarded as an extensor due to its moment arm. The activity of flexors was ceased during the deceleration phase of the movement. On the other hand, extensor muscles were activated in a braking manner to terminate movement at the desired final point. The LT and IL were the most active muscles in this phase. Other back muscles such as LD muscles were also considerably activated. Other trunk muscles such as PL and PS were not included in the graphs as their activation level was relatively low (with maximum average value of 0.15% and 0.06%, respectively). Application of the stability criterion resulted in an increase in antagonistic activities of IL and LT during the first phase of the movement

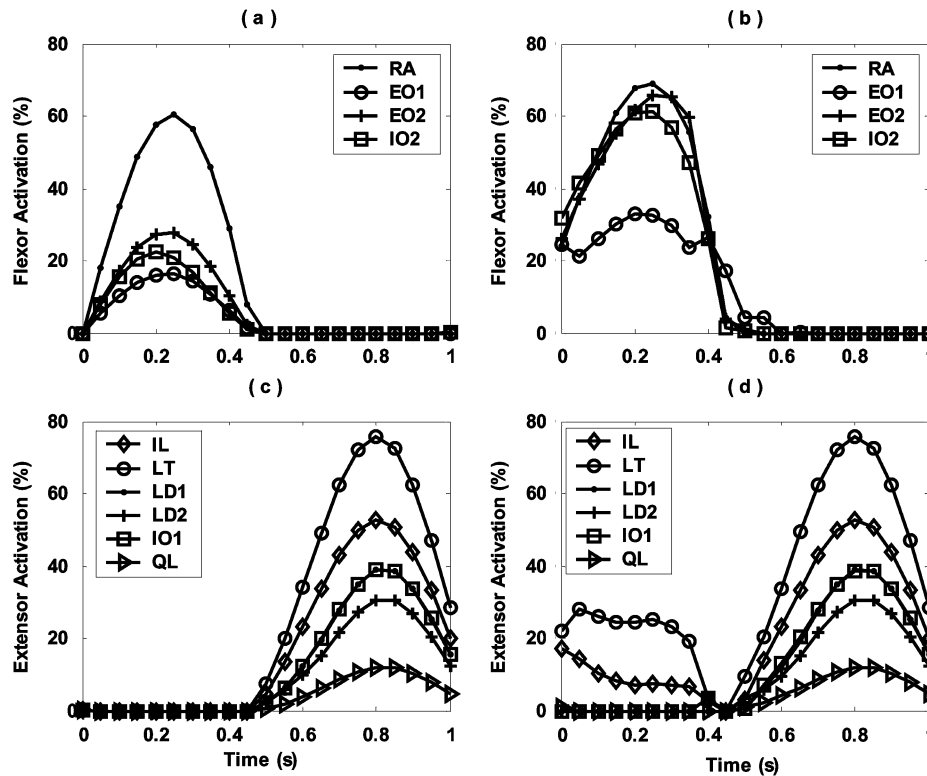


Fig. 3. Simulation Case 1: muscle activation profiles without [(a) and (c)] and with [(b) and (d)] application of the stability constraint without the effects of spindle feedback. Top panels show flexor muscles and the bottom panels illustrate the extensor muscles.

[Fig. 3(d)]. As a result, activation of the flexors was also increased to keep the net moment at the desired value [Fig. 3(b)]. Co-activation was not seen in the second phase of the movement indicating that there was no need for the additional stiffness during this phase of the flexion movement in agreement with earlier results that system stability substantially increases at larger flexion angles [5], [41]. Peak muscular force in the RA and EO was 320 N and 110 N, respectively, at the time 0.2 s. Peak force in the extensor muscles such as the LT and IL was 700 N and 350 N, respectively, at the time 0.8 s. Stability constraint caused increases of 200 N and 80 N, respectively, in the LT and IL muscle forces at the initiation point of the movement. In order to evaluate the relative passive/active muscle force contribution to the net joint moment without and with the stability constraint, total active and passive components of all muscle forces are illustrated in Fig. 4(a) and (b), respectively. Obviously, the passive muscle force was lower at the beginning of the movement and gradually increased reaching about 60% at the end of the movement [Fig. 4(c) and (d)]. Constraining the stability dramatically changed the relative contribution of active forces at the beginning of the movement (the active contribution shifted from 0 to 85%).

