Dynamic iso-resistive trunk extension simulation: Contributions of the intrinsic and reflexive mechanisms to spinal stability

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Abstract. The effects of external resistance on the recruitment of trunk muscles and the role of intrinsic and reflexive mechanisms to ensure the spinal stability are significant issues in spinal biomechanics. A computational model of spine under the control of 48 anatomically oriented muscle actions was used to simulate iso-resistive trunk movements. Neural excitation of muscles was attained 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. In this study, the trunk extension movement at various resistance levels while extending from 60° flexion to the upright posture was investigated. Incorporation of the stability condition as an additional constraint in the optimization algorithm increased antagonistic activities for all resistance levels demonstrating that the co-activation caused an increase in the intrinsic stiffness of the spine and its stability in a feed-forward manner. During the acceleration phase of the movement, extensors activity increased while flexors activity decreased in response to the higher resistance. The co-activation ratio noticed in the braking phase of the movement increased with higher resistance. In presence of a 30 Nm flexion perturbation moment, reflexive feed-back noticeably decreased the induced deviation of the velocity and position profiles from the desired ones at all resistance levels. The stability-generated co-activation decreased the reflexive response of muscle spindles to the perturbation demonstrating that both intrinsic and reflexive mechanisms contribute to the trunk stability. The rise in muscle co-activation can ameliorate the corruption of afferent neural sensory system at the expense of higher loading of the spine.

Keywords: Dynamic simulation, stability, spine, iso-resistive extension, spindle reflex

1. Introduction

Dynamic trunk motion has recently been identified as a potential risk factor for low back disorders [15], particularly when the motion occurs in multiple planes simultaneously [20,21]. The imposed trunk moments have been observed to increase by 15–170% during dynamic exertions. Greater trunk

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moments occur along with increased agonistic and antagonistic muscle activities resulting in greater spinal loads [15]. On the contrary, there have been few studies which reported lower trunk moments in response to higher trunk motion velocities [28]. Investigation of the dynamic trunk extension movement against external resistance is helpful to understand how the neuromuscular system produces and stabilizes spinal movements during dynamic lifts. Commercial rehabilitation devices providing resistance to trunk exertion (i.e. passive and active dynamometers) are used in both trunk muscle training and quantification of trunk performance [43,56]. Recently, Shirazi-Adl and colleagues [3,17] developed a kinematics-based finite element model of the trunk to evaluate the contribution of active and passive tissues to the spine loading and 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 the spine during various quasi-dynamic activities has previously been investigated in an EMG-assisted model study by taking the determinant of the Hessian matrix as the index of stability [10].

The contribution of neuromuscular elements during dynamic activities has been estimated by static optimization and inverse dynamics approach which are computationally efficient, especially in presence of redundancy with multiple muscles [2,66]. However, traditional optimization models of the trunk poorly predict antagonistic muscle coactivity essential to enhance stability of the trunk during dynamic tasks [27,49]. In addition, few studies have quantitatively examined the association between dynamic spinal stability and antagonistic co-activation. The effect of muscles co-activation on the lumbar spine stability has been investigated by increasing the lower bond on antagonistic muscle activations [1,24, 39]. However, this method cannot directly evaluate the minimum co-activation level needed to meet the stability requirements. The direct incorporation of the stability criterion, as an additional constraint in optimization models, seems to improve the prediction of antagonistic activity as well as the spine loading [8,31,55,63].

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

1) *Intrinsic mechanism*, which provides appropriate stiffness to the trunk by preactivating antagonistic muscles in anticipation of a 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 a perturbation [14]. Optimization models based on stability condition are supposed to provide the intrinsic stiffness to the trunk and partially contribute to its stability in advance of any disturbance.

2) *Reflexive mechanism*, which stiffens the trunk and contributes to its stability with proper feed-back in response to the perturbation. It has been shown that intrinsic stiffness alone is insufficient to stabilize trunk extension exertions in presence of disturbances [50]. Muscle spindles via "stretch reflex" are responsible for providing muscle's reflexive responses that change the muscle activation and hence its stiffness and viscosity [11].

The goal of this study was to use an efficient three-dimensional computational model for simulation of trunk dynamic extension movements and its stability when opposed by a constant resistance using both intrinsic mechanism provided by the stability-based optimization and reflex mechanism provided by the muscle spindle's action. The hypothesis that co-activation can reduce the need for reflex contributions was also investigated.
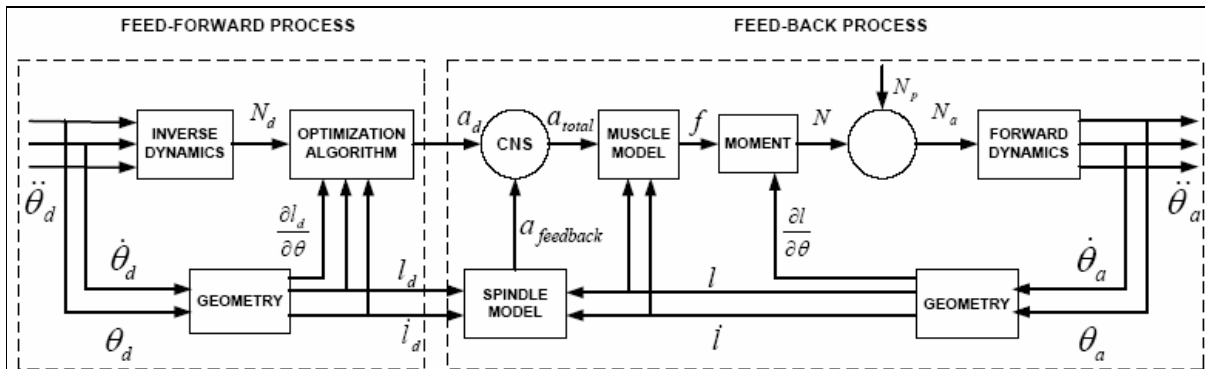
Fig. 1. Computation algorithm. Feed-forward Process: Muscle neural activations are computed based on the desired kinematics with or without application of the stability constraint in the optimization routine. Feed-back Process: The effects of muscle spindles on total muscle activations and the kinematics 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, 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 moment. A delay of 50 ms is applied in the transmission of feed-back signals.

