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BIODYNAMIC OF THE HUMAN SPINE

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ÉCOLE POLYTECHNIQUE DE MONTRÉAL

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To my wife Parisa,

my parents, my sister, and my brothers

your work ethic and endless encouragement have constantly inspired me

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Résumé

La douleur dorsale est un fardeau socio-économique significatif pour notre société vieillissante. La gestion efficace des troubles du dos dépend de l'évaluation fiable du comportement mécanique du tronc (c.-à-d. charges sur la colonne vertébrale, forces des muscles, déformation/contrainte dans les composants passifs et actifs du tronc, et la stabilité du système) dans diverses activités professionnelles et récréationnelles. Les difficultés techniques et les considérations éthiques associées avec les méthodes de mesures invasives ainsi que des limitations dans les méthodes de mesures indirectes font des modèles biomécaniques un outil indispensable pour l'évaluation de la biomécanique du tronc. Les modèles biomécaniques existants de la colonne ont négligé ou trop simplifié la résistance passive non linéaire, la géométrie/chargement/dynamique complexe de la colonne, les conditions d'équilibre dans tous les niveaux et les directions spinales, et la trajectoire courbée des muscles d'extenseur.

Dans cette étude, une approche itérative basée sur la cinématique a été développée pour alléger les limitations dans les modèles biomécaniques antérieurs. Des données des mesures cinématiques ont été introduit dans un modèle non linéaire d'éléments finis et des équations du mouvement ont été numériquement résolues pour calculer la réponse dans le temps sujet d'un chargement de la gravité et de l'inertie. De cette manière, l'équilibre satisfait à tous les niveaux et directions de la colonne ainsi que les forces calculées des muscles, les charges sur la colonne, et la stabilité du système étaient entièrement en accord avec la cinématique prescrite et les propriétés non linéaires passives. La méthode a été utilisée pour évaluer l'effet sur la biodynamique du tronc de certaines conditions de charge et paramètres mécaniques tels que les différentes techniques de levage (c.-à-d. penché et accroupie), vitesse du mouvement du tronc (vitesses lentes, moyennes et rapides), la vibration du corps entier et le choc, la trajectoire courbée des muscles globaux d'extenseur, les activités antagoniques des

muscles du tronc, les changements des postures, la co-activité abdominale et les changements des propriétés passives de la colonne et des fesses.

Notre recherche sur la biodynamique du tronc sous différentes techniques de levage a prouvé que les moments totaux, les forces des muscles, les forces passives (muscles ou ligaments) et les forces internes de compression/cisaillement étaient plus grandes pour la technique du levage penchée que celle accroupie. Pour la tache du levage relativement lent performée dans cette étude avec les phases de flexion et extension chaque ~2s de durée, l'effet de l'inertie et de l'amortissement n'étaient pas généralement important. En plus, le changement de position de la charge externe du levage penchée atteignant le même bras du levier (par rapport à S1) que celui du levage accroupi n'a pas influencé les conclusions de cette étude sur la suprématie des levages accroupis vis a vis de celui penché. Ayant comme conséquence, pour les taches considérées, les résultats recommandent le levage accroupi que celui penché comme technique de choix en réduisant les moments totaux, les forces des muscles et les charges internes de la colonne. Les changements des propriétés passives de la colonne ont sensiblement influencé les forces des muscles, les charges vertébrales et la stabilité du système dans les deux techniques de levage, mais bien plus dans la posture penchée que celle accroupie. La stabilité de la colonne s'est nettement améliorée avec les grandes propriétés passives, la flexion du tronc et les charges externes. La simulation des muscles globaux d'extenseur courbé a considérablement diminuée les charges vertébrales et améliorée la stabilité du système au cours des taches que celle avec des muscles droits.

Les charges de la colonne et les forces musculaires estimées pendant les mouvements libres de la flexion-extension aux vitesses variables étaient sensiblement plus grandes dans le cas le plus rapide que celui le plus lent indiquant ainsi l'effet des forces d'inertie. La stabilité de la colonne a été améliorée dans de plus grands angles de flexion du tronc et le cas du mouvement le plus rapide. La relaxation partielle ou complète des muscles globaux d'extenseur s'est produite seulement pendant le mouvement le plus lent. Quelques muscles lombaires locaux, particulièrement dans les sujets avec une plus grande flexion lombaire et à des rythmes plus lents, ont également démontré la relaxation en flexion. Les résultats ont confirmé le rôle crucial de la vitesse du mouvement sur la biomécanique de la colonne. Les prédictions ont également démontré le rôle important de l'amplitude de la rotation lombaire maximale et de sa variation temporelle sur la réponse.

Notre étude de la réponse humaine à une vibration du corps entier a montré que, l'excitation de l'input à la base, par l'intermédiaire des forces d'inertie et des muscles, a sensiblement influencée les charges de la colonne et la stabilité du système. La posture fléchie d'un sujet assis a augmenté le moment total, les forces des muscles et les charges passives de la colonne tout en améliorant la stabilité du tronc. D'une façon similaire, l'introduction du bas à modérer la co-activité antagonique dans les muscles abdominaux a augmenté les charges passives de la colonne et a amélioré sa stabilité. Un compromis, par conséquent, existe entre les petites forces des muscles et les charges de la colonne d'une part et la colonne la plus stable d'autre part. Les excitations à la base avec une plus grande accélération augmentent sensiblement les forces des muscles/charges de la colonne et, par conséquent, le risque de blessures.