The effects of perturbation on the position and velocity profiles for the simulation Case 2 are depicted in Fig. 5. Perturbation caused the position and velocity profiles to deviate from the desired ones in absence or presence of the spindle response with different gain factors. When spindles were not included in the simulation (i.e., $g = 0$), the error in position and velocity reached to about 14° and $45^\circ/\text{s}$, respectively, at the end of the movement. However, incorporation of the spindle feedback into

the algorithm (i.e., $g = 300$) decreased these deviations to about 4° and $10^\circ/\text{s}$, respectively. Initially, an increase in the gain factor decreased the position and velocity errors with respect to the desired ones. But further increases in gain (> 450) yielded an overshoot in the position and velocity profiles and consequently increased the errors.

The effect of variation of the spindle gain factor on the feedback activation of muscles is shown in Fig. 6. The amplitude of feedback activation of extensor muscles (such as IL) increased with increasing feedback gain factor while their duration decreased [Fig. 6(a)]. With high gain factor even the flexor muscles (such as RA) were activated via reflexive activation [Fig. 6(b)].

Interestingly, incorporation of the stability constraint decreased the spindle responses to the perturbation [Fig. 7(b)]. Due to co-activation (inclusion of the stability constraint), the error of position and velocity profiles also decreased [Fig. 7(c) and (d)]. For example, at the final instance of the movement, the error in position and velocity decreased to about 3° and $7^\circ/\text{s}$, respectively.

Maximum joint compressive force reached 3000 N and 4000 N at the time 0.2 and 0.8 s, respectively. Maximum shear force was about 200 N at the time 0.3 s. Satisfying the stability condition significantly increased the compressive force especially at the beginning of the movement which was due to the increased antagonistic co-activation. McGill *et al.* [42] predicted the L4-L5 compressive force in terms of joint moments by introducing a simple polynomial. Comparison of our compressive force profile with the result of their polynomial revealed a correlation of 0.98 and 0.81 without and with the stability condition, respectively. The corresponding root mean