2. Methods

2.1. Neuromuscular 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 three-dimensional space under the control of 48 muscle fascicles. The effect of gravity was considered as a concentrated load (300 N) at the center of mass with a distance of 40 cm from the pendulum center of rotation [1,34]. The dynamic equations of motion of this inverted pendulum were derived in a compact form [1,34,55]. The muscle and spindle models used in our study are described in the Appendix A. Linearization of the equations of motion around an equilibrium point yielded the system state matrix which should have eigenvalues with negative real parts to ensure stability (Appendix A). The anatomical and geometric data of muscle fascicles were based on the literature [10]. Only those muscle fascicles that can create moment about the L5-S1 joint were included in this study (muscle abbreviations are described in the Appendix B).

2.2. Computation algorithm

The "Geometry" process in the simulation algorithm (Fig. 1) involved the computation of the instantaneous muscle lengths (based on the angular position and the origin-insertion coordinates of muscles in the up-right posture [41]), 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 was run in MATLAB®12 and Optimization Toolbox was used for the optimization process.

2.2.1. 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 generated by trunk muscles, was computed using inverse

dynamics. To simulate the iso-resistive trunk extension, additional constant levels of resistance torque was considered against the movement. The performance criterion (P) was taken as the sum of the squared muscle activations [2,42,66]. The optimization process was modeled by the following set of equations (see Appendix A for more descriptions):

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

Subject to:

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

2.2.2. Feed-back process

The second stage, as depicted in Fig. 1, involved a feed-back process in which kinematics parameters were computed through forward dynamics subject to neural excitation of muscles. In order to evaluate the performance of muscle spindles in providing the reflexive stiffness, a perturbation moment, N_p, was added to the existing moment, N (Fig. 1). Having the net muscular moment, resulting kinematics parameters were obtained by numerical integration (depicted as Forward Dynamics in Fig. 1). Spindles were responsible to compensate any deviation of the predicted kinematics and the desired one which could be due to numerical errors or the perturbation.

A neural transmission delay of 50 ms was considered in the transmission of feed-back signals which lies within the range of reported spinal muscle reflex delays [23,30,73].

2.3. Simulated cases

The sagittally-symmetric extension movement from $60°$ of flexion to the upright posture during 1 s with various resistance torques of 0 (no resistance), 30 Nm (8% MVC), 70 Nm (20% MVC) and 100 Nm (40% MVC) was considered. The maximum voluntary contraction (MVC) was simulated by requiring the model to maximize torque at the upright posture. 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 prescribed 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)).

Table 1 summarizes the simulations performed in this study. In the first simulation (Case 1), the performance of the feed-forward process was evaluated with different levels of resistance and both with and without the stability constraint while the perturbation and spindle feed-back were absent. In the second simulation (Case 2), the performance of the feed-back process was evaluated with the application of the 30 Nm flexion perturbation moment at 0.65 s for the duration of 0.07 s without inclusion of the stability constraint and resistance in the optimization. In the third simulation (Case 3), to illustrate how the feed-forward and the feed-back mechanisms interact to provide stability, the same perturbation moment was applied when the stability constraint was considered in the optimization through feed-forward process while no resistance was included. In Case 4, a combination of all conditions including the resistance and the stability constraint was considered while spindles were active with the same feed-back gain as in Cases 2 and 3 against the external perturbation.

Table 1
Four simulation cases for the trunk movement from 60° of flexion to the up-right posture in presence (+) or absence (−) of the resistance, perturbation, stability constraint and spindle feed-back

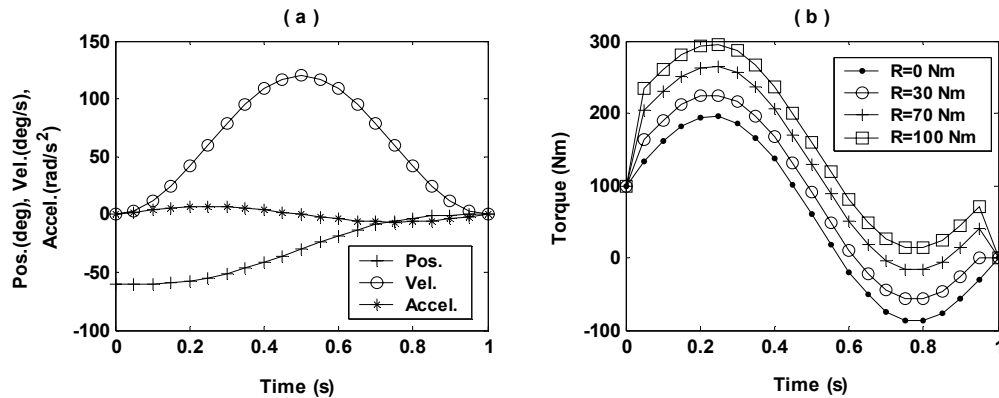|	Case 1	Case 2	Case 3	Case 4
Resistance	+	−	−	+
Stability Constraint	+/−	−	+	+
Spindle Feed-Back	−	+/−	+	+
Perturbation	−	+	+	+

Fig. 2. Prescribed kinematics profiles of the extension movement (a) and the net trunk muscular torque against four resistance levels (b).

3. Results

In simulation Case 1, as expected, activation of trunk extensor muscles increased initially to initiate the trunk extension movement (Fig. 3). The contribution of LT and IL to movement acceleration was significant with an increasing trend in activity (to saturation) and duration against higher levels of resistance. Other trunk extensors such as LD and PL were also activated in the acceleration phase of the movement (with the peak values of 30% and 25%, respectively, when no resistance was present) in addition to one portion of the Internal Oblique (IO1, with the peak value of 35%) which was recruited as an extensor due to its moment arm. During the deceleration phase of the movement, the activity of extensor muscles was ceased and, in turn, flexor muscles were activated in a braking manner to terminate movement at the desired final point (Fig. 4).

