A COMPUTATIONAL METHOD FOR SIMULATION OF TRUNK MOTION: TOWARDS A THEORETICAL BASED QUANTITATIVE ASSESSMENT OF TRUNK PERFORMANCE

M. Parnianpour
Department of Industrial, Welding & System Engineering, The Ohio State University, Columbus, Ohio, USA

J. L. Wang
Biomechanics Research Laboratory, Department of Orthopaedic & Rehabilitation
Yale University School of Medicine, New Haven, Connecticut, USA

A. Shirazi-Adl
Department of Mechanical Engineering, Ecole Polytechnique, Montreal, Canada

B. Khayatian
Department of Electrical Engineering, The Ohio State University, Columbus, Ohio, USA

G. Lafferriere
Department of Mathematics, Portland State University, Portland, Oregon, USA

Correspondence to:
Jaw-Lin Wang, Ph.D.
Biomechanics Research Laboratory
Department of Orthopaedic and Rehabilitation
Yale University School of Medicine
P.O. Box 208071, New Haven, CT 06520-8071, USA
Phone: 1-203-785-7139, Fax: 1-203-785-7069
E-mail: JLWang@biomed.med.yale.edu

ABSTRACT

Quantitative assessment of trunk muscle performance is important in documenting the extent of impairment and disability due to low back disorders (LBD). The statistical pattern recognition problem of classifying LBD patients and normal subjects based on dynamic trunk performance has been data driven. To provide clinical insight for interpretation of the distinctive features in the movement profiles, we have suggested an optimization-based approach for simulation of dynamic point-to-point sagittal trunk movement. The effect of strength impairment on movement patterns was simulated based on minimizing different physical cost functions: Energy, Jerk, Peak Torque, Impulse, and Work. During unconstrained simulations, uni-modal velocity patterns are predicted???, while time to peak velocity is distinct for each cost function. The significant differences between unimpaired optimal movement profiles were diminished by imposing an 80% reduction in extensor muscle strength. The results indicate that the search for finding the objective function being used by central nervous system is an ill-posed problem since we are never sure if we have included all the active constraints in the simulation. The four application areas of this study are: 1) providing optimized trajectories for bio-feedback to patients during the rehabilitation process; 2) training workers to lift safely; 3) estimating the task demand based on the global description of the job; and 4) aiding the engineering evaluation to develop ergonomic and workplace interventions which needed to accommodate individuals with prior disability.

Keywords: Trunk muscles, quantitative assessment, ergonomics, low back disorders, simulation

INTRODUCTION

The performance of voluntary movements and the maintenance of trunk posture during various mechanical conditions are achieved by coordinated and controlled trunk muscle activity. The large number of degrees of freedom in the human spine allows movements to be carried out in an infinite number of ways with multiple muscle combinations. Despite the availability of these options, the central nervous system (CNS) demonstrates consistent regularities or ``invariant patterns'' in execution of movement. Several models and theories have been developed to explore these "regularities" by motor control researchers [1,2].

The literature on motor control is sparse with respect to the systematic study of trunk movement. The studies of Thorstensson and colleagues [3,4,5] are among the few systematic studies of trunk movement and muscle recruitment within a motor control paradigm. The plane of movement, direction, amplitude, velocity, and initial posture were varied while the patterns of EMG activity and trunk movements were studied. During voluntary trunk flexion at maximum speed of varying amplitude, both variant and invariant characteristics were revealed [5]. The relative timings of peak velocity, acceleration, and deceleration were invariant whereas the duration of the movement varied, which in part agrees with the concept of a generalized motor program. The duration of initiating activity in primary movers and the time to onset of antagonist activity were highly correlated with amplitude, duration, and peak velocity of the movement, while low correlation was found between temporal aspects of EMG and the peak acceleration. Several of the existing theories of motor control such as the ``speed control system'' [6], the ``pulse step model'' [7] or the ``equilibrium-point model'' [8] did not fit the data. The large mass of the trunk, the intrinsic instability of the spine and the multiple degrees of freedom of the spine should provide a rich medium for refining the existing theories.

The presence of regular spatial and temporal patterns in the muscle activation and movement profiles has led to the development of the following hypothesis. The information content in the movement patterns of the trunk can be used to identify normal healthy subjects and classify low back pain patients into various low back disorder categories. Sharafeddin et al. [9], and Marras et al. [10] have reported more than 90% accuracy in identifying the low back pain patients using two diverse sets of databases and performance assessment tools. Sharafeddin et al. defined the motion patterns measured with a triaxial dynamometer, which provided external resistance to subjects. Meanwhile, Marras et al. used an exoskeletal electro-goniometer to measure the relative motion of trunk with respect to pelvis.

The key limitations in the development of the discriminate functions for classification purposes, are the data-driven nature of the algorithms and the lack of theoretical orientation in the process of development and validation of these models. It is suggested that the mathematical simulation of flexion or extension trunk movement may identify an objective basis for the evaluation and assessment of trunk kinematic performance. A catalog of movement patterns that are optimal with respect to physical and biomechanical quantities may contribute to the emergence of a more theoretically based computational paradigm for the evaluation of kinematic performance of normal subjects and patients. It must be emphasized that in this study we have no intention to claim that the central nervous system actually optimizes any single or composite cost function. The purpose of this paper is to develop a computational tool for movement simulation of the trunk to conduct a number of numerical experiments such as determining the effect of impairment of strength on the kinematics of trunk performance.