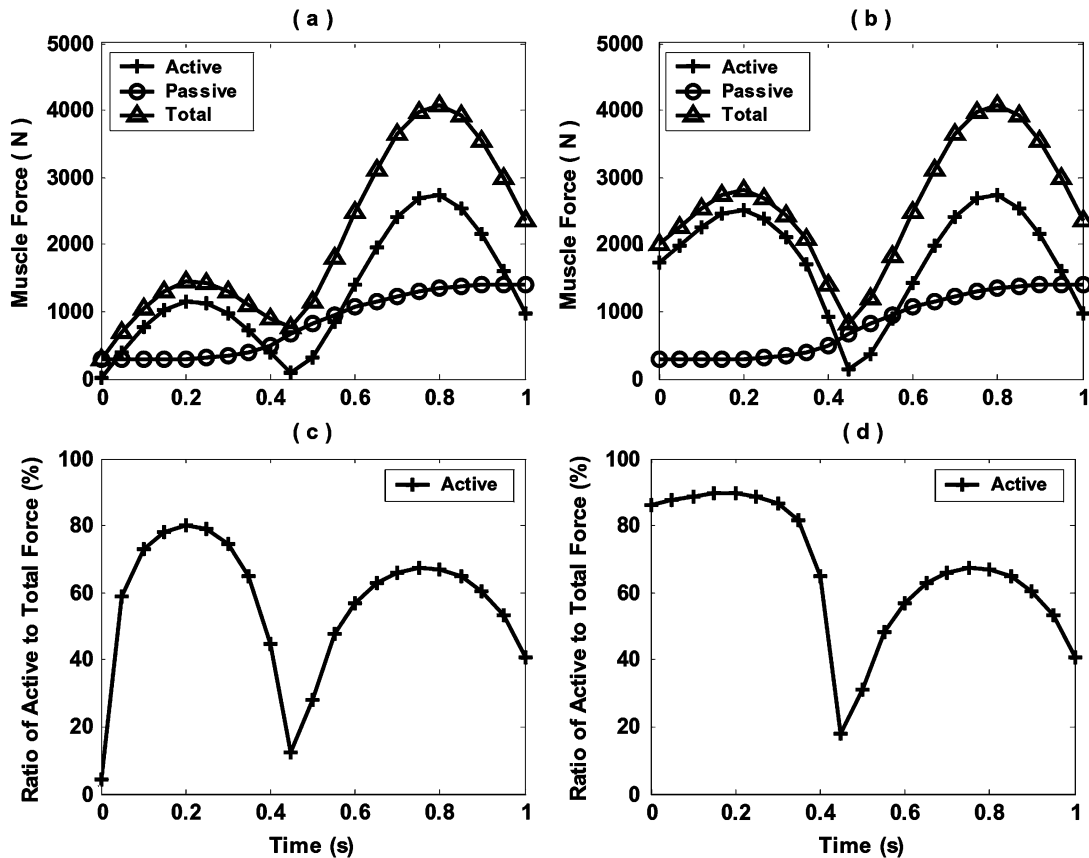


Fig. 4. Simulation Case 1: total muscle force profile (algebraic sum of all muscle forces) and its active and passive components without (a) and with (b) the stability constraint. Contribution of active force component as the percent of total muscle force without (c) and with (d) the stability constraint.

square error was 260 N and 820 N, respectively. Moreover, comparison of our computed compressive force with that of Fathallah *et al.* [43] yielded correlations of 0.84 and 0.63 without and with the stability condition, respectively.

IV. DISCUSSION

Such model studies could improve much-needed understanding of the pathomechanics of the spine that are essential in effective management of low back disorders. Preventive and performance enhancement programs also substantially benefit from the knowledge gained. The strength of the current rather simple computational approach is to combine the intrinsic and reflexive stabilizing subsystems in assessments of the spinal stability in 3-D dynamic tasks.

Most of the studies assessing the spinal stability have considered static [13], [17], [18], [44], [45] or quasi-dynamic [7], [46] models for the trunk musculoskeletal system. Cholewicki and McGill [7] developed a comprehensive electromyography (EMG)-assisted multirigid body model of lumbar spine with 18 rotational degrees-of-freedom. Nevertheless, their model was quasi-dynamic since movements were performed at low or constant velocity. The squat and stoop lifts simulated by the Kinematics-based approach were also performed at rather slow pace with generally insignificant effect of inertia [8], [41]. Fully-dynamic model of the trunk is required especially for the case of fast and sudden jerky movements.

In the present study, similar to the computational model developed by Thelen *et al.* [10], we used feedback in addition to

feed-forward controls to derive the kinematic trajectories toward a set of desired kinematics. In our study, the feedback process used the biologically identified reflex mechanism [27]–[29].

The role of co-activation to enhance spinal stability has long been recognized [11], [13], [18], [46], [47]. To compensate for the inability of the conventional optimization approaches in predicting antagonistic activities some studies have set nonzero lower bounds on muscle activations or stresses to evaluate the effect of co-activation on the spinal stability [13]–[15], [47]. This method, however, would not result in an acceptable co-activation level as it does not allow for any silence states of muscle activity [48] and, hence, would not yield the minimum required co-activation level based on the stability challenge [49].

We observed that the stability-based optimization is able to predict the minimum required co-activation of antagonist muscles. Increased antagonistic activities during initial acceleration phase of flexion from the upright position (reaching about 30% for LT and 20% for IL) imply that co-activation of muscles is needed during this period of movement, which is due to rather low trunk muscle activities. The passive stiffness of ligamentous and muscular systems in the neighborhood of the upright posture is relatively low, hence, higher stiffness must be provided by greater active muscular contribution [5], [18], [50]. Expectedly, the passive contribution should increase in the second phase of the movement (from 0.5 to 1 s) in agreement with the results of Arjmand and Shirazi-Adl [5] who noticed the important role of passive components in the stability of spine in 45° and 60° isometric flexion postures.