Les prédictions étaient en concordance bien qu'aux forces de réaction à la base et qu'aux accélérations mesurées à différents niveaux de la colonne. En outre, l'accord qualitatif avec l'activité enregistrée d'électromyographie à différents muscles superficiels a été également trouvé. L'approche dynamique basée sur la cinématique a été démontrée afin de rapporter des données fiables sous diverses taches. De telles données sont cruciales pour la prévention et le traitement efficaces des maux de la colonne. Les futures applications de l'approche basée sur la cinématique pour l'étude des manutentions avec la torsion et la flexion latéral du tronc, la vibration du corps entier avec une plus grande accélération mettent certainement beaucoup de lumières sur la biodynamique du tronc dans des circonstances à haut risque de la lésion dorsale. En outre, le développement et l'intégration d'un modèle d'éléments finis non linéaire de la colonne cervicale au modèle courant fourniraient un meilleur terrain pour la future recherche sur la biodynamique de la colonne humaine.

Abstract

Back pain is a significant socioeconomic burden on our aging society. Effective management of back disorders depends on reliable estimation of trunk mechanical behaviour (i.e. spinal loads, muscle forces, stress/strain in passive and active trunk components, and spinal stability) in various occupational and recreational activities. Practical difficulties and ethical considerations associated with invasive direct and indirect measurement methods along with limitations in indirect measurement methods leave biomechanical models as the indispensable tool for assessment of the trunk biomechanics. Existing biomechanical models of the spine have either neglected or oversimplified the nonlinear passive resistance, complex geometry/loading/dynamics of the spine, equilibrium requirements in all spinal levels and directions, and the wrapping of extensor muscles.

In the present study, an iterative dynamic kinematics-based method was developed to alleviate limitations in earlier biomechanical models. Measured kinematics data were input into a nonlinear finite element model and differential equations of motion were numerically solved to calculate required joint moments and forces, subject to gravitational and inertial loading. In this manner, while satisfying equilibrium at all spinal levels and direction, calculated muscle forces, spinal loads, and system stability were in full accordance with prescribed kinematics and nonlinear passive properties. The method was employed to evaluate the effect on trunk biodynamic of some loading conditions and mechanical parameters such as different lifting techniques (i.e. stoop and squat), velocity of trunk movement (slow, medium and fast velocities), whole body vibration and shock, wrapping of global extensor muscles, antagonistic trunk muscle activities, changes in posture, abdominal co-activity and alterations in passive properties of the spine and buttocks.

Our investigation on trunk biomechanics under different lifting techniques showed that net moments, muscle forces, passive (muscle or ligamentous) forces and

internal compression/shear forces were larger in stoop lifts than in squat ones. For the relatively slow lifting tasks performed in this study with the lowering and lifting phases each lasting ~2s, the effect of inertia and damping was not, in general, important. Moreover, posterior shift in the position of the external load in stoop lift reaching the same lever arm with respect to the S1 as that in squat lift did not influence the conclusions of this study on the merits of squat lifts over stoop ones. Results, for the tasks considered, advocate squat lifting over stoop lifting as the technique of choice in reducing net moments, muscle forces and internal spinal loads (i.e., moment, compression and shear force). Alterations in passive properties of spine substantially influenced muscle forces, spinal loads and system stability in both lifting techniques, though more so in stoop than in squat. Stability of spine substantially improved with greater passive properties, trunk flexion and load. Simulation of global extensor muscles with curved rather than straight courses considerably diminished loads on spine and increased stability throughout the task.

Estimated spinal loads and muscle forces during free flexion-extension movements were significantly larger in fastest pace as compared to slower ones indicating the effect of inertial forces. Spinal stability was improved in larger trunk flexion angles and fastest movement. Partial or full flexion relaxation of global extensor muscles occurred only in slower movements. Some local lumbar muscles, especially in subjects with larger lumbar flexion and at slower paces, also demonstrated flexion relaxation. Results confirmed the crucial role of movement velocity on spinal biomechanics. Predictions also demonstrated the important role on response of the magnitude of peak lumbar rotation and its temporal variation.

Finally, our study of human response to a whole body vibration showed that, the input base excitation, via inertial and muscle forces, substantially influenced spinal loads and system stability. The flexed posture in sitting increased the net moment, muscle forces and passive spinal loads while improving the trunk stability. Similarly, the introduction of low to moderate antagonistic coactivity in abdominal muscles increased the passive spinal loads and improved the spinal stability. A trade-off, hence, exists between lower muscle forces and spinal loads on one hand and more stable spine on the other. Base excitations with larger acceleration contents substantially increase muscle forces/spinal loads and, hence, the risk of injury.

Predictions agreed well with measured base reaction forces and accelerations at different spinal levels. Moreover, qualitative agreement with recoded electromyography activity at different superficial muscles was also found. The kinematics-based approach was demonstrated to yield reliable data under various occupational tasks. Such data are crucial for effective prevention and treatment of spinal disorders. Future applications of kinematics-based approach to investigate manual material handling task with twisting and lateral bending of the trunk, whole body vibration with much larger acceleration contents will certainly shed light on the biomechanics of the trunk under circumstances with high risk of back injury. Moreover, development and integration of a nonlinear finite element model of the cervical spine to the current model would provide a better ground for future investigation on the biodynamic of the human spine.