METHOD

The most fundamental class of trunk movement is the sagittally symmetric point-to-point flexion and extension. Although the spine is made of complex interconnected six degrees of freedom joints, which include flexible intervertebral discs and nonlinear contact due to engagement of facet joints, we have approximated the system in terms of a rigid body rotation of the torso about the L5/S1. The requirements for an acceptable approximation to our physical system (trunk) is very different for the physician or the physical therapist evaluating one's motion and the neurophysiologist or biomechanist studying the intersegmental motions of the spine in the laboratory [11]. The experimental protocols providing dynamic inter-segmental kinematic data non-invasively are lacking, while numerous technologies have been utilized to get the gross trunk motion. The classical inverted pendulum model for the study of the flexion/extension movement of trunk in the sagittal plane will be used [12]. For this first approximation, we will assume that all the trunk muscles combine to generate a resultant torque about the L5/S1, the center of rotation. The dynamic equation of motion is given by:

$$J\frac{d^2\theta}{dt^2} - B\sin\theta = \tau \qquad L \le \tau \le U \tag{1}$$

where $J = I_c + ml^2, B = mgl$, m is the mass of the trunk, I_c is the moment of inertia of the trunk at its center of mass, l is the distance from the center of mass of the trunk to the axis of rotation, g is the gravitational acceleration, τ is the torque generated by the muscles with its respective lower and upper bounds (L, U), and θ is the angle of trunk about the vertical upright position. The anthropometric data for an individual with height and weight of 1.7 m and 80 kg were used in these simulations [13].

While this is a simplified model it is already fully nonlinear in its characteristics. The forward dynamics problem requires one to integrate this nonlinear differential equation by inputting the unknown muscular torques. By considering the desired angular position, velocity, and acceleration at the beginning and the end of the movement, the system of equations becomes the classical two point boundary value problem or initial boundary value problem. There are infinite trajectories that will satisfy the equations of motion and the boundary/initial conditions. However, to select the optimal trajectory amongst the multiple solutions, the following cost functions can be suggested:

Energy	$\frac{1}{2}\int_0^T \tau^2(t)\,dt$
Jerk	$\frac{1}{2}\int_0^T (\frac{d^3\theta}{dt^3})^2\,dt$
Peak Torque	$\max_{t\in(0,T)} \lvert\tau(t)\rvert$
Impulse	$\int_0^T \lvert\tau(t)\rvert\,dt$

Work $\int_0^T \left|\tau(t)\,\dot{\theta}\right| dt$

In addition, one can constrain the net muscular torque, τ, to simulate the concept of strength impairment in patients. It should be noted that the movement time could be fixed or set as a free parameter (in particular for the minimum time solutions). To allow systematic comparisons of results we have considered the movement time to be fixed at 1 second based on the previous experimental results of Schmitz [14].

No analytical closed form solution can be suggested, despite the simplicity of the model. Nagurka and Yen [15] proposed a Fourier-based approximation to generate near optimal trajectories of general dynamical systems. By representing the time history of each generalized coordinate by the sum of an auxiliary polynomial and a finite-term Fourier-type series, the optimal control problem of the two point boundary/initial value differential equations are reduced to a nonlinear programming problem. The optimal trajectory *θ(t)* is approximated by a fifth degree polynomial plus a linear combination of Fourier terms with weight coefficients.

$$\theta(t) = p(t) + \sum_{i=1}^{k} a_i \cos(\frac{2i\pi t}{t_f}) + \sum_{i=1}^{k} b_i \sin(\frac{2i\pi t}{t_f}) \qquad (2)$$

The angular velocity and acceleration can be obtained analytically using Equation (2). Based on Equation (1), we obtain the torque corresponding to the given trajectory as a function of the coefficients a_i, b_i and the final time t_f (the coefficients of the polynomial are also functions of a_i, b_i, and t_f). Hence, in this approach, instead of forward integration of the equation of motion (forward dynamics), we use the inverse dynamics to compute τ, which is much faster and simpler due to its algebraic structure. The cost function is then also a function of the unknown coefficients a_i, b_i and the final time t_f.

The cost (objective) function will be minimized subject to the equality constraint (nonlinear dynamic equation of motion; initial and final boundary conditions) and the inequality constraints (which could be imposed on τ, $\theta, \dot{\theta}$, and/or $\ddot{\theta}$). The constrained nonlinear programming algorithm was used to simulate the following trunk movements [16].

The movement time was set to one second and the range of motion used for the simulation was 60 degrees for both flexion and extension movements. The motion started from rest and terminated with zero velocity and acceleration. The number of terms in the Fourier series was increased from 2 to 6 in order to determine their effect on the optimal trajectory for various costs. The effect of a global upper bound for the extensor strength was also evaluated for each cost function - *L*= -200 Nm (absolute strength reduction). In addition, we constrained the extensor strength to 80% of its peak value during unconstrained simulations for the same cost function. These latter simulations are presented to determine the effects of a relative extensor strength reduction.

RESULTS

Depending on the cost function, only a few Fourier terms were needed to stabilize the approximation of the optimal trajectory. Figure 1 presents the two extreme cases depicting the influence of the number of Fourier terms used in the model. For the case of minimum Jerk, no improvement was observed, while for the minimization of Work, Peak Torque and Impulse significant differences were predicted in the trajectories and net muscular torques. Minimum Energy profiles were also slightly affected by the number of terms in the Fourier series. The higher accuracy in representation of the optimal trajectory will require larger number of terms which also increases the CPU time needed for convergence. We selected to keep 6 Fourier series terms for the simulations performed in this study.

The optimized unconstrained trunk flexion and extension trajectories are represented for each of the five cost functions in Figure 2. The effect of movement direction on the optimal trajectories is significant for all cost functions with the exception of the minimum Jerk case. The kinematic results indicate that the time to peak velocity is an important feature separating the different cost functions. During extension, the time to peak velocity occurs in the initial stage of the movement while it is delayed to the latter stages during the flexion when minimizing Energy, Work and Impulse. However, the time to peak velocity is delayed during extension which means the acceleration phase is prolonged relative to the deceleration phase of the movement when minimizing the Peak Torque. The distinctions are enhanced in the profiles of the higher derivative measures of velocity and acceleration. The Minimum Impulse and Work profiles are more similar during flexion than in extension movement. The results of these simulations indicate the trade off between using the gravitation moments or muscular moments to drive the trunk. It should be added that these trajectories represent the extreme cases, since the presence of constraints on strength will moderate these behaviors (Figure 3). Nonetheless, the reliance on rapid deceleration of the trunk by the antagonist muscles will require rapid high force development during eccentric muscle action. High eccentric muscle activity has been implicated in muscular injuries and self--limiting myogenic delayed soreness often present after completion of unaccustomed physical exertions [2,14].