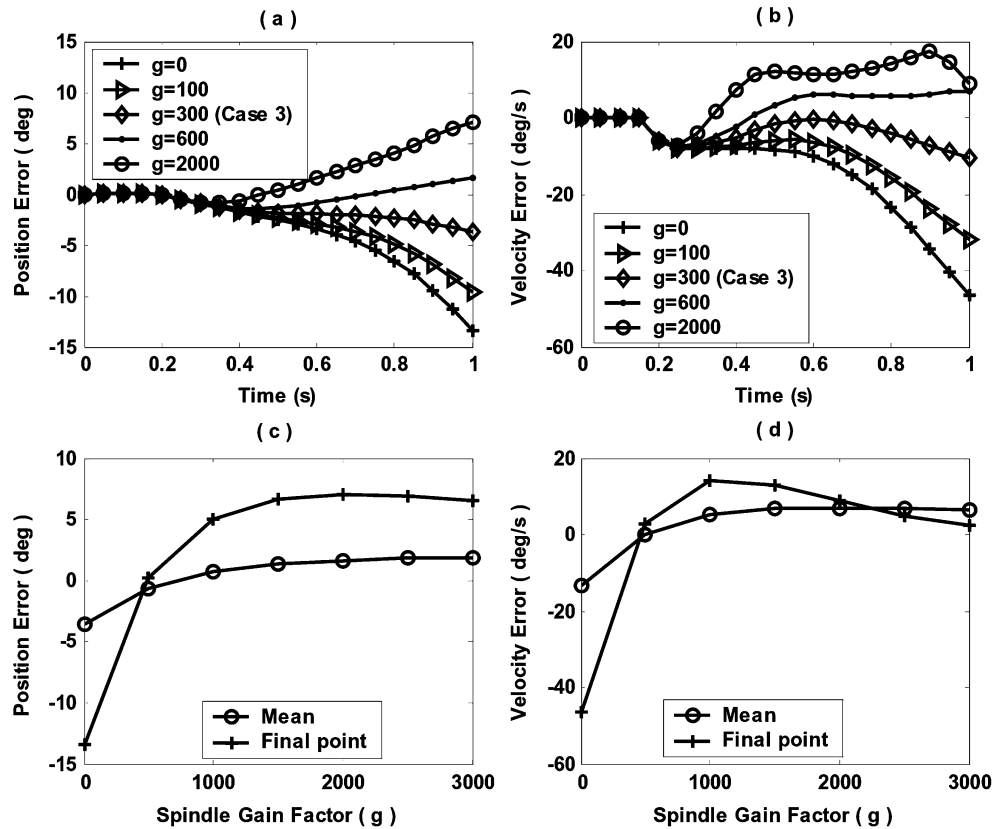


Fig. 5. Simulation Case 2: error in actual position (a) and velocity (b) profiles with respect to the desired profiles of the movement with different spindle gain factors in presence of the perturbation. When g is set equal to 0, no reflexive feedback is considered. Mean and final point errors in position (c) and velocity (d) with respect to the spindle gain factor g .

The stability requirement during dynamic activities is, therefore, phase-dependent. The higher threat of instability in the upright posture under low loads has been identified during isometric [5], [6], [47] and dynamic [8] conditions. Higher flexion angles or loads in hands tend to reduce the need for co-activation due to greater stiffness associated with higher muscle forces and passive bending moments. Consistently, Cholewicki and McGill [7] found that tasks which demand higher muscular activities show higher stability level than tasks which require less muscular activities.

In our study, feed-forward process, based on the stability requirement, contributed to the stability of the trunk movement by preactivating antagonistic trunk muscles. Brown and Potvin [17] also showed that inclusion of the stability constraint in the optimization simulation can provide more realistic prediction for antagonistic activity. However, they mentioned that adding the stability constraint into the optimization did not improve prediction of overall agonistic force levels which was due to the nature of the objective function used in their study. Although we cannot deduce that the CNS optimizes sum of the muscle activation while satisfying the stability requirement, the results are encouraging as a computational means to estimate dynamic loading on the spine.