Condensé en Français

Introduction

Les maux de dos lombaires sont les troubles musculosquelettiques les plus répandus et les plus couteux dans l'industrie. À travers les années, le coût associé à ces troubles a augmenté d'une façon constante. On estime que les coûts directes de compensation des ouvriers ont dépassé les 11.4 milliards de dollars aux EU pour l'année 1989 (Webster et Snook 1994) tandis que le coût médical direct des maux lombaires a dépassé les 24 milliards de dollars pour l'année 1990 (Frymoyer et Chats-Baril 1991). En additionnant les coûts associés à la perte des salaires et de production, à la réhabilitation ainsi qu'au coût de formation et de recrutement des ouvriers remplaçants, le coût total engendré par ces troubles s'avère très significatif pour la société.

Les troubles lombaires peuvent émaner de différentes origines, certaines sont associées à des facteurs professionnels, alors que d'autres plutôt d'un aspect personnel n'ayant aucun rapport avec le travail physique. Les facteurs mécaniques sont cependant reconnus comme la cause principale des douleurs lombaires. Selon des études épidémiologiques, les facteurs de risque reliés au travail physique (mécanique) peuvent se subdiviser en cinq catégories à savoir: (1) manutention manuelle (MM), (2) tache répétitive et fatigue, (3) vibration du corps entier (VCE) et choc, (4) posture maladroite, et (5) posture de travail statique combiné avec l'un des facteurs mentionnés (NIOSH 1997, Pope et al 2002). Les facteurs psychosociaux qu'ils soient reliés ou non au travail ont été également associés aux troubles lombaires, pourtant on leur a attribué le fait d'influencer la chronicité et le comportement des douleurs plutôt que d'en être leur cause (Adams, 2004.).

Des études pathologiques associent un large éventail de lésions lombaires à des blessures de compression et de torsion (Bogduck et Twomey 1991). Par conséquent, la détermination précise de la répartition des charges parmi les composants passifs et actifs du tronc humain dans diverses activités est essentielle pour la prévention, l'évaluation et le traitement efficaces des troubles de la colonne vertébrale.

L'infaisabilité de la mesure directe des forces musculaires et des charges vertébrales chez les êtres humains d'une part, le souci du coût, les limitations et difficultés associées à la mesure indirecte de ces variables de l'autre, font que les modèles biomécaniques soient des outils indispensables pour l'évaluation biomécanique du tronc pendant diverses activités professionnelles et récréationnelles. En général, pour calculer les forces inconnues des muscles en utilisant les modèles mécaniques, les moments de charges externes sont d'abord déterminés en utilisant la dynamique inverse soit analytiquement s'il s'agit d'un système de segments de corps rigides ou numériquement pour le cas d'un système de corps déformables en utilisant la méthode des éléments finis. Les moments calculés des charges externes devraient être équilibrés intérieurement par les composantes passives et actives du tronc. Ce problème d'équilibre s'avère être redondant puisque le nombre des forces inconnues de muscles est supérieur au nombre d'équations disponibles.

Pour résoudre la redondance dans les équations d'équilibre, quatre approches de modélisation ont été introduites: la réduction, l'optimisation, l'approche basée sur les signaux d'électromyographie (EMG-assistée), et l'approche basée sur la cinématique ('Kinematics-based approach'). Dans la méthode de réduction, les muscles sont groupés pour former des ensembles synergiques (c.-à-d. ayant la fonction semblable) afin de réduire le nombre de forces inconnues au nombre d'équations. Évidemment, cette méthode n'estime pas les forces dans chaque muscle du tronc (comme entité individuelle) incluant divers fascicules des muscles extenseurs du dos qui sont individuellement attachés à différents niveaux spinaux pour avoir des rôles distincts pendant différentes activités (Bogduk et al 1992). Le principe de la méthode d'optimisation est qu'il présuppose l'existence d'une fonction de coût (par exemple la minimisation de somme de tensions des muscles) qui peut être optimisée par le système nerveux central, tout en respectant des égalités (c.-à-d. équations d'équilibre) et inégalités (c.-à-d. les limites sur les contraintes des muscles). Cependant, lorsque les méthodes d'optimisation sont soumises uniquement aux contraintes d'équilibre, elles sont incapables de prévoir la co-activation dans les muscles antagonistes. Il a été démontré que l'introduction de la contrainte de stabilité dans la méthode d'optimisation allait automatiquement générer une co-activité des muscles antagonistes (Zeinali et al, 2007, Stokes et Gardner-Morse, 2001). La méthode 'EMG-assistée' emploie les signaux d'électromyographie traités des muscles du tronc pour estimer les forces musculaires. En dépit de la polémique concernant la nature des relations linéaires ou non-linéaires entre les forces musculaires et les données d'EMG et de son extension aux conditions dynamiques, on note une certaine évolution de ces modèles. Toutefois, on note que les forces musculaires calculées par la méthode EMG-assistée ne satisfait pas nécessairement les équations d'équilibre et ont besoin d'être calibrées en utilisant des facteurs de 'Gain'. Finalement, dans l'approche basée sur la cinématique, des données cinématiques sont appliquées généralement dans toutes les directions et niveaux vertébraux et par conséquent, diminuent la redondance du problème (Kiefer et al. 1997). Cette méthode satisfait les équations d'équilibre à tous les niveaux et dans toutes les directions comparativement aux modèles fréquemment utilisés tel que le modèle à niveau unique dans lequel l'équilibre est vérifié à un seul niveau seulement (Arjmand et al. 2007).