RA and EO1 muscles were the most active muscles in this phase with a decreasing trend in magnitude and duration of activity against higher resistances (Fig. 4(a) and 4(c)). Application of the stability criterion resulted in an increase in activities of extensor muscles (mainly IL and LT) during the second phase of the movement against all resistance levels (Fig. 3(b) and 3(d)). Flexor muscles (RA and EO1) also showed an increase in activity (to about 20% and 30%, respectively) in the acceleration phase at higher resistance levels (70 Nm and 100 Nm) which caused recruitment of the extensors with lower mechanical efficacy such as LD, PL and QL due to the saturation in IL and LT. However, at the highest resistance level (100 Nm) extensors seemed to be active in all movement duration (even in the deceleration phase).

The effects of resistance levels on the co-activity were investigated by defining an overall co-activation ratio considering the fact that the role of trunk flexors and extensors (as agonists or antagonists) changed in
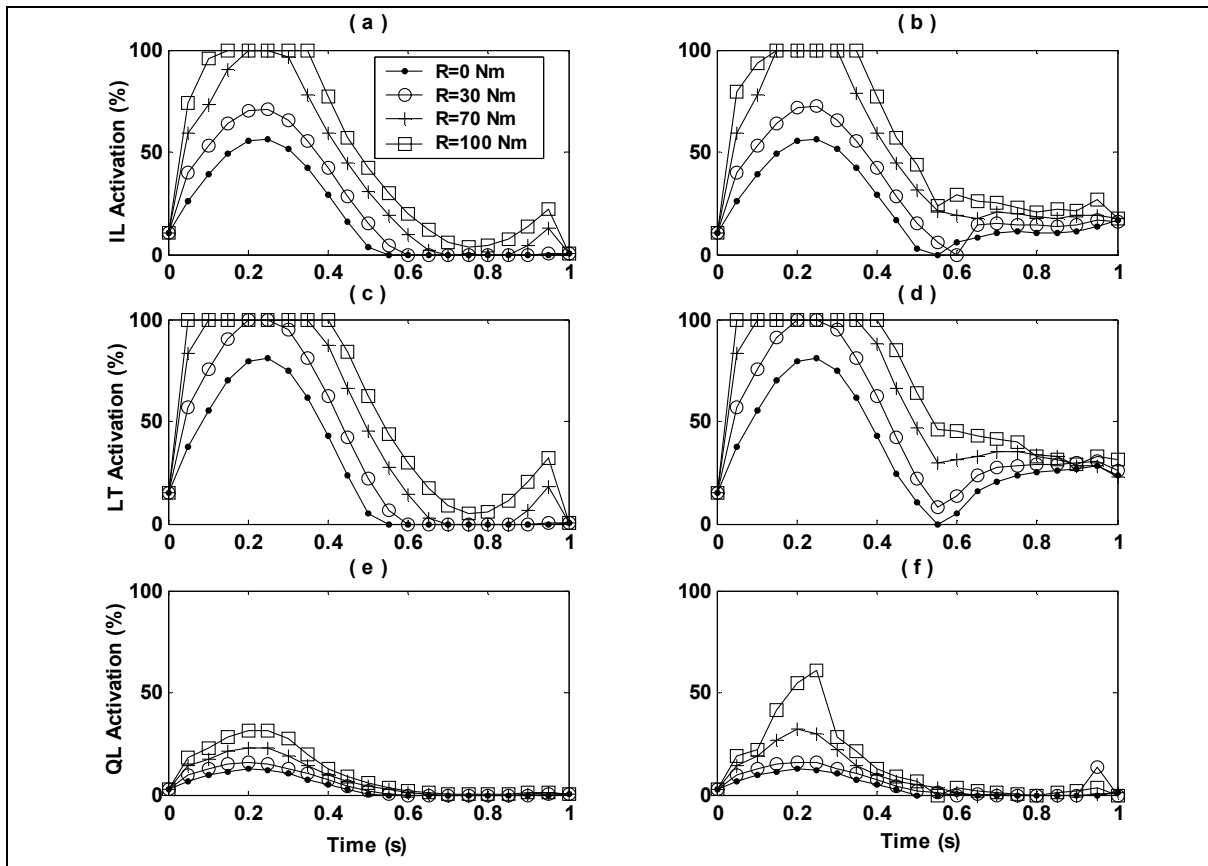
Fig. 3. Simulation Case 1: Trunk extensors (IL, LT, QL) activation profiles in extension movement against different resistance levels without the stability constraint [(a), (c) and (e)] and with the stability constraint [(b), (d) and (f)].

acceleration and deceleration phases. For instance, LT muscle in the acceleration phase of the movement was an agonist as was the case for RA muscle in the deceleration phase. The co-activation ratio was computed by the following equation:

$$Coactivation\ ratio = \frac{\sum_{j} a_j^{\mathrm{antagonist}}}{\sum_{i} a_i^{\mathrm{agonist}} + \sum_{j} a_j^{\mathrm{antagonist}}}$$

In which, a_i^{agonist} and $a_j^{\mathrm{antagonist}}$ are the agonistic and antagonistic activities, respectively. The co-activation ratio, shown in Fig. 5(a), was calculated during movement against different resistance levels accounting for the stability criterion. The arithmetic mean of co-activation ratio is also shown in Fig. 5(b).

Co-activation, through application of the stability constraint, was less pronounced in the first phase of the movement indicating that there was less need for additional stiffness during this initial phase. In the highest resistance case ($R = 100$ Nm), an increase in co-activity (to about 0.1) was also noticed during the first phase of the movement in addition to the braking phase implying the need for a modest level of co-activity in almost entire movement duration.