The existence of impairment in the maximum peak extension strength, reduced the distinction between the predicted optimized trajectories (Figure 3). The left panels of Figure 3 present the profiles when the lower bound of τ was set to -200 Nm. The right panels of Figure 3 represent the effect of an 80% strength reduction relative to the peak extensor moment predicted for the unconstrained simulations for each cost. These profiles must be compared with the full strength flexion profiles (right panels of Figure 2).

DISCUSSION

For the unconstrained minimum Jerk cases, the Maximum Principle (based on Euler-Poisson's theorem) indicates that the optimal trajectory $\theta(t)$ must have the form $\theta(t) = \sum_{i=0}^{5} d_i t^i$ (i.e. a polynomial of degree 5) [17]. This cost function results the smoothest trajectory, and is purely based on kinematic considerations. This is also the reason that it may not be as relevant since both the postural and inertial loads must be considered during trajectory planning of the trunk (especially during the more complex multi-link lifting tasks) [18].

The results of our predicted minimum Jerk trajectory for unconstrained simulations matched the analytical results since the same set of the polynomial coefficients were predicted (Figure 1). As expected, the addition of 2 to 6 terms in Fourier series did not affect the minimum Jerk optimal trajectory, validating our algorithm. The discrimination between global and local minimal can not be provided under these complex conditions; however, numerous simulations starting from different initial guesses were performed to gain confidence in the reliability of the results. Nevertheless, we cautiously suggest that the identified trajectories are the "*optimized*" rather than optimal solutions. A number of sensitivity analyses were performed to investigate the effects of uncertainty regarding the anthropometric inputs to the model on the optimized trajectories across different cost functions. The moment of inertia of the head, arms and trunk (HAT), J, was perturbed +/- 20%, while keeping the value of B in Equation 1 constant. The kinematic profiles were not significantly affected and the general patterns remained the same for all cost functions. The Peak Torque profiles were most sensitive to the variations of J. The results of the sensitivity analysis are shown for kinematic profiles minimizing Energy and Work (Figure 4).

The cross tabulation of the costs for different optimization trials suggests the existence of the trade-off between various strategies. For the case of unconstrained flexion task, minimizing Energy increases the jerk more than 13.5 fold over its minimum value, while minimizing Jerk only increases energy cost by 20% (Table 1). Minimizing Impulse created the largest jerk, 54 fold higher than its minimum value. Minimum Jerk trajectory, which represents lower wear and tear due to smooth trajectory, provides the lowest relative cost with respect to the other criteria. It is most likely that multiple costs may be optimized at the same time [19,20], and the relative importance of each criterion may be phase dependent during complex multi-joint coordinated tasks [21].

The global or absolute limit to extensor strength caused the predicted trajectories to become more similar to the minimum Jerk solution (Figure 3). The weakness in antagonist muscle group during trunk flexion also created a dynamical constraint in the maximum acceleration that the flexor muscles could generate, hence reducing the flexion angular velocity and acceleration and consequently a lower required maximum flexion torque. The reduction of extensor strength not only causes the reduction in maximum deceleration but also expedites the final deceleration phase of motion. In addition, the duration of the deceleration phase of flexion movement increased as a result of reduction in the extensor strength. Hence, the kinematics and kinetics of trunk flexion are affected in terms of their amplitudes and timings.

The results of this study are in general agreement with the experimental trunk dynamic profiles of 25 normal healthy male subjects (Figure 5). Subjects were asked to move as fast and as accurately as possible to 60 degrees of flexion with a target width of +/- 5 degrees [14]. Both the predicted and measured profiles were ballistic with a single peak in the velocity profiles. However, the most important lesson of this numerical experiment is the fact that identification of the cost function used by CNS is an ill-posed problem [23], which can not be solved by any experimental study. The simulation of strength impairments showed that the optimized profiles became more similar for distinct cost functions (Figures 2 and 3) . Hence, we can not find the appropriate cost function by assessing the similarity between the predicted and measured movement profiles. We also have no idea whether all the active constraints have been included in the model, since the optimized trajectory depends both on the cost function and the constraints imposed on the system (Figure 3).

In this study, the movement time was fixed at one second, based on the experimental study of Schmitz [14]. Treating the movement time as a free variable would have created more variations in the profiles (especially the minimum Work and Impulse trajectories), but the systematic evaluation of the effects of strength impairment and cost functions would not have been possible. However, based on minimum Peak Torque flexion simulation (Figure 2), if the individual has extensor strength less than 175 Nm, the flexion task cannot be performed in one second. Hence, the model suggests that the movement time should be increased to accommodate functional capacity/impairment of this individual. The following observation confirms the clinical findings that low back disorder patients have lower angular velocity and acceleration during maximum trunk flexion and extension [9,10]. The following simulation indicates that although there is controversy regarding the selection of cost function in optimization based simulations, there is no controversy in the predictive power of these simulations to determine the feasibility of a task. These results can aid in identifying the ergonomic intervention needed to accommodate the capability constraints of the individual with disability. This functionality of movement simulation is extremely important with respect to implementation of reasonable accommodation for individuals with prior disability as mandated by the Americans with Disabilities Act (1992).

There are a number of applications for the results of these models. The optimized trajectories could provide biofeedback to low back pain patients during their rehabilitation process. Given the inability to use the uninjured joint as a reference for impairment and disability evaluation of spine, the objective and quantitative benchmarks provided by the model may prove invaluable in assessment of the kinematic performance of low back pain patients. Khalaf et al. [27] have extended the model to include the dynamics of the multi-link system to simulate generalized manual material handling tasks. The present model allows the detailed trajectory planning to be relegated to the computational algorithm after a number of global characteristics of the task are specified (i.e., movement time, initial and final boundary conditions). The ability to simulate trunk movement can allow us to generate the range of demand profiles when we are considering the design of new workplaces or modifications of tasks [28]. Training workers in safe lifting methods is also a promising application, although large validation studies are required.