To date, there have been few experiments investigating patterns of muscular activity during voluntary trunk flexion for quantitative validation of the results. In the study by Thorstensson *et al.* [39] volunteers were requested to perform

flexion movement with the maximum possible velocity as accurately as possible. Consistent with our study, they reported an initial burst of EMG activity from prime movers (flexors) during acceleration followed by an opposing muscle group (extensors) burst for deceleration. They also reported a final prime mover burst to ensure target position which was not observed in our simulations. Schmitz, in his study of iso-resistive trunk flexion movement [51], reported a similar two-phase muscular activity. He studied the flexion movement from upright to 55° flexion with varying resistance. Results of the current study for muscle activities are supported by his nominal resistance case study. His reported average of peak activities for flexor muscles is 8% more than that obtained in our study. This difference can be explained, since his study was not a freestyle movement as it was in our study. Due to nominal resistance, more activity of flexors with a shorter time to peak is anticipated in the acceleration phase of movement. For extensors such as erector spinae muscles he reported a peak activity similar to the average of peak activities of LT and IL muscles in our study but again with shorter time to peak. It should be reminded that acceleration profiles are not identical hence temporal difference in muscle activities should be expected.

General trend of compressive force profile was in fairly good agreement with the results of McGill *et al.* [42]. The difference was more pronounced when the stability condition was applied, maybe because the polynomial was basically derived from quasi-dynamic studies while the effect of stability is more

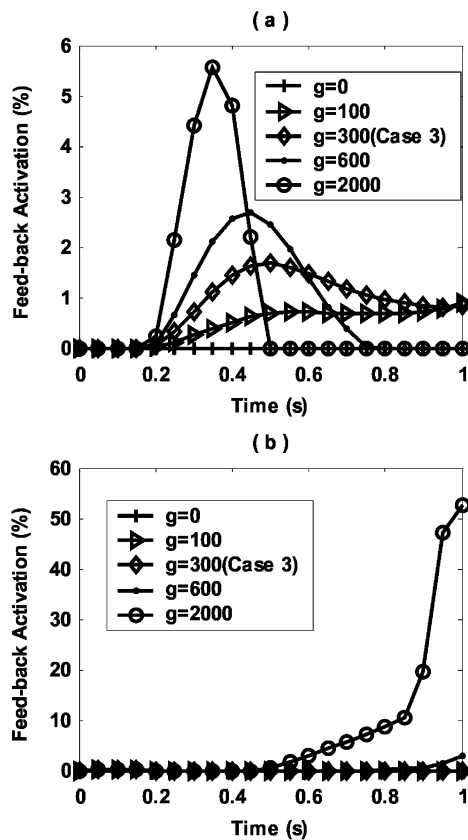


Fig. 6. Simulation Case 2: effect of different values of spindle gain factor on the feedback activation of one extensor muscle, IL, (a) and one flexor muscle RA (b). When g is set equal to 0, no reflexive feedback is considered.

significant in dynamic conditions. Large muscle and compressive forces are mainly due to the acceleration profile assumed in our simulation study.

The flexion perturbation moment considered in this study can simulate, for example, a sudden temporary additional weight. The impulse of the applied perturbation was within the range of that used by Granata *et al.* [37]. The range of force perturbations used in other recent studies generated moments almost similar to our perturbation moment [52], [53]. We observed that incorporation of the muscle spindle reflex activity into the simulation regulated the muscle activations and stabilized movement in response to the external perturbation. Since the perturbation was a flexion moment, extensor muscles were expected to be activated via reflexive mechanism. Actually, the reflex gain, g , is an input to our simulation and a wide range of values (i.e., 100–2000) has been tested for. Within this range, the selected value of 300 for the gain has shown the best performance. Increasing the feedback gain to higher values (i.e., 2000) yielded a nonphysiological increase in flexor muscles reflexive activity in order to make up for the errors of position and velocity profiles in overshooting the desired ones (Figs. 5 and 6). In other words, increasing gain to higher values caused unstable behavior of the system which was also affected by the reflex delay. Moreover, it was shown that the response of the reflex contribution was nonlinear since both amplitude and duration of feedback activation were affected by higher reflex gain. Due to the nonlinearity and latency in reflex response, fine tuning is needed to select an

appropriate reflexive gain factor in compensation for the perturbation effect. A range of 10–60 ms has been reported and also used for the latency of spinal muscle reflex activities [14], [23], [37], [38], [54]. Reducing the reflex delay has been shown to improve the performance of the reflex mechanism allowing for higher reflex gain prior to the system instability [54].