In the simulation Case 2, before the perturbation was applied, position and velocity profiles calculated
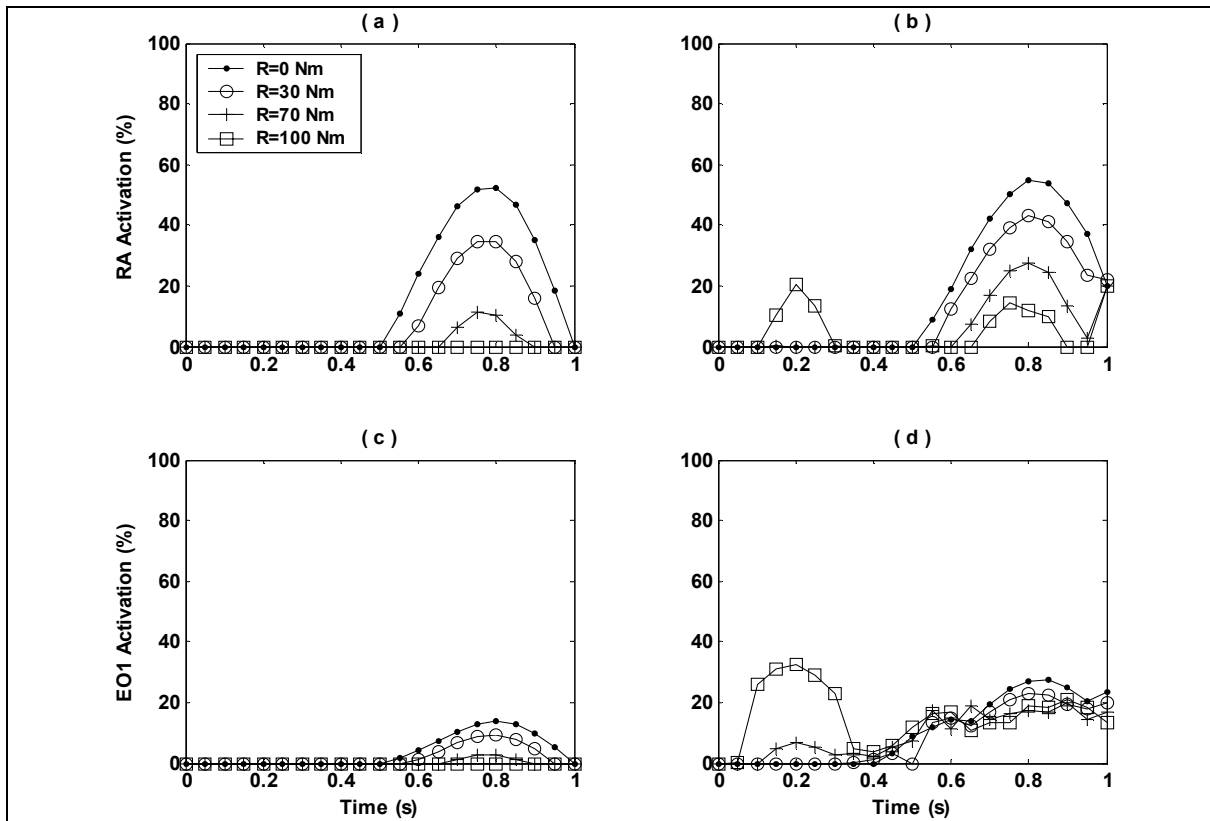
Fig. 4. Simulation Case 1: Trunk flexors (RA and EO2) activation profiles in extension movement against different resistance levels without the stability constraint [(a) and (c)] and with the stability constraint [(b) and (d)].
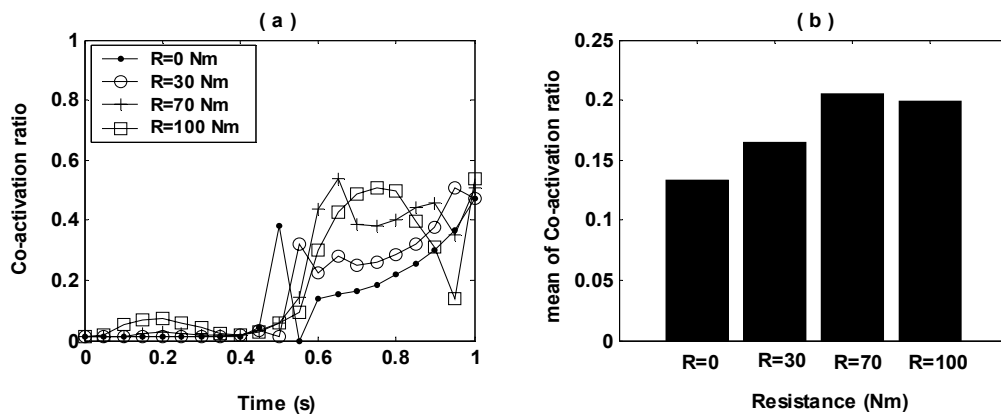

Fig. 5. Simulation Case 1: Co-activation ratio profiles (a) and its mean (b) during extension movement against different resistance levels with the stability constraint.

based on the obtained muscle activations showed negligible deviations of 0.05 ° and 0.2°/s, respectively, which was due only to the computational errors. It has been shown that the equations of motion are ill-posed [33]. The effects of perturbation on the position and velocity profiles for the simulation Case
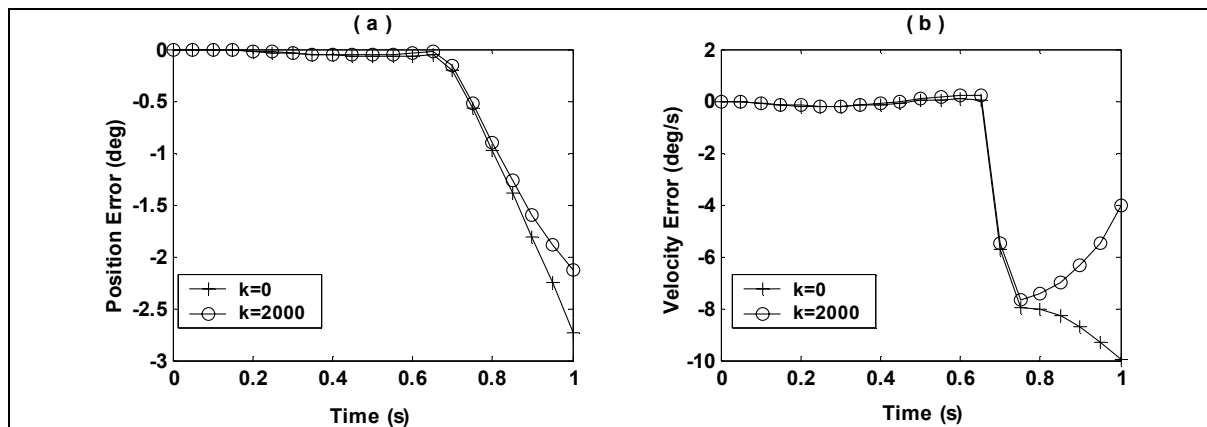
Fig. 6. Simulation Case 2: The effects of spindle feed-back on the error of position (a) and velocity (b) profiles with respect to the desired ones. The perturbation flexion moment of 30 Nm is applied at 0.65 s for the duration of 0.07 s.