Les modèles biodynamiques de la colonne vertébrale dont le but est d'évaluer les forces musculaires et les charges vertébrales, peuvent être divisés en trois groupes distincts traitant la manutention manuelle, la vibration du corps entier ainsi que l'impact/chute. Les modèles de segment ('link segment models') fournissent simplement les charges d'inertie et de gravite à une section transversale spécifique sans la contribution des forces internes. Par la suite, il est nécessaire d'utiliser un modèle de distribution pour le calcul des forces musculaires du tronc et des charges de la colonne en prenant en compte les forces passives de la colonne vertébrale. Le calcul de telles forces musculaires est souvent basé sur la considération de l'équilibre à un seul niveau, ce qui cause une violation de l'équilibre aux niveaux restants (Arjmand et al. 2007). Bien que, la méthode des éléments finis soit reconnue pour être l'approche la plus appropriée pour modéliser la colonne vertébral (Shirazi-Adl et Parnianpour 2001, Reeves et Cholewicki 2003, Liu 1982, Sidel 2005), l'application de cette méthode dans les activités de manutention manuelle est rare et se limite généralement qu'aux analyses statiques. Les modèles par éléments finis qui existent présentement pour l'étude de la vibration du corps entier et de l'impact ne considèrent pas correctement les propriétés non linéaires des segments de la colonne et la redondance dans le système musculo-squelettiques du tronc. En plus, la trajectoire courbée des muscles d'extenseurs dorsaux sous l'effet des mouvements avec des flexions larges ou modérées du tronc a souvent été présumé linéaire ou incorrectement simulée (Cholewicki et McGill1996).

La stabilité de la colonne vertébrale est une autre considération importante qui influence le recrutement des muscles. Les charges critiques en flambement des colonnes thoraco-lombaires et lombaires dans le plan coronal prennent les valeurs d'environ 20 N et 88 N, respectivement (Crisco et Panjabi 1991, Lucas et Bressler 1961, Shirazi-Adl et Parnianpoour 1993 et 1996). Néanmoins, pendant les activités quotidiennes, la colonne subit des charges axiales beaucoup plus importantes sans présenter de signes d'instabilité (Arjmand et Shirazi-Adl 2005, 2006, EL-Riche et autres 2004). On a montré que la stabilité de la colonne vertébrale est liée à l'angle de flexion du tronc (Arjmand et Shirazi-Adl 2006, Cholewicki et McGill 1996), aux charges externes (Elrich et autres 2004, 2005, Arjmand et Shirazi-Adl 2006), à la co-activité des muscles antagoniques (El-Rich et Shirazi-Adl 2004, Cholewicki et al 1999), à la lordose lombaire (Arjmand et Shirazi-Adl 2005), ainsi qu'à la pression intra-abdominale (Cholewicki et autres 1999, Arjmand et Shirazi-Adl 2005). Cependant, aucune étude sur la stabilité de la colonne dans les taches dynamiques n'a examiné le rôle probable des techniques de levage, des propriétés passives de la colonne, des changements dans la vitesse de mouvement et de la vibration du corps entier.

Puisque l'approche basée sur la cinématique a la capacité de surmonter plusieurs lacunes des anciens modèles, l'objectif principal de cette étude est de développer une méthode pour des conditions transitoires afin d'estimer les forces musculaires, charges spinales et la stabilité du tronc dans les taches dynamiques. Cette méthode va servir par la suite à évaluer l'effet de quelques facteurs de risque importants sur la biomécanique du tronc. Ceux-ci incluent l'évaluation de la biomécanique du tronc sous différentes techniques de levage (c.-à-d. penchée et accroupie), mouvement libre de flexion-extension du tronc (à trois vitesses) et vibration du corps entier et choc. De plus, l'effet sur les résultats de certains paramètres additionnels tels que la trajectoire courbée des muscles extenseurs globaux, des propriétés passives de la colonne et des fessiers, la co-activité des muscles abdominaux ainsi que les caractéristiques dynamiques du tronc ont été évalués.

L'approche dynamique basée sur la cinématique (Dynamic Kinematics-Based Approach)

Cette approche itérative exploite des données de la cinématique pour produire des équations additionnelles à chaque niveau de la colonne afin d'alléger la redondance de la cinétique dans le système. Au début, la cinématique mesurée du tronc (rotations au plan sagittal à différents niveaux vertébraux et aux translations soit au niveau S1 ou au niveau des fesses) et les charges externes/gravité sont prescrites dans un modèle non linéaire d'élément fini de la colonne. L'algorithme implicite a été employé pour résoudre le problème transitoire non linéaire, ayant pour résultat une variation de temps des moments de réaction à chaque niveau vertébral qui doivent êtres équilibrer par des muscles attachés à ce niveau. Pour résoudre la redondance restante à chaque niveau, une approche d'optimisation avec une fonction objective de la somme minimale des contraintes cubiques de muscle est employée. Les équations d'inégalité sont basées sur des forces musculaires positives inconnues qui sont plus grande que leurs composantes passives restantes (calculées sur la longueur instantanée des muscles et sur la relation tension-longueur (*Davis et al 2000*)) mais plus petite que la somme de leurs forces actives maximales (c.-à-d., 0.6 MPa par section physiologique de muscle, PCSA (Winter 2005)) et les composantes passives de ces forces. À la fin de chaque itération, la pénalité des forces des muscles à différents niveaux dans les directions axiale et du cisaillement est appliquée avec les charges externes à la colonne et ce procédé est répété jusqu'à ce que la convergence soit réalisée (c.-à-d. les forces calculées des muscles dans deux itérations successives demeurent presque les mêmes). Une fois que les forces des muscles sont calculées tout au long de la période de la tache, la stabilité du système est étudiée en remplaçant des muscles avec des éléments uniaxiaux. La rigidité, k, de chaque élément uniaxial est assignée en utilisant la relation linéaire de rigidité-force (c.-à-d. k=q F/L) où la rigidité de muscle est proportionnelle à la force musculaire instantanée, F, et inversement proportionnel à sa longueur actuelle, L, avec q comme un coefficient sans dimensions de rigidité des muscles pris pour être le même pour tous les muscles (Bergmark 1989). À chaque instant, la marge de stabilité pour différentes valeurs de q est étudiée à des configurations déformées chargées par les analyses de la fréquence naturelle et de la perturbation linéaire. Le code de calcules d'éléments finis ABAQUS (version ABAQUS 6.5) est employé pour des analyses linéaires et non-linéaires de stabilité alors que le procédé d'optimisation est analytiquement résolu par la méthode de multiplicateur de Lagrange.