A number of previous investigators have used the optimal control theory to understand the coordination amongst the multiple muscles during the performance of a goal oriented multiple joint movement such as maximum vertical jumping [22]. Pandy et al. [21] realized that most other physical activities may not have as clear cut objective or goal as athletic or maximum performances. The consideration of kinematic constraints and utilization of a composite objective function that was phase dependent improved the correlation between the observed muscle activation and optimal control solutions during a non-ballistic functional task such as rising from a chair [21]. It must be strongly restated that we are not suggesting that the CNS solves the problem of trajectory planning by use of optimal control. There are numerous other theories that propose more viable alternatives, using the massively parallel distributed networks to solve the problem of managing the redundancy of the neuromuscular system [24,25,26].

Pandy et al. [29] have evaluated the Fourier series approximation techniques in optimal control problems that require bang-bang solution for their control. It can be argued that the constraint for physical realizability of the controls should limit the discontinuity in the controls as predicted by the bang-bang, or bang-coast-bang control strategies of the minimum time and minimum impulse strategies. During the unconstrained simulations, the predicted torques for minimum Work or Impulse strategies are already outside of the maximum strength capability of the normal subjects. The higher Fourier terms would allow to further delay the deceleration of the trunk which will require much higher magnitude of deceleration and extensor torque, when minimizing Work or Impulse during flexion. Hence, the limitation of the approximation in state representation should not significantly affect the results of this study.

The more significant limitation of this study is the simplifying assumption that the multiple muscles act as a torque actuator about L5/S1. The energetic [30,31] and biomechanical considerations of the muscular action such as its tension-length and tension-velocity relationships [28] have not been incorporated in the model. The added complexity of such a refined model was deemed unjustified at this stage. The present model deals with easily measurable or estimated quantities such as overall trunk angular position, velocity and acceleration and its net muscular torque. The governing equation already is fully nonlinear without any closed form solution. The key concepts that this model is trying to assess are still validly illustrated with all its simplifications. We wanted to provide physical bases for evaluation of trunk motion considering the experimental protocols used in numerous industrial or clinical settings [9,32,33]. Muscle driven trunk models [12] will be considered in future simulations as more biomechanical studies incorporate recording of multiple trunk muscle EMG as part of the quantitative assessment of trunk performance [2,34,35]. In addition, future work must incorporate the overlooked effects of spinal deformation, changes in lordosis, the viscoelastic response of passive spine on the load sharing between passive and active neuromuscular spine [36,37,38].

CONCLUSIONS

The computational method proposed for simulation of the trunk movement can contribute to the rehabilitation engineering and biomechanical literature. The four application areas of interest are: 1) providing optimized trajectories for the biofeedback to patients during rehabilitation process; 2) training workers to lift safely (once the model is extended to multiple links); 3) estimating the task demand based on global description of the job; and 4) aiding the engineering evaluation in terms of ergonomics and workplace changes needed to accommodate individuals with prior disability. The present study also emphasizes the importance of including realistic constraints on the maximum performance of the neuro-muscular system.

Acknowledgments: The authors acknowledge partial support from NIDRR, REC grant #H133E30009 and the Ohio Bureau of Workers' Compensation, Division of Safety and Hygiene. Special thanks to Prof. Tom J. Schmitz for providing the experimental data presented in Figure 5. and P. J. Sparto for his invaluable assistance with editing of the text.

REFERENCES

[1] Gottlieb GL, Corcos DM, Agarwal GC: Strategies for the control of voluntary movements with one mechanical degree of freedom. Behav & Brain Sci 1989; 12; 189-250.

[2] Ross EC, Parnianpour M, Martin D: The effects of resistance level on muscle coordination patterns and movement profile during trunk extension. Spine 1993; 18; 1829-1839.

[3] Thorstensson A, Oddsson L, Carlson H: Motor control of voluntary movements in standing. Acta Physiol Scand 1985; 125; 309-321.

[4] Oddsson L: Coordination of a simple voluntary multi-joint movement with postural demands: Trunk extension in standing man. Acta Physiol Scand 1988; 134; 109-118.

[5] Oddsson L, Thorstensson A: Fast voluntary trunk flexion movements in standing: Motor patterns. Acta Physiol Scand 1987; 129; 93-106.

[6] Freund HJ, Budingen HJ: The relationship between speed and amplitude of the fastest voluntary contractions on human arm muscles. Exp Brain Res 1978; 31; 1-12.

[7] Ghez AR, and Vicario D: The control of rapid limb movement in the cat I. Response latency. Exp Brain Res 1978; 33; 173-189.

[8] Feldman AG: Once more on the equilibrium-point hypothesis (λ model) for motor control. J Mot Behav 1986; 18; 17-54.

[9] Sharafeddin H, Parnianpour M, Hemami H, Hanson T, Goldman S, and Madson T: Computer aided diagnosis of low back disorders using the motion profile. 15th South Biomed Engr Conf, Dayton Ohio 1996; 431-432.

[10] Marras WS, Parnianpour M, Ferguson SA, Kim JY, Crowell RR, Bose S, and Simon SR: The classification of anatomic- and symptom-based low back disorders using motion measure models. Spine 1995; 20; 2531-2546.

[11] Steffen T, Rubin RK, Barmaki HB, Antonio J, Marchesi D, Aebi M. A new technique for measuring lumbar segmental motion *in vivo*. Spine 1997; 22; 156-166.

[12] Parnianpour M, Ahmed L, Hemami H, Barin K, and Crowell R. Mathematical formulation of dynamic stability of three dimensional spine during point to point movements: The Role of Coactivation. Proceedings of the 12th Triennial Congress of the International Ergonomics Association 1994; 3; 179-181.

[13] Winter D: Biomechanics and motor control of human movement. New York, Johm Wiley 1979.

[14] Schmitz TJ. The effects of direction and resistance on iso-inertial trunk movement profiles, motor outputs, and muscle coordination patterns during

unidirectional trunk motion. Doctoral Dissertation, New York University, 1992.

[15] Nagurka ML and Yen V: Fourier-based optimal control of nonlinear dynamic systems. ASME J Dynamic Systems, Measurements, and Control 1990; 112;17-26.