We also demonstrated that preactivation of the trunk muscles to higher degree through co-activation decreased the reflex response to the same perturbation. Granata *et al.* [37] have examined the trunk reflex response to different levels of force perturbation. They also found a declining trend for the response gain with the flexion preload which induced enhanced muscular activity prior to the perturbation. Similar results were reported by Stokes *et al.* [55], [56] who speculated that minimal adjustment by muscle reflex response is needed when the stability level is sufficiently high prior to the perturbation. As suggested by Panjabi [57] and [58], the spine stabilizing system could rely more on preactivating spinal muscles when sensory feedback from mechanoreceptors is corrupted through injuries to soft tissues. On the other hand, it has been suggested that muscle reflex response to perturbation can augment the lumbar spine stability level achieved prior to the perturbation so that it can compensate for the insufficient initial stability [22]. Alternatively, if we include reflexes within optimization, we would also predict the lower co-activation and minimum gain similar to Franklin and Granata [54]. Both the trunk stability level prior to the perturbation and the trunk reflex response after the perturbation, therefore, combine to influence the trunk stiffness and its kinematics response to perturbations especially in dynamic tasks [59].

The present model has some limitations because of the assumptions and simplifications in the model. In the current study, trunk was considered as a rigid inverted pendulum ignoring the intervertebral motions. This compelled us to consider only muscle fascicles that can produce moment about L5-S1 joint. The contribution of passive tissues like intervertebral discs and ligaments to the spine stiffness was overlooked in this study. Nevertheless, like passive muscle forces, their resistive contribution becomes more significant in larger flexion angles while active mediated mechanisms were identified during initial phase of the flexion. The magnitude of compressive and shear forces may be affected by the simplicity of the spinal model. We expect lower muscle forces at greater flexion angles if we include passive contributions of discs and ligaments attached to individual lumbar vertebrae. Future studies are underway to test Panjabi's recent hypothesis [58] by inducing partial passive tissue injuries and looking for compensation of neuromuscular response to adapt the potential impairment. The Transverse Abdominis was not included in our model as it does not produce any moment about lumbosacral joint. But it has been identified to have the potential of stabilizing the lumbar spine via intraabdominal pressure [60], [61]. Exclusion of pelvic rotation and the spine–pelvic coordination affects not only the amount of spine rotation for a single trunk orientation [62] but also the orientation of muscles and their moment arms. Neglected base excitations at pelvis in fast movements could also affect the results. We have a single concentrated mass while others more properly modeled the mass distribution and

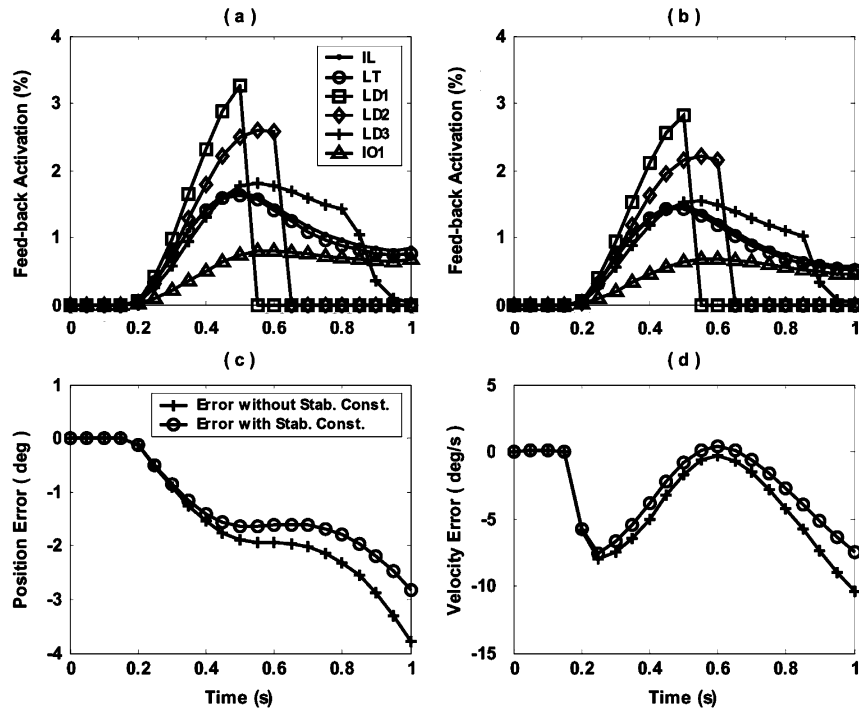


Fig. 7. Simulation Case 3: feedback activation provided by spindle action in response to the perturbation when the stability condition is (a) and is not (b) applied. Error in position (c) and velocity (d) profiles of the movement with respect to the desired profiles when the stability constraint is and is not considered in the optimization algorithm. The spindle gain factor g is set equal to 300.