Le modèle sagittal symétrique du tronc est fait de six poutres déformables non linéaires pour représenter les segments T12-S1 et sept éléments rigides pour représenter tête-T12 (comme un seul corps) et vertèbres (L1-S1) lombosacral (Arjmand et Shirazi-Adl, 2005, 2006 ; Bazrgari et al, 2007 ; Bazrgari et Shirazi-Adl, 2007). La relation non linéaire entre la charge et le déplacement dans différentes directions, et les différences de la flexion par rapport a l'extension basés sur les résultats numériques et ceux mesurés précédemment sont représentées (Oxland et al, 1992 ; Pop, 2001 ; Shirazi-Adl et al, 2002 ; Yamamoto et al, 1989). Une architecture sagittale symétrique de muscle contenant 46 muscles locaux (fixé aux vertèbres lombaires) et 10 globaux (attaché à la cage thoracique) a été employée. Les masses, les moments de masse d'inertie, et le centre de masse correspondant à différents niveaux du tronc le long de la colonne sont définis en se referant sur des données publiées (de Leva, 1996 ; Pearsall et al, 1996 ; Zatsiorsky et Seluyanov, 1983). L'amortissement inter segmentaire est assigné en utilisant des valeurs mesurées (Kasra et al, 1992 ; Markolf, 1970). Des fesses à la base sont modélisées par un élément de connecteur (compression seulement) avec les propriétés non-linéaires de rigidité basées sur des données rapportées dans la littérature (Aimedieu et al, 2003 ; Kitazaki et Griffin, 1997a) et un amortissement semblable à celui des segments lombaires.

Pour évaluer l'effet des différentes techniques de levage sur la biomécanique du tronc (chapitre 2), des mesures in vivo de cinématique ont été prescrites dans le modèle pour estimer les forces des muscles du tronc et les charges internes de la colonne pour le levage dynamique en posture accroupie et pour celle penchée avec et sans une charge de 180 N. Des mesures ont été effectuées sur des sujets en bonne santé pour rassembler les rotations segmentaires pendant les levages requis comme données d'input pour des modélisations. La contribution des charges d'inertie a été étudiée en comparant les résultats des analyses isométriques à ceux dynamiques. En conséquence, la stabilité de la colonne et la distribution de la charge externe entre les composantes passives et actives du tronc au cours des levages en posture dynamique accroupie et penchée ont été étudiées, lorsque la rigidité en flexion de la colonne vertébrale a été changée (chapitre 3). Des effets sur les prévisions des changements d'amortissement segmentaire, la position des charges externes ainsi que la trajectoire courbée des muscles extenseurs globaux ont été également étudiés. Dans les études suivantes (chapitre 4), des mesures in vivo de la cinématique et les forces de réaction du plancher ont été effectuées sur des jeunes sujets asymptomatiques tout en exécutant la flexion et l'extension libre à trois vitesses différentes pour évaluer l'effets des changements de la vitesse sur l'activation des muscles, les charges de la colonne, les forces de réaction et la stabilité du système. En conclusion, l'approche dynamique itérative basée sur la cinématique (Dynamic

Kinematics-Based Approach) a été utilisée pour évaluer les forces des muscles, les charges de la colonne et la stabilité du système pour un sujet soumis à une excitation verticale aléatoire de base avec des chocs d'accélération de ~ ±1 g (chapitre 5). L'accélération verticale du siège d'input était une accélération non pondérée à l'interface du siège conducteur d'une excavatrice hydraulique (Seidel et al 1997). Les effets de la posture, la co-activité des muscles abdominaux et les changements de la rigidité des fesses ont été également étudiés.

Résultats

Les mesures in vivo ont démontré des rotations plus grandes du thorax du bassin, et des rotations de la colonne lombaire (T12-S1) dans des élévations penchées comparés à des élévations accroupis ($p < 0.05$, 0.05 et 0.03, respectivement). L'importance de la charge externe (0 N contre 180 N), n'a eu aucun effet significatif sur ces rotations. Le moment externe maximal calculé au disque L5-S1 augmente sensiblement lorsque la technique d'élévation penchée est effectué (~28% comparé aux élévations accroupis). Ces moments sont dus principalement par les muscles avec une petite contribution (~10-30% selon la technique de levage) des segments passifs de la colonne. Au niveau T12 et dans les deux conditions de charge, les moments de résistances des muscles globaux d'extenseur et de la colonne vertébrale étaient les deux plus grands pour le levage penché que dans le levage accroupis. La contribution des forces actives des muscles, particulièrement dans le cas avec une charge de 180 N, était plus grande que des forces passives des muscles. Les différences relatives dans les forces globales des muscles dans la posture accroupie contre celle penchée étaient dues principalement à des composants passifs plus petits dans la première technique de levage. Les forces maximales des muscles à différents niveaux locaux et globaux étaient plus grandes dans le cas du levage penché que celui du levage accroupis. Les forces locales internes de compression et de cisaillement à différents niveaux intervertébraux de disque étaient également plus grandes dans le cas du levage penché avec des différences maximales atteignant ~800 N dans la compression et ~200 N dans le cisaillement au niveau du L5-S1. Les forces de cisaillement calculées ont montré une augmentation importante du niveau L4-L5 au L5-S1 pour tous les cas. En raison des plus grandes rotations lombaires, les moments segmentaires passifs étaient également plus grands pour le même cas.