[16] Grace A: Optimization toolbox: for use with Matlab. The Mathworks Inc. 1992.

[17] Flash T and Hogan N: The coordination of arm movements: an experimentally confirmed mathematical model. J Neurosci 1985; 7; 1688-1703.

[18] Uno Y, Kawato M, Suzuki R: Formation and control of optimal trajectory in human multijoint arm movement. Biol Cybern 1989; 61; 89-101.

[19] Nelson WG. Physical principles for economics of skilled movements. Biol Cybern 1983; 46; 135-147.

[20] Seif-Naraghi AH and Winters JM: Optimized strategies for scaling goal directed dynamic limb movements. In Winters JM and Woo SLY, eds. Multiple Muscle Systems: Biomechanics and Movement Organization. Springer-Verlag, New York, 1990; 312-334.

[21] Pandy MG, Garner BA, Anderson FC: Optimal control of non-ballistic muscular movements: a constraint-based performance criterion for rising from a chair. Trans. ASME J Biomech Eng 1995; 117; 15-26.

[22] Pandy MG and Zajac FE: Optimal muscular coordination strategies for jumping. J Biomechanics 1991; 24; 1-10.

[23] Latash ML: Control of human movement, Human Kinetics Publisher, 1993.

[24] Pellionisz AJ, Ramos CF: Geometrical approach to neural net control of movements and posture. In Allum JHJ, Allum-Mecklenburg DJ, Harris FP, Probst R, eds., Progress in Brain Research 1993; 97; 245-256.

[25] Pellionisz AJ. Coordination: a vector-matrix description of transformations of overcomplete CND coordinates and a tensorial solution using the Moore-Penrose generalized inverse. J Theor Biol 1984; 110; 353-375.

[26] Naussbaum MA, Martin BJ, Chaffin DB: A neural network for simulation of torso muscle coordination. J Biomechanics 1997; 30; 251-258.

[27] Khalaf KA, Parnianpour M, Wade L, and Sparto PJ: Biomechanical simulation of manual multi-link coordinated lifting. 15th South Biomed Engr Conf, Dayton Ohio, 1996; 197-198.

[28] Khalaf KA, Parnianpour M, Sparto PJ, Simon SR: Modeling of functional trunk muscle performance: Interfacing ergonomics and spine rehabilitation in response to the ADA. J Rehab Res Development, In press.

[29] Pandy MG, Anderson FC, Hull DG: A parameter optimization approach for the

optimal control of large scale musculoskeletal systems. J Biomechanical Eng 1992; 114; 450-460.

[30] Oguztoreli MN and Stein RB: Optimal task performance of antagonistic muscles. Biol Cybern 1990; 64; 87-94.

[31] Stein RB, Oguztoreli MN, Capaday C: What is optimized in muscular movements? In Jones NL, McCartney N, McComas AJ, eds. Human Muscle Power, Human Kinetics Publishers, Champaign, IL, 1986; 131-150.

[32] Marras WS, Lavender SA, Leurgans SE, Rajulu SL, Allread WG, Fathallah FA, and Ferguson SA: The Role of dynamic Three-dimensional trunk motion in occupationally-related low back disorders. Spine 1993; 18; 617-628.

[33] Parnianpour M: Applications of quantitative assessment of human performance in occupational medicine, In Bronzino J. (ed.). Handbook of Biomedical Engineering. CRC Press, Boca Raton. 1995; 1230-1239.

[34] Tan J, Parnianpour M, Nordin M, Willems B, Hofer H: Isometric maximal and sub-maximal trunk extension at different flexed positions in standing: Triaxial torque output and EMG. Spine 1993; 18; 2480-2490.

[35] Gracovetsky S: Musculoketal function of the spine. In Winters JM and Woo SLY, Multiple Muscle systems: Biomechanics and movement organization, Springer-Verlag; 1990; 411-437.

[36] Shirazi-Adl A, Parnianpour M: Stabilizing role of moments and pelvic rotation on the human spine in compression. ASME J Biomech Eng~~r~~ 1996; 118; 26-31.

[37] Shirazi-Adl A, Parnianpour M: Role of posture in mechanics of the lumbar spine. J Spinal Disorders 1996; 9; 277-286.

[38] Wang JL, Shirazi-Adl A, Engin AE, Li S, and Parnianpour M: A model study of the effect of rate of loading on the internal stress of an L2-L3 lumbar motion. ASME Advances in Bioengineering, BED-31, 1995; 19-20.

Table 1. The cross-tabulation of the evaluated five normalized costs for the simulation of unconstrained trunk flexion.

	The evaluated normalized cost function				
Min Cost	Energy	Jerk	Peak Torque	Impulse	Work
Energy	1.000	13.476	1.734	1.427	1.035
Jerk	1.199	1.000	1.201	1.811	1.164
Peak Torque	1.712	10.126	1.000	2.511	1.652
Impulse	1.461	54.624	2.791	1.000	1.007
Work	1.323	24.456	2.322	1.098	1.000