2 are depicted in Fig. 6. Higher errors in position and velocity profiles were seen in response to the perturbation when spindles were not active (i.e., $k = 0$) in which case, the error in position and velocity reached to about $2.7°$ and $10°/s$, respectively, at the end of movement. However, incorporation of spindle feed-back into the algorithm (i.e., $k = 2000$), decreased these deviations to about $2.2°$ and $4°/s$, respectively.

Interestingly, incorporation of the stability constraint decreased the spindles response to the perturbation (Fig. 7). Due to co-activation (inclusion of the stability constraint), the errors in position and velocity profiles were also decreased (Fig. 7(c) and 7(d)). For example, at the final instant of the movement the error of position and velocity decreased to about $1.7°$ and $2°/s$, respectively.

For Case 4, similar to Case 3, the stability condition brought about a decrease in the position and velocity errors irrespective of the resistance levels (Fig. 8). Both resistance and the stability constraint increased the L5-S1 joint reaction forces, especially the compressive force, as depicted in Fig. 8(c) and 8(d).

4. Discussion

The main objective of this study was to employ a rather simple computational model for simulation of fully-dynamic trunk extension with variable resistance using a stability-based static optimization as a feed-forward process and a reflexive mechanism as a feed-back process. 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.

Most of the studies assessing the spinal stability have considered static [8,12,24,31,64] or quasi-dynamic [10,48] models for the trunk musculoskeletal system. The squat and stoop lifts simulated by the kinematics-based approach were also performed at rather slow pace with generally insignificant effect of inertia [6]. A fully-dynamic model of the trunk is required, especially for the case of fast and sudden jerky movements. Additionally, the role of time-dependent dynamic neural feed-back, as the third stabilizing subsystem [50], has been ignored in many earlier woks. In a recent study, nonlinear dynamic analysis has been used to characterize the neuromuscular control of stability during repetitive trunk movements [26].
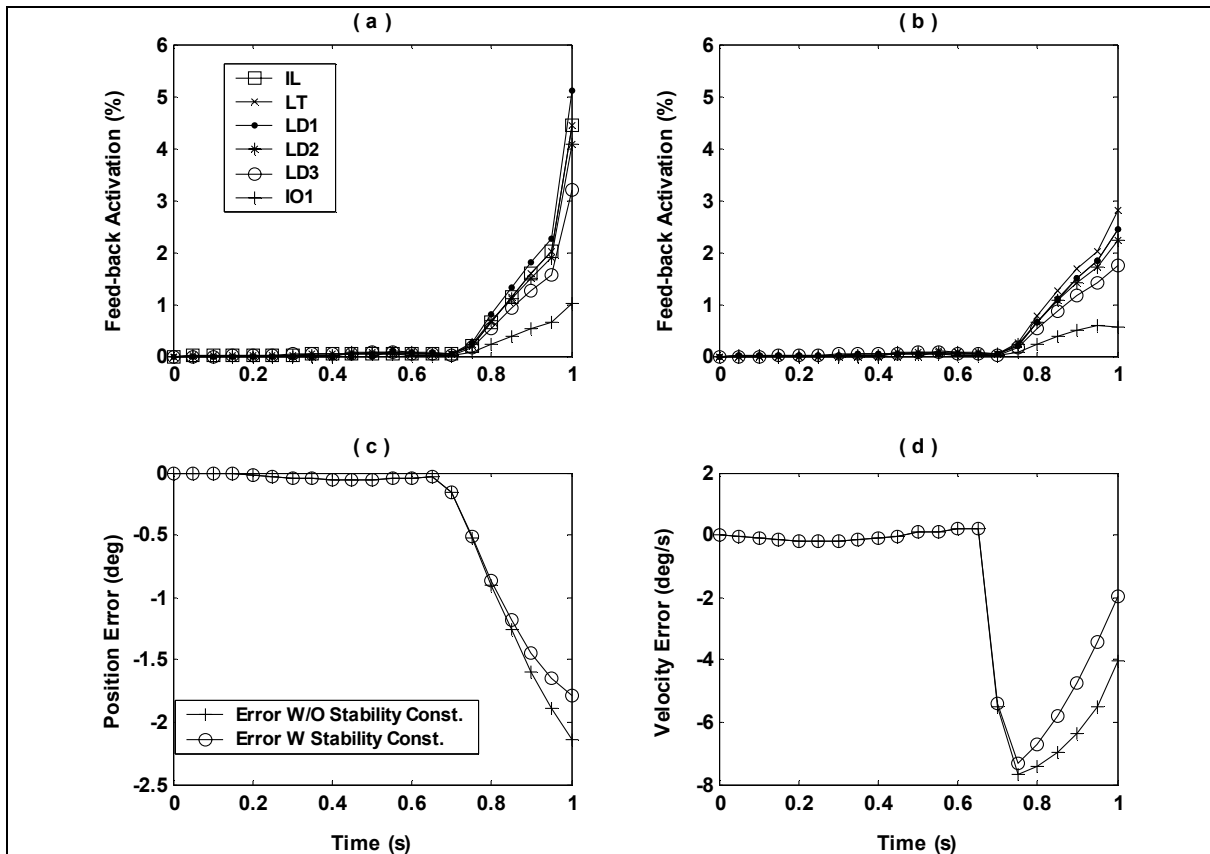
Fig. 7. Simulation Case 3: Spindles feed-back activation profiles of a number of extensor muscles without (a) and with (b) the stability constraint in response to the perturbation. The error of position (c) and velocity (d) profiles with respect to desired ones in presence of the perturbation and the spindle feed-back (k=2000) with (W) and without (W/O) the stability constraint.