Le coefficient critique de rigidité des muscles change de manière significative en fonction de l'angle de flexion du tronc et de la charge externe (c.-à-d., flexion vers l'avant sans la charge et l'extension avec charge). La marge relative de stabilité dans les deux techniques de levage changées dans la phase de flexion sans charge, étant plus petite dans les périodes de début de la flexion (c.-à-d. de plus grandes valeurs critiques de q) et plus grande au fur et à mesure que la flexion du tronc augmentait (c.-à-d. de plus petites valeurs critiques de q). Les changements de la rigidité passive de la colonne influencent sensiblement les résultats des techniques de levage penchée et accroupie, mais plus dans le premier cas. La réduction de la rigidité passive augmentait les forces de compression et de cisaillement par contre les tendances inversées étaient calculées quand une plus grande rigidité passive de la colonne a été assumée. Le moment sagittal au L5-S1 change par conséquent par des valeurs maximales de 32.2 Nm et 12.9 Nm pour le levage penché et accroupi, respectivement. Ces changements des moments ont par conséquence des changements compensatoires des forces de muscles et, par conséquent, des charges de compression et de cisaillement. La diminution de 40% de la rigidité de flexion n'a sensiblement pas changé les forces de cisaillement du disque supérieur, mais elle augmentait celui au niveau L5-S1 de 89 N et 106 N du levage penché et accroupi, respectivement. Dans ce cas-ci, des forces de compression ont été augmentées par des valeurs maximales de 602 N et 271 N au niveau L5-S1 du levage penché et accroupi, respectivement. En raison des plus grandes rotations lombaires, l'influence des changements des propriétés passives de la colonne sur les forces de contact entre les muscles extenseurs et la colonne vertébrale était beaucoup plus prononcée pour le levage penché. Les changements des propriétés passives de la colonne ont influencé la stabilité, principalement pendant les débuts du levage lorsque les sujets ne supportaient aucune charge et la flexion du tronc était relativement petite.

Excepté les périodes de temps au début et à la fin des taches et cela juste après le levage de la charge externe, les résultats étaient presque les mêmes pour des analyses statiques et dynamiques le long de la durée du levage. L'inclusion de l'inertie de la charge externe dans l'analyse a eu des effets négligeables sur les résultats. L'augmentation de l'amortissement du disque n'a pas changé les résultats, alors que les fluctuations dans la réponse (±10 Nm le moment requis du thorax) ont été observées en l'absence d'amortissement dans le modèle. En s'approchant de la positionnement de la charge externe dans le levage penché (<88 millimètres afin d'arriver au même bras de levier relatif en ce qui concerne le S1 considéré dans le levage accroupi) toutes les forces des muscles ont été réduite aussi bien pour le moment externe, que la compression et le cisaillement au niveau L5-S1 à des valeurs entre celles prévues pour le levage accroupi et penché; mais encore toujours plus grande que celles du levage accroupi. La considération de la trajectoire non linéaire des muscles globaux d'extenseur par rapport au modèle avec des muscles globaux droits a sensiblement influencé les résultats ; la force de compression a diminué à tous les niveaux (par exemple à peu près de 543 N au niveau L2-L3) et la force de cisaillement a augmenté aux niveaux les plus bas. Négliger les trajectoires courbées pour les muscles globaux et leurs forces de contact respectives (dus à l'interaction avec la colonne) a comme conséquence, un système moins stable durant toute la flexion du tronc mais en dépit des forces de muscles plus grandes.

Dans le chapitre 4, des mesures in vivo en cinématique de 14 sujets portant sur des mouvements libres de flexion-extension à trois vitesses différentes ont été employées comme données d'entrée dans le modèle. Plutôt de simuler un sujet simple d'une part ou tous les 14 sujets d'autre part, on a décidé de prendre trois sujets avec des rotations lombaires extrêmes (maximale et minimale) et moyennes pour évaluer l'effet du mouvement du tronc sur la biomécanique de la colonne. De cette manière et selon nos objectifs avec une quantité raisonnable d'effort, des effets des variations de la vitesse du mouvement et la rotation lombaire sur des résultats ont été pris en considération. Les moyennes des rotations maximales (écart-type) du tronc, du bassin, et du lombaire (en degrés) pour tous les sujets étaient respectivement 113.5 (7), 57.5 (10.2), 56 (6.5) en vitesse lente, 113.5 (6.2), 59.1 (10.3), 54.4 (7.3) en vitesse intermédiaire, et 122.6 (5.2), 65.5 (8.7), 57.1 (7.7) en vitesse rapide du mouvement. Les mouvements les plus rapides ont duré ~1.9, 2.2, et 3 secondes pour des sujets respectivement `Min', `Max', et `Mean', comparé avec la moyenne de ~2.4 secondes (écart-type = 0.7) pour tous les sujets. La signification des forces d'inertie pour le mouvement le plus rapide comparé à celui le plus lent était apparente dans le cas des forces de réaction calculées et mesurées dans chacun des trois sujets. Les forces d'inertie maximales de 317 (405), 230 (309) et 104 (118) N ont été calculées selon la composante verticale des forces de réaction au S1 (et par la plateforme de la force) pour la vitesse la plus rapide pour des sujets `Min', `Max', et `Mean', respectivement.