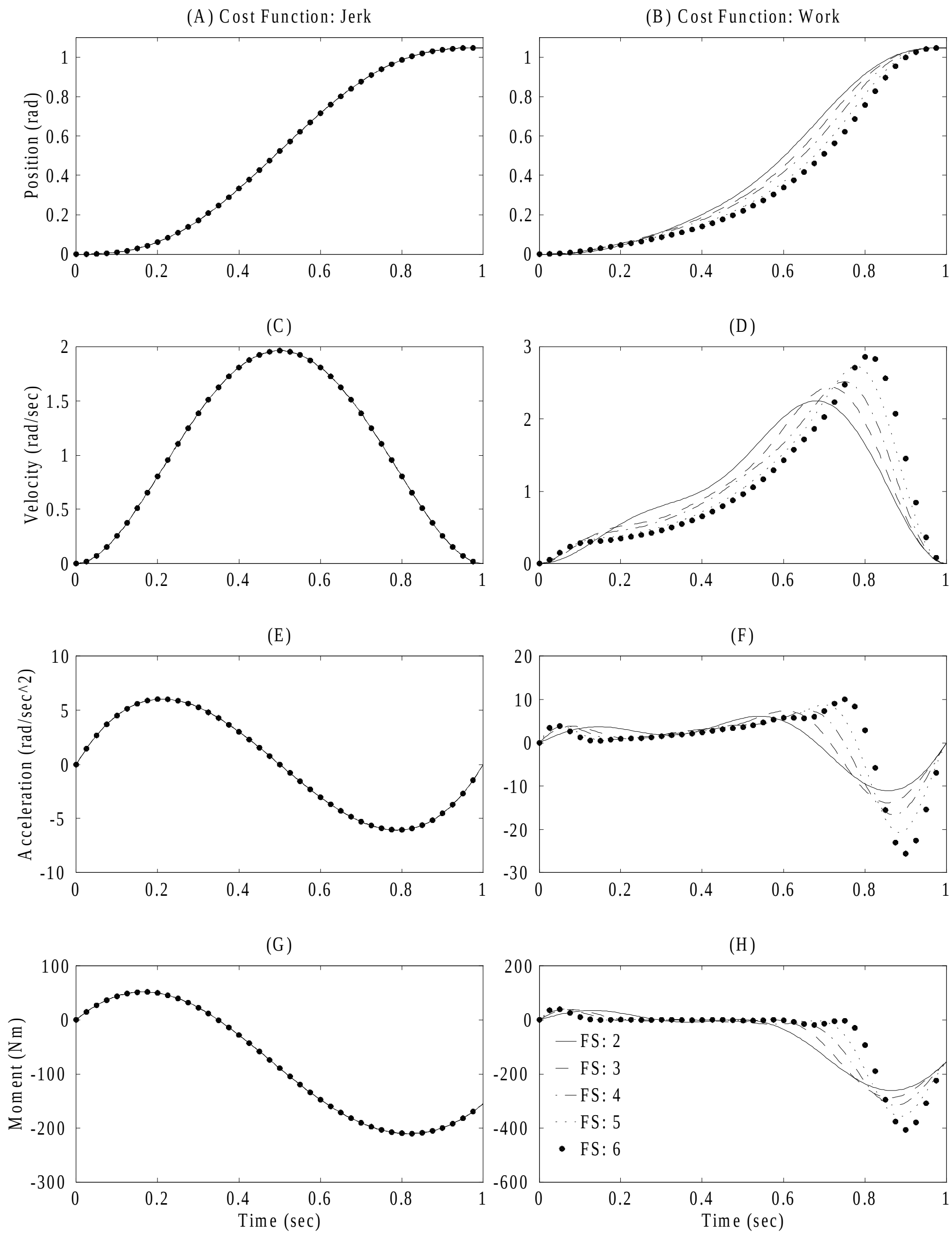
Figure 1. The effect of number of Fourier terms used in the predicted optimized flexion trajectories for minimizing the Jerk (left panels) and Work (right panels).

Figure 2. The optimized unconstrained trunk flexion and extension trajectories for the five cost functions.

Figure 3. The optimized trunk flexion trajectories for the five cost functions with the relative and absolute peak extensor strength impairments (see text for further descriptions).

Figure 4. The sensitivity analysis to demonstrate the effect of +/- 20\% variation of the estimated value of moment of inertia of HAT (head, arms and trunk) about the center of rotation, $J = 10\ kg\dot{m}^2$, while keeping B constant for minimum Energy and Work profiles.

Figure 5. The ensemble averages with +/- 1 standard deviations of the trunk flexion kinematic profiles for 25 normal male subjects (Schmitz 1992).



(A) Cost Function: Jerk
(B) Cost Function: Work
(C)
(D)
(E)
(F)
(G)
(H)
Position (rad)
Velocity (rad/sec)
Acceleration (rad/sec^2)
Moment (Nm)
Time (sec)
Time (sec)
FS: 2
FS: 3
FS: 4
FS: 5
FS: 6

Figure 1.