its anterior location to individual motion segments center of rotation [5], [6], [8]. In the current study, lines of action of global muscles such as the erector spinae are assumed to be linear in all postures, whereas their moment arm is affected by its curved path in forward flexion postures influencing muscle and compressive forces as well as the spinal stability [63].

In this investigation, like any other simulation studies, the availability of experimental data, required structural complexity in the model and the nature of enquiry directly affects the selection of muscle model. We acknowledge that we have sacrificed two important features of muscle contraction: 1) the excitation–activation contraction dynamics and 2) the dissociation of the contractile element and musculotendon lengths. These features require complex models with ordinary or partial differential system of equations [32], [64]–[66]. Kaufman *et al.* using similar cost function and muscle model structure investigated knee muscle forces during isokinetic exertions [35]. Unlike animal studies such as Lieber *et al.*, measurement of sarcomere, tendon, aponeurosis lengths under physiological conditions is not possible [67]. Hence, we opted for simplicity of the muscle model due to complexity of the overall system (Fig. 1).

The greatest concern is when $(\partial F/\partial L)$ and $(\partial F/\partial \dot{L})$ from the muscle model is used to determine stability of the system. Many investigators have used a simple equation relating muscle stiffness to its force [68] (i.e., $K = qF/L$, where q is taken as a constant ranging from 0.2 to 40 with an average of 10). Following Bergmark [69], numerous researchers have used this relationship for their analytical studies of spinal stability [5], [17], [54], [70].

A parametric study employing this relation for muscle stiffness with q varying from 2 to 10 while repeating the simulation Case 1 confirmed our initial results (Fig. 8). For q larger than 8, the intrinsic stiffness of muscles were sufficient to ensure stability since the maximum eigen-values were always negative during the entire flexion cycle (Table II). At lower values of q , the positive eigen-values became more prevalent and co-activation was needed to stabilize the spine (Fig. 8). Spine was most vulnerable to instability close to the upright posture, which is consistent with both earlier studies [5], [18], [50] and physical intuition provided by the inspection of the governing equations of motion. Hence, the results indicate a qualitative agreement for two methods of assessing stiffness of muscles in this study. One cautionary note is warranted, Cholewicki and McGill [71] stated that simple linear relation between stiffness and force is only valid for isometric conditions.

Muscle contraction is so complex and analytical methods are so limited that any single model can capture only a part of the complexity. As more accurate relationship between muscle force, stiffness and viscosity becomes available general formulation above could be used with higher confidence. Franklin and Granata [54] have suggested one such relation (3) for isometric conditions. The isometric perturbation-based paradigm needs to be extended to dynamic conditions to better characterize the interplay of the intrinsic and reflexive stiffness and viscosity with muscle length and velocity that determine the overall stability of spine.

In conclusion, the present computational model combined feed-forward (open loop) and feedback (closed loop) control of the trunk stability for simulation of the trunk dynamics during