Pour lancer la flexion vers l'avant avec la vitesse la plus rapide, une activité abdominale au début de cette flexion a été observée. En revanche, presque aucune activité abdominale n'a été constatée au début des taches lentes. Tous les sujets ont démontré une activité abdominale à des grands angles de flexion au cours des mouvements lents. Pour de plus grands angles de flexion, tous les sujets ont montré une plus grande activité des muscles globaux d'extenseur au cours du mouvement le plus rapide et un phénomène de relaxation en flexion dans les mouvements lents. Les forces des muscles lombaires locaux ont suivi des tendances semblables à ceux pour les muscles globaux. La relaxation complète en flexion des muscles lombaires locaux a été constatée seulement à quelques niveaux dans les sujets avec de plus grandes rotations lombaires (c.-à-d., 'Mean' et 'Max').

Le moment externe total calculé au S1 a sensiblement changé entre le mouvement le plus rapide et les deux mouvements lents. Contrairement au mouvement le plus rapide, des mouvements plus lents ont démontré même une diminution du moment sagittal aux grands angles de flexion. Le moment total maximal a sensiblement augmenté de ~ 83%, 22% et 65% de la vitesse intermédiaire à celle la plus rapide pour les sujets 'Min', 'Mean' et 'Max', respectivement. Les charges de la colonne et les moments totaux atteignent leur maximum au niveau L5-S1. Les charges de la colonne à différents niveaux étaient plus grandes dans le mouvement le plus rapide que ceux intermédiaire et lent, mais avec des différences négligeables entre le deux derniers mouvements. Les angles extrêmes de flexion considérés dans cette étude ont comme conséquence des changements considérables des trajectoires des muscles globaux d'extenseur. Le contact des muscles globaux d'extenseur avec la colonne vertébrale a rapporté de grandes forces de contact aux différents niveaux, ces forces augmentent au fur et à mesure que les forces de muscle et la rotation lombaire augmentent. L'analyse de stabilité pour les sujets 'Min' et 'Mean' avec des vitesses plus rapides et intermédiaires a démontré que la colonne était tout à fait stable dans la flexion profonde; et n'a exigée aucune rigidité de muscle quand les angles vers l'avant de flexion du tronc atteignent 19.5° (38°) et 52° (62°) pour les sujets 'Min' (et 'Mean') avec des vitesses plus rapides et intermédiaires, respectivement.

L'input d'accélération verticale du siège, employée pour évaluer l'effet de la vibration du corps entier sur la biomécanique du tronc (chapitre 5), changeait principalement dans la gamme de ±5 ms^{-2} avec la valeur de RMS de 1.4 ms^{-2} et deux crêtes relativement pointues (c.-à-d. -12 et +10 ms^{-2}). Le moment total au niveau S1 restait presque positif pendant la période de vibration et étaient plus grand dans la posture fléchi que celle droite. La contribution passive de la colonne pour équilibrer le moment externe, était initialement (au début de la vibration) 48% et 87% et a diminué pour atteindre des valeurs minimales de 13% et 22% au moment de l'accélération positive maximale pour les postures droites et fléchis, respectivement. Les forces locales de compression et de cisaillement ont atteint leurs valeurs maximales au niveau L5-S1 et étaient plus grandes dans la posture fléchie avec des valeurs maximales 1608 N et 771 N, respectivement, par rapport aux valeurs 1131 N et 508 N de la posture droite. Les forces associées calculées dans les extenseurs globaux (LG et IC) et les muscles de l'abdomen (RA, IO et EO) n'ont montré aucune co-activité et ont suivi les tendances temporelles prévues pour les moments totaux et les charges de la colonne. Le coefficient critique de rigidité de muscle pour maintenir la stabilité du tronc a été sensiblement influencé par l'activité du muscle et la posture. Le tronc était beaucoup plus stable (c.-à-d., petites valeurs critiques de q) pendant les périodes d'activité des muscles abdominaux et au contraire moins stable aux périodes d'activité maximale des muscles d'extenseur. La posture fléchie a comme conséquence une amélioration globale de la stabilité du tronc, due principalement à la plus grande résistance fournie par les tissus passifs.

L'introduction de la co-activité antagonique dans les muscles abdominaux IO, EO et le RA à 2%, 1% et 0.5%, respectivement, a sensiblement amélioré la stabilité du tronc dans la posture droite (c.-à-d., exigence des valeurs critiques beaucoup plus petites de q). Les co-activités abdominales antagoniques prescrites avant ont diminué le coefficient critique de rigidité des muscles de 89 à 21 au début de la vibration et de 131 à 39 à l'accélération maximale positive. Les charges de la colonne, au contraire, sont augmentées dans les mêmes périodes à peu près de 196 N pour la force de compression et de 138 N pour la force de cisaillement. La diminution de la rigidité des fesses n'a pas affecté la réponse autant. Par contre, l'augmentation de cette rigidité a augmenté la valeur de l'accélération; par exemple les accélérations maximales à la tête du sujet ont atteint ±17 ms^{-2} pour le cas des fesses rigides. En outre la fréquence naturelle du système sous la charge de la gravité et les postures prescrites était de ~5.5 Hz, pour devenir ~12.4 Hz ou 3.9 Hz dans le cas des fesses complètement rigides ou plus molles, respectivement.