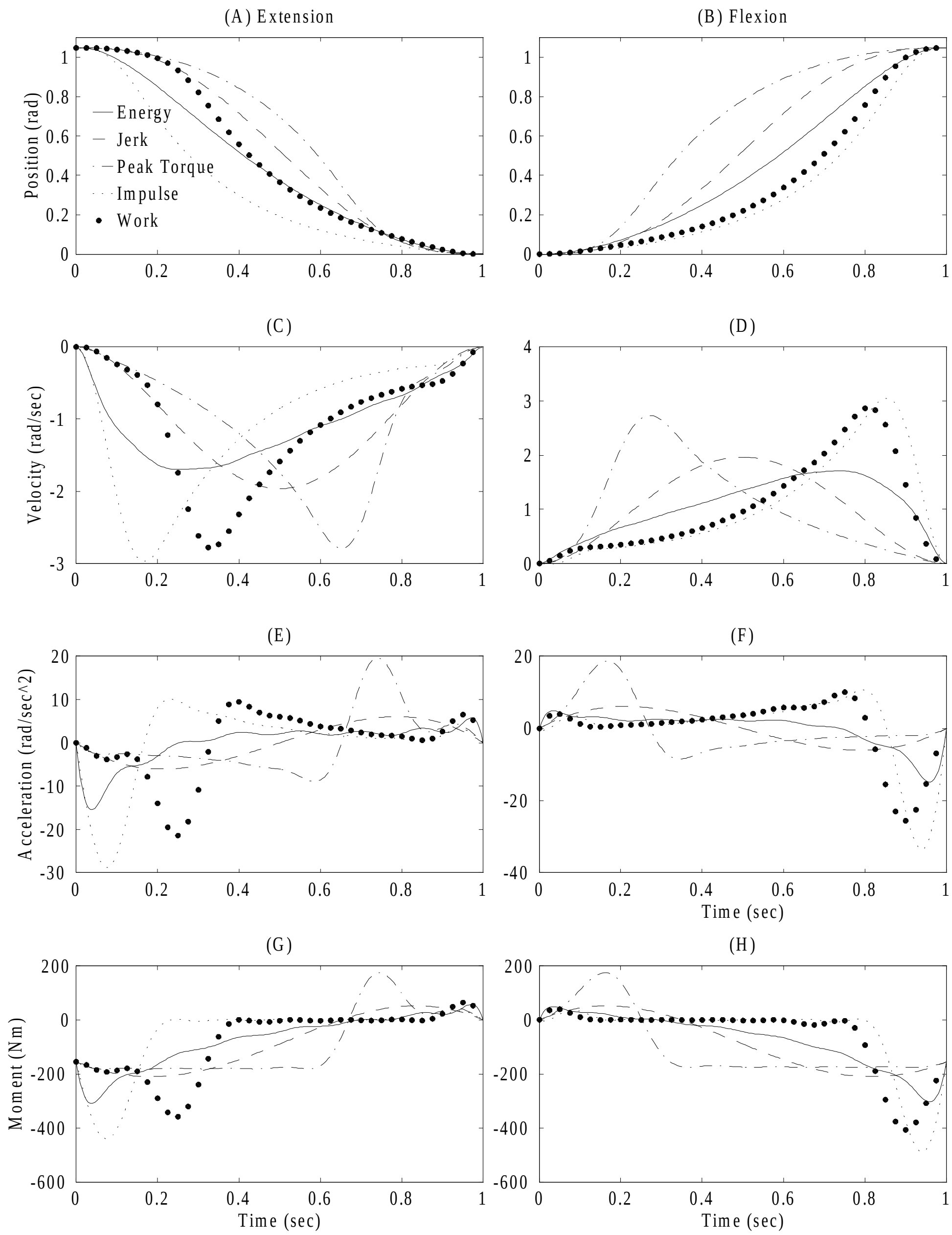

(A) Extension
(B) Flexion
Position (rad)
Energy
Jerk
Peak Torque
Impulse
Work
(C)
(D)
Velocity (rad/sec)
(E)
(F)
Acceleration (rad/sec^2)
Time (sec)
(G)
(H)
Moment (Nm)
Time (sec)
Time (sec)

Figure 2.

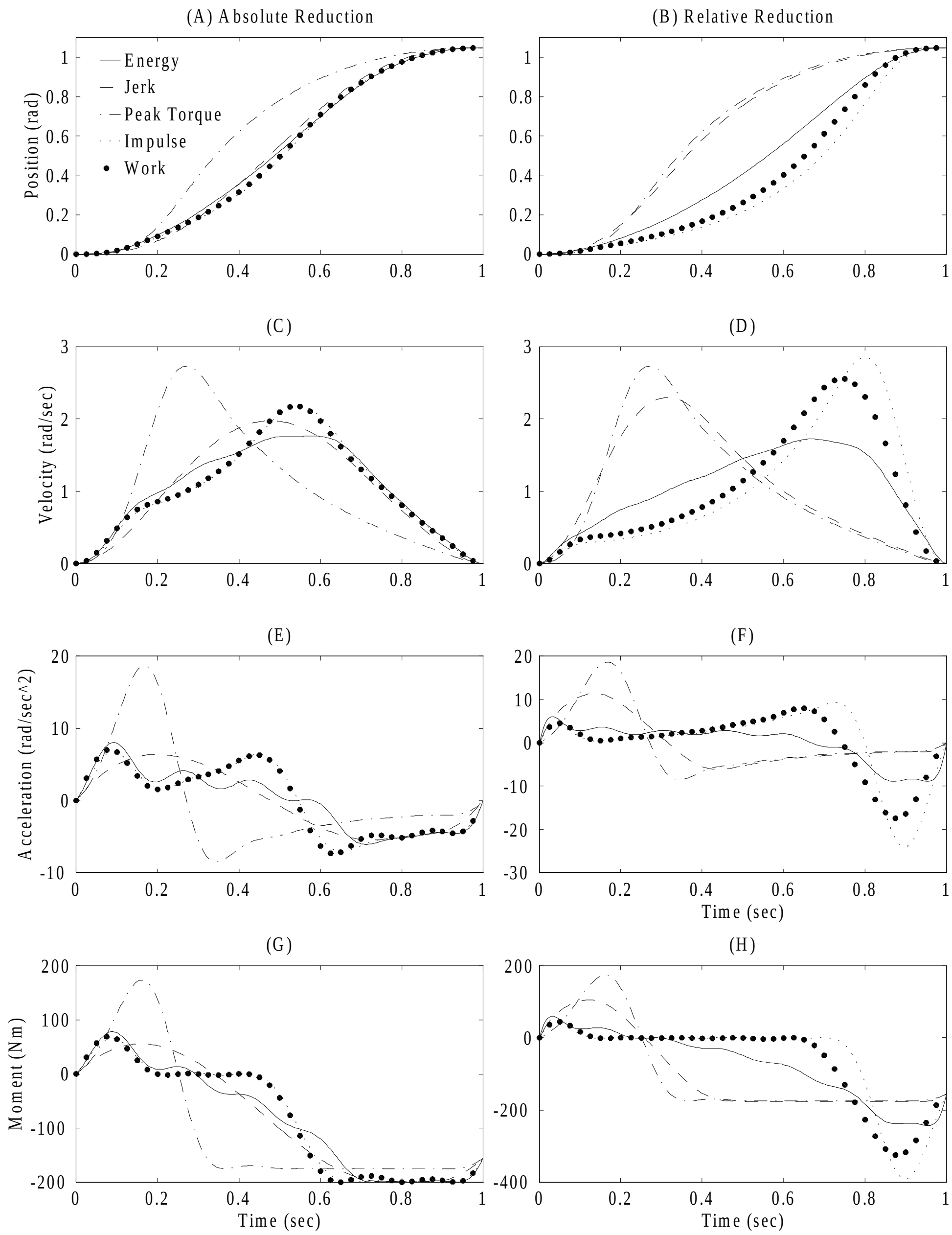

(A) Absolute Reduction
(B) Relative Reduction
Energy
Jerk
Peak Torque
Impulse
Work
Position (rad)
(C)
(D)
Velocity (rad/sec)
(E)
(F)
Acceleration (rad/sec^2)
(G)
(H)
Moment (Nm)
Time (sec)

Figure 3.