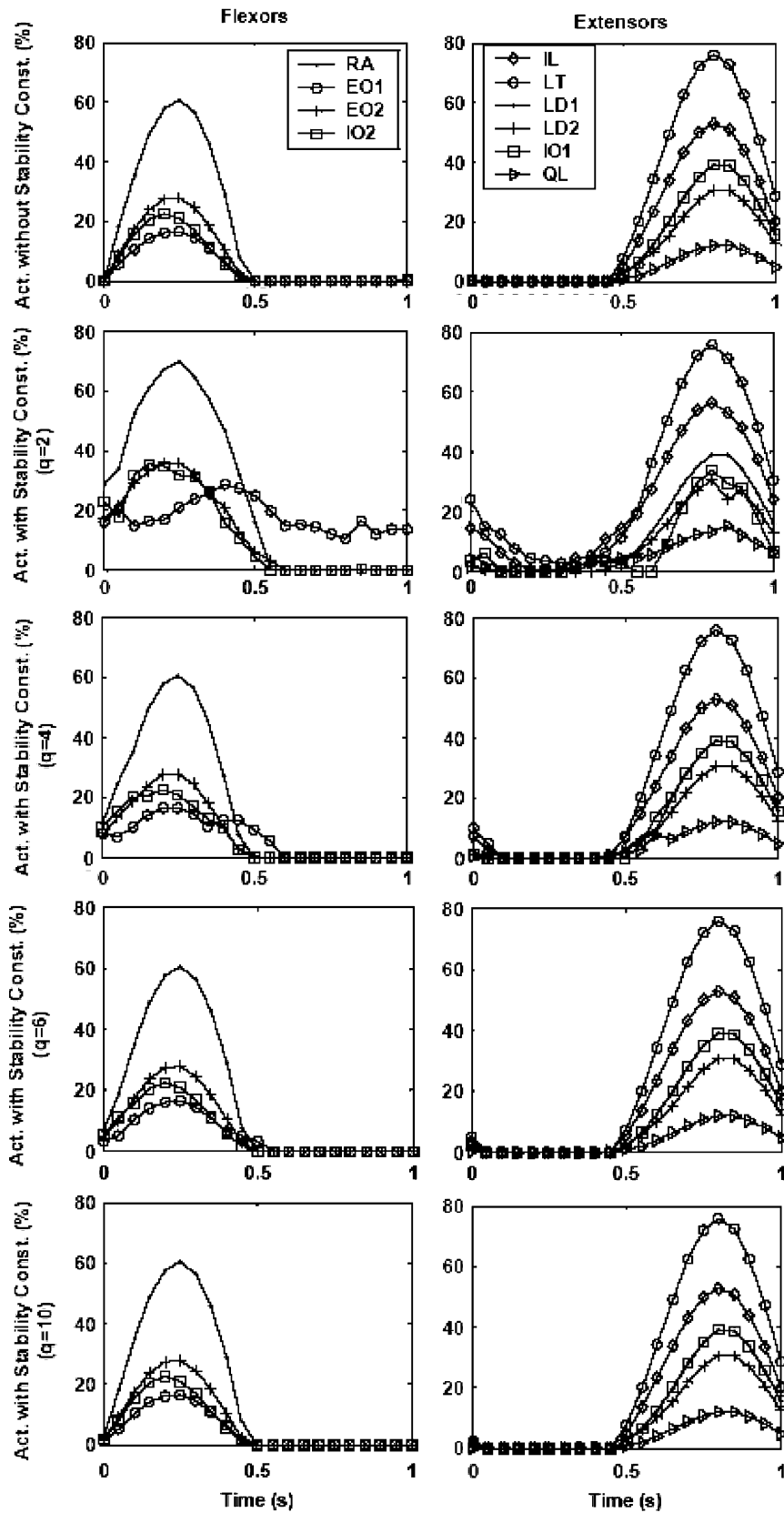


Fig. 8. Activation of flexors and extensors without or with stability constraint with different values of q .

3-D fully-dynamic tasks. A unique feature of this study is the ability to simulate the trunk dynamics and its stability during combined movements. In spite of shortcomings in the model,

it can be considered as a positive step toward more accurate evaluation of the trunk stability and its neuromuscular control during fully-dynamic tasks.

TABLE II
MAXIMUM EIGEN-VALUES OF THE SYSTEM STATE MATRIX AT
DIFFERENT PERCENTS OF THE FLEXION CYCLE
EVALUATED WITH DIFFERENT VALUES OF Q

q	Percent of Flexion movement						
	5	15	25	35	65	75	85
2	21.33	7.02	4.13	10.47	3.28	1.93	1.65
4	14.75	-0.43	-0.46	0.12	-0.29	-0.47	-0.53
5	10.92	-0.43	-0.47	-0.36	-0.26	-0.45	-0.52
6	6.52	-0.43	-0.47	-0.27	-0.25	-0.45	-0.51
7	1.17	-0.43	-0.48	-0.29	-0.24	-0.45	-0.51
8	-0.16	-0.43	-0.48	-0.29	-0.24	-0.45	-0.51
10	-0.16	-0.43	-0.48	-0.29	-0.24	-0.45	-0.51

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