Similar to the computational model developed by Thelen et al., we used feed-back in addition to feed-forward control to derive the kinematics trajectories toward a set of desired kinematics [66]. In our study, the feed-back process used the biologically identified reflex mechanism [25,32,38].

The role of co-activation to enhance the spinal stability has long been widely recognized [18,24, 27,31,48]. To compensate for the inability of the conventional optimization approaches in predicting antagonistic activities some studies have enforced non-zero lower bounds on muscle activations or stresses to evaluate the effect of co-activation on the spinal stability [1,18,24,39]. This method, however, would not necessarily result in an acceptable co-activation level as it does not allow for any silence states of muscle activity [22] and, hence, would not yield the minimum co-activation level required based on the stability challenge [29]. Brown and Potvin [8] showed that inclusion of the stability constraint in the optimization simulation can provide more realistic prediction for antagonistic activity. We also observed that the stability-based optimization in feed-forward process is able to predict the required co-activation of antagonist muscles. Increased antagonistic activities during the braking phase of extension (to about 30% for LT and 20% for IL) imply that co-activation of muscles is needed during this period of movement irrespective of the resistance level. The need for co-activation in the acceleration phase of the flexion movement from the up-right posture has also been reported [74]. Consistently, higher threat of instability in the upright posture under low loads has been identified during isometric [3,17,18] and
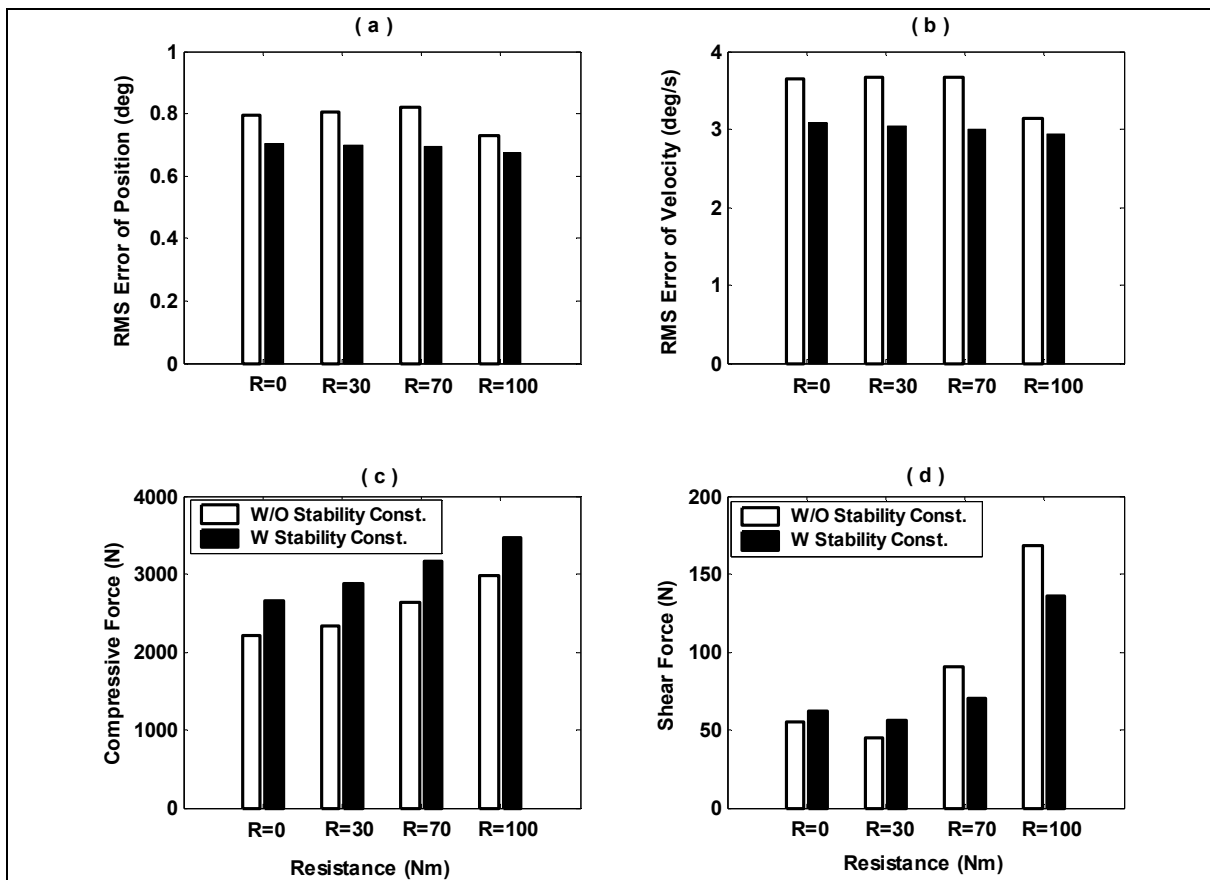
Fig. 8. Simulation Case 4: Combined effects of the stability constraint and resistance levels on the RMS errors of position (a) and velocity (b) profiles with respect to the desired ones in presence of the perturbation and spindle feed-back ($k = 2000$). The effects of the stability constraint and resistance levels on the mean compressive (c) and shear (d) forces at the L5-S1 joint.

dynamic [6] conditions. The 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 [3, 4,6]. The calculated mean co-activation ratio showed that the movement against higher resistance may require higher intrinsic stiffness for stability. Interestingly, we noticed another phase of co-activity in the acceleration phase of the movement when resistance exceeded 70 Nm.

The increase in back extensors activity against higher resistance was expected since they are the primary accelerator in the first phase of the extension movement, an observation in agreement with earlier works [59,60]. Decreased activity of abdominal muscles predicted under greater resistance levels is in contrast to results of Schmitz [60] who did not report any significant change in abdominal muscle activity with higher resistance.