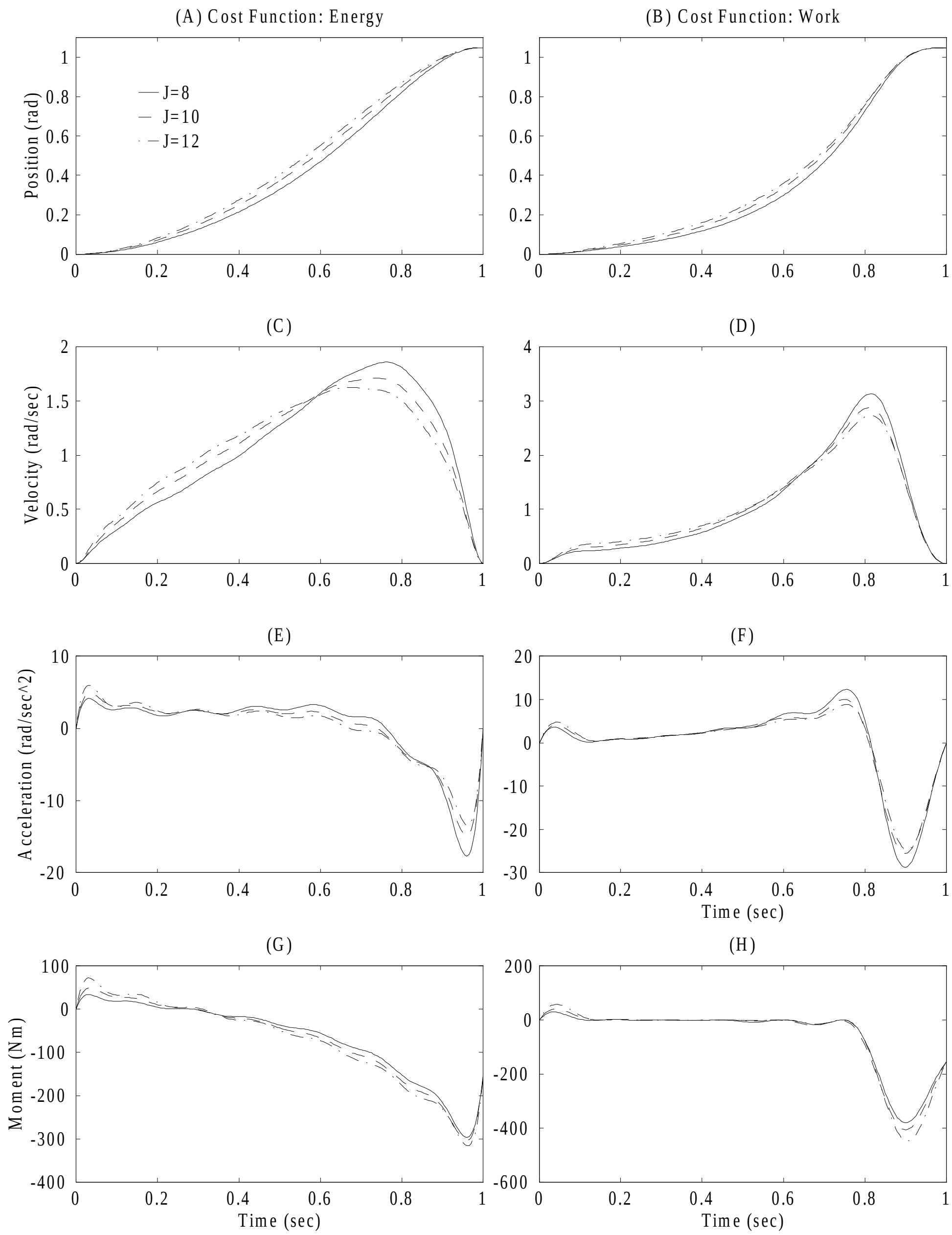

(A) Cost Function: Energy
(B) Cost Function: Work
J=8
J=10
J=12
Position (rad)
(C)
(D)
Velocity (rad/sec)
(E)
(F)
Acceleration (rad/sec^2)
(G)
(H)
Moment (Nm)
Time (sec)

Figure 4.

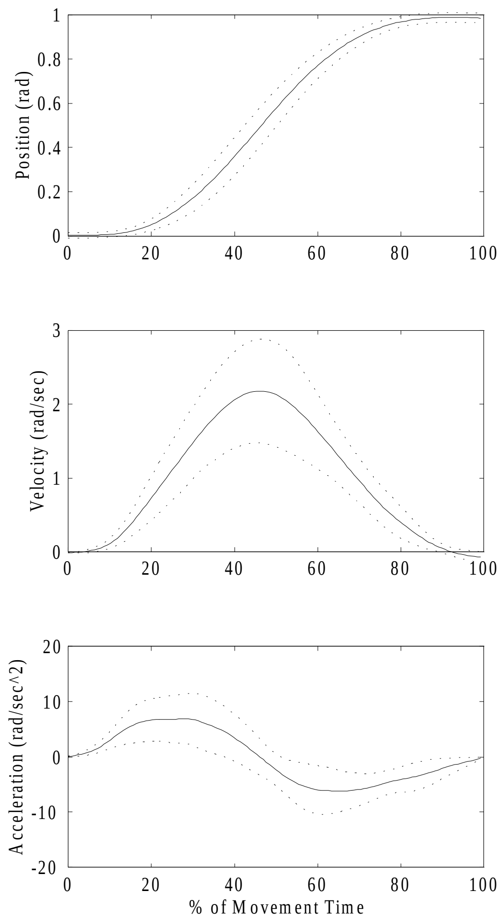

Position (rad)
1
0.8
0.6
0.4
0.2
0
0 20 40 60 80 100
Velocity (rad/sec)
3
2
1
0
0 20 40 60 80 100
Acceleration (rad/sec^2)
20
10
0
-10
-20
0 20 40 60 80 100
% of Movement Time

Figure 5.