To date, there have been few experiments investigating the coordination patterns of muscular activity during voluntary trunk dynamic extension particularly against external resistance. In the study by Thorstensson et al. [67] volunteers were requested to perform bidirectional flexion/extension movements with various velocities and movement amplitudes. Similarly, they reported a strict reciprocity between agonist and antagonist muscles. Marras and Mirka [46,47] also studied the muscle activities during the generation of trunk acceleration in extension or in combination with twisting. However their acceleration

levels were much lower than what we prescribed in our simulations. Ross et al. [59] and Schmitz [60], in their studies of iso-resistive trunk extension movement from 55° of flexion to the upright posture, reported a similar two-phase muscular activity. During nominal resistance case, peak acceleration and velocity seen by Ross et al. [59] was similar to that prescribed as the desired kinematics in our simulation. Similar activity levels were reported for abdominal muscles except for oblique muscles activity which was 20% higher than our result on average. They reported 20% lower activity in lateral extensors (LT and IL muscle group) and observed that the duration of extensor activities increased with higher resistance as was shown in our results [59]. Abdominal muscles, in that study, showed little variation with higher resistance while in our study the duration of their activity decreased with higher resistance. It should be noted that our input acceleration profile was different from that used in earlier studies. The assumed acceleration and resistance profiles caused the appearance of a short third phase of muscle activities against high resistance levels (Fig. 3(a) and 3(c)) which may not be experienced in in-vivo studies with dynamometers basically because of different kinematics profiles. Future simulations should allow trajectory planning to be realized based on the optimal control to further allow investigation of the effects of resistance on both kinematics and kinetics of spine [68].

Most of the existing biomechanical models of the spinal stability have overlooked the role of reflex despite the evidence on its contribution [50]. 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 in an earlier study [30]. The range of force perturbations used in other recent studies generated moments almost similar to our perturbation moment [44,51]. We observed that the incorporation of muscle spindle reflex activity into the simulation regulated the muscle activations and reduced the deviation of movement profiles in response to an external perturbation. It has been shown that the reflex gain, k, is a significant factor determining the efficacy of the reflex response due to the nonlinearity and latency in the reflexive system [74]. A range of 10 to 60 ms has been reported and also used for the latency of spinal muscle reflex activities [1,16,23,30,73]. Reducing the reflex delay has been shown to improve the performance of the reflex mechanism allowing for higher reflex gain to be reached before the system becomes unstable [23]. Abnormally large reflex delays have been associated with the impaired postural control observed in patients with low back pain [57].

We demonstrated that greater preactivation of trunk muscles through co-activation decreased the reflex response to the perturbation. Granata et al. [30] have examined the trunk reflex response to different levels of force perturbation. They also found a declining trend for the response gain with flexion preload which enhanced muscular activity prior to the perturbation. Similar results were reported by Stokes et al. [62,65] who speculated that minimal adjustment by muscle reflex response is needed when the stability level is sufficiently high prior to the perturbation. It has been suggested that co-contraction strategy may be used to compensate for impaired reflex dynamics and corruptions in sensory feed-back from mechanoreceptors [53,54,69]. On the other hand, muscle reflex response to the perturbation can augment the lumbar spine stability while compensating for insufficient intrinsic stability [11]. In fact it provides a more energy efficient control of stability than steady state recruitment of intrinsic stiffness alone [23]. All of above-mentioned results reinforce the notion that both the trunk stability level prior to the perturbation and the trunk reflex response after the perturbation combine to influence the trunk stiffness and its kinematics response to perturbations especially in dynamic tasks. However, we did not observe a significant effect of resistance on the kinematics errors with respect to the desired ones in presence of the same perturbation and spindle feed-back, even though higher spinal loads were observed while the stability constraint was imposed. As suggested by Reeves and Cholewicki [58] the next step toward lumbar spine modeling involves dynamic models of loading and movement, which include motor control responses. We believe the present approach is an initial step in the right direction.

The present model has some limitations; the availability of experimental data, required structural complexity in the model and the muscle model. Two important features of muscle contraction have been overlooked: 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 systems of equations [9,45,71,72].

Another concern is when $\partial F/\partial L$ as the muscle stiffness is used to determine the stability of the system. Additional parametric studies employing the well-known relation qF/L for muscle stiffness instead showed a qualitative consistency between the two approaches [74]. The model was most vulnerable to instability close to the upright posture, which is consistent with both earlier studies [3,31] and physical intuition provided by the inspection of the governing equations of motion.

In the current study, the intervertebral motions were ignored and the trunk was considered as a rigid inverted pendulum. This compelled us to consider only muscle fascicles that can produce moment about L5-S1 joint. However, exclusion of intersegmental muscles should not affect the overall stability of the spine [13]. Moreover, the contribution of passive tissues such as intervertebral discs and ligaments to the spine stiffness was overlooked in this study. Nevertheless, inclusion of passive tissues would not affect the results qualitatively since, like passive muscle forces, their resistive contribution becomes more significant in larger flexion angles while active mediated mechanisms were identified during the second phase of extension. We expect lower muscle forces at greater flexion angles if we include passive contributions of discs and ligaments and larger shear components by inclusion of local muscles attached to individual lumbar vertebrae. 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 intra-abdominal pressure [5,36,37]. Exclusion of pelvic rotation and the spine-pelvic coordination affects not only the amount of spine rotation for a single trunk orientation [19], but also the orientation of muscles and their moment arms. Neglected base excitations at pelvis in fast movements could also affect results. We have a single concentrated mass while others have modeled the mass distribution and its anterior location to individual motion segments center of rotation [3,6,17]. In the current study, lines of action of global muscles such as erector spinae were assumed to follow a straight line in all postures. Moment arms of global muscles are affected by their curved paths in forward flexion postures influencing muscle and compressive forces as well as the spinal stability [4]. These limitations are being re-examined in our current ongoing extension of this model.

5. Conclusion

The present computational model combined feed-forward and feed-back controls of the stability for simulation of dynamic trunk extension in presence of an external resistance with and without a perturbation. In spite of shortcomings in the model, it demonstrates a great potential toward more accurate evaluation of the trunk stability and its neuromuscular control during fully-dynamic tasks. Incorporation of the stability condition as an additional constraint in the optimization algorithm increased antagonistic activities. The co-activation ratio noticed in the braking phase of the movement increased with higher resistance. In presence of an external perturbation, reflexive feed-back noticeably decreased deviations in movement profiles. The intrinsic stability decreased the reflexive response of muscle spindles to the perturbation. The higher co-activation observed during the dynamic movements of low back pain patients may also be a compensatory manifestation of the larger delay in their reflexive response which stiffens spine exploiting intrinsic stability mechanism. The clinicians should consider the predicted compressive and shear forces at different levels of iso-resistive extension movements when designing

the optimal exercise programs (i.e. specifying the resistance levels) to avoid undue risk while taxing the neuromuscular resources for facilitating their rehabilitation course.

Appendix A

The dynamic equations of motion of the inverted pendulum were derived in a compact form:

$$J_1 \dot{W} = -WW J_1 W + N_{\text{input}} - G(\Theta) + N_R \tag{A1}$$

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 L5-S1 in body coordinate system. $WW J_1 W$ represents the torque due to coriolis forces. N_R is a constant resistance torque opposing the direction of motion defined as:

$$N_R = -R \cdot Sign(\dot{\theta}) \tag{A2}$$

where, R is the magnitude of resistance.

Muscle model

$$f = f_{\max} \cdot \left\{ a \cdot f(l) \cdot f(\dot{l}) + f_p(l) \right\} \tag{A3}$$

where, $f(l)$, $f(\dot{l})$, $f_p(l)$ are force-length, force-velocity and passive force relations and a is the muscle activation level. $f_{\max}$ is the maximum muscle force based on maximum muscle stress $\sigma_{\max}$ taken as 0.55 MPa which is within the range reported in literature [1,3,17].

Spindle model

The model of stretch reflex invoked in our algorithm was derived from earlier works [25,61]:

$$\begin{aligned} r &= k(l - l_r)\dot{l}^{.3} \quad \text{when} \quad l - l_r > 0 \,, \; \dot{l} > 0 \\ r &= 0.1k(l - l_r) \quad \text{when} \quad l - l_r \geqslant 0 \,, \; \dot{l} \leqslant 0 \end{aligned} \tag{A4}$$

where, r is the spindle discharge rate (firing rate), l is the muscle length, $\dot{l}$ is the muscle velocity, k is the spindle gain factor and l_r is the reference length which is set to be the desired muscle length computed from the desired kinematics at each instance of simulation time. The constant value of 2000 is considered for k in our study. A linear transformation is adopted from the earlier studies to map the spindle firing rate, r, to the muscle activation level a_{feedback} [9,25].

Stability criteria

Substituting the net muscular torque with corresponding muscle forces and their moment arms:

$$J_1 \dot{W} + WW J_1 W + G(\Theta) - N_R = -\mathrm{B}^{\mathrm{T}} \frac{\partial L}{\partial \Theta}^{\mathrm{T}} F \tag{A5}$$

where, B is the transformation matrix between the body coordinate system and that of inertial coordinate system, $\frac{\partial L}{\partial \Theta}^T$ 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 yields, we obtained:

$$J_1 \dot{W}\big|_e + \left(\frac{\partial G(\Theta)}{\partial \Theta}\big|_e + \mathrm{B}^\mathrm{T} \frac{\partial L}{\partial \Theta}^\mathrm{T}\big|_e \frac{\partial F}{\partial L}\big|_e \frac{\partial L}{\partial \Theta}\big|_e \right) (\Theta - \Theta_e)$$
$$+ \left(\mathrm{B}^\mathrm{T} \frac{\partial L}{\partial \Theta}^\mathrm{T}\big|_e \frac{\partial F}{\partial L}\big|_e \frac{\partial L}{\partial \Theta}\big|_e \mathrm{B} \right) W = 0 \tag{A6}$$

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

$$J_1 \dot{W}\big|_e + K(\Theta - \Theta_e) + VW = 0 \tag{A7}$$

where, $J_1 \dot{W}\big|_e$ is the inertial torque about the equilibrium point, K is the stiffness matrix that arises from conservative forces (gravitational and muscle stiffness), V is viscosity matrix that arises from non-conservative forces due to muscle viscosity. In state space form, the Eq. (A8) can be written as:

$$\begin{bmatrix} \dot{\Theta} \\ \dot{W} \end{bmatrix}_{6\times1} = A_{6\times6} \begin{bmatrix} \Theta - \Theta_e \\ W \end{bmatrix}_{6\times1} \tag{A8}$$

In which,

$$A = \begin{bmatrix} zeros_{3\times3} & \mathrm{B}_{3\times3} \\ -J_1^{-1}K_{3\times3} & -J_1^{-1}V_{3\times3} \end{bmatrix}_{6\times6} \tag{A9}$$

To ensure stability of the linearized system, eigenvalues of A should have negative real parts [14,16,35]:

$$REAL\,(Eig(A)) < 0 \tag{A10}$$

Appendix B

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 about L5-S1 in the body coordinate system
N_R	= resistance torque about L5-S1 in the body coordinate system
WW	= skew symmetric matrix corresponding to W
B	= transformation matrix between the body and inertial coordinate systems
$\frac{\partial L}{\partial \Theta}^T$	= matrix of moment arm of muscles (3×48)
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
$PCSA$	= muscle physiologic cross sectional area
$\sigma_{\max}$	= maximum muscle stress
a_m	= muscle activation level of each muscle $(m = 1 \text{ to } 48)$

$f_{\max}$	=	maximum muscle force
l	=	muscle length
l_d	=	desired muscle length
$\dot{l}$	=	muscle velocity
r	=	spindle discharge rate (firing rate)
k	=	muscle spindle gain factor
l_r	=	muscle spindle reference length
$f(l)$	=	force-length relationship of muscle
$f(\dot{l})$	=	force-velocity relationship of muscle
$f_p(l)$	=	passive force relationship of 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

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