Conclusion

Une architecture détaillée des muscles du tronc avec les propriétés non-linéaires de la colonne vertébrale, la trajectoire courbée des muscles globaux d'extenseur et les caractéristiques dynamiques du tronc (inertie et amortissement) ont été employés dans notre approche dynamique basée sur la cinématique afin d'évaluer les forces des muscles, les charges de la colonne et la stabilité du tronc au cours des levages en posture accroupie et penchée, des mouvements de flexion-extension, et la vibration aléatoire du corps entier à la base. Les prédictions ont satisfait la cinématique et les conditions dynamiques d'équilibre à tous les niveaux et directions de la colonne. Pour les taches considérées, les résultats recommande le levage accroupi que penché comme technique de choix en réduisant les moments totaux, les forces des muscles et les charges internes de la colonne. Ces valeurs sont demeurées plus grandes, même lorsque le bras du levier de la charge externe du levage penché ait été réduit pour égaler celui du levage accroupi. En outre, puisque les levages relativement lents ont été performés et modélisé dans ce travail, les caractéristiques dynamiques du tronc n'ont pas démontré des effets significatifs sur les résultats. Les changements des propriétés passives de la colonne semblables à des changements de la flexion du tronc et aux charges externes ont influencé d'une manière significative les charges de la colonne pendant les levages dynamiques penchés et accroupis, mais plus pour le premier que le dernier. De tels changements des propriétés passives de la colonne n'ont pas cependant affecté la stabilité du système. D'ailleurs, négliger la trajectoire réaliste des muscles globaux d'extenseur en modélisant les taches à de plus grands angles de flexion, ont comme conséquence une évaluation plus grande des charges de la colonne mais de plus petites marges de stabilité.

Notre étude a confirmé le rôle crucial de la vitesse du mouvement et la rotation lombaire sur la dynamique de la réponse, l'activation des muscles, la relaxation en flexion, les charges internes de la colonne et la stabilité du tronc. La vibration du corps entier avec un certain contenu élevé d'accélération (c.-à-d. le choc) augmente sensiblement les charges de la colonne et détériore la stabilité du système. Les charges et la stabilité de la colonne ont été aussi démontrées sensibles à la posture lombaire, au rythme lombaires/pelviens, au bras du levier de la charge et les muscles, à la trajectoire courbée des muscles globaux, aux propriétés passives de la colonne et aux muscles du tronc. Des prédictions ont été vérifiées pour être en accord général avec les mesures disponibles. Les résultats promettent également un grand soutien de l'évaluation fiable des forces des muscles, des charges de la colonne et de la stabilité du tronc et, par conséquent, dans la gestion adéquate des désordres de la colonne. Les futures applications de l'approche basée sur la cinématique afin d'étudier la tache de la manipulation manuelle du matériel avec la torsion et la flexion latérale du tronc, la vibration du corps entier avec un contenu beaucoup plus grand d'accélération mettent certainement beaucoup de lumières sur la biodynamique du tronc dans des circonstances à haut risque de la lésion dorsale. D'ailleurs, le développement et l'intégration d'un modèle d'éléments fini non linéaire de la colonne cervicale au modèle courant fourniront un excellent terrain pour les futures recherches sur la biodynamique de toute la colonne humaine.

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List of Abbreviations

ALL	Anterior Longitudinal Ligament
C	Compression force
C1 to C7	Cervical Spine Vertebrae
CNS	Central Nervous System
CSI	Cervical Spine Injury
EMG	Electromyography
EO	External Oblique
Fa	Active Muscle Force
Fp	Passive Muscle Force
FRP	Flexion Relaxation Phenomenon
Ft	Total Muscle Force
IAP	Intra-Abdominal Pressure
ICpl	Iliocostalis lumborum pars lumborum
ICpt	Iliocostalis lumborum pars thoracic
IDP	Intra-Discal Pressure
IO	Internal Oblique
IP	Iliopsoas
L1 to L5	Lumbar Spine Vertebrae
LBP	Low Back Pain
LED	Light Emitting Diode
LGpl	Longissimus thoracis pars lumborum
LGpt	Longissimus thoracis pars thoracic
M	Moment
MF	Multifidus
MMH	Manual Material Handling
MSD	Musculoskeletal Disorder
NIOSH	The National Institute for Occupational Safety and Health

PCSA	Physiological Cross Sectional Area
PLS	Posterior Ligamentous System
QL	Quadratus Lumborum
RA	Rectus Abdominus
S	Shear force
S1	Sacrum Vertebra #1
T1 to T12	Thoracic Spine Vertebrae
TLF	Thoraco-Lumbar Fascia
WBV	Whole Body Vibration

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CHAPTER ONE

INTRODUCTION

1.1 Back Pain

Musculoskeletal Disorders (MSDs) are the major component of work related illnesses in terms of cost and work absenteeism (NIOSH, 1997). Among others, Low Back Pain (LBP), is the most prevalent and costly MSD in industry. A life time prevalence of up to 70% has been associated with LBP (Lawrence et al., 1998) while 4-5% of the population experiences an acute low back pain episode every year (Plante et al., 1997). LBP accounts for 16-19% of all workers compensation claims, but 33-41% of the total associated cost (Spengler et al., 1986; Webster and Snook, 1994). On average, health care cost of individuals with back pain is about 60% higher than that of individuals without LBP ($3498 vs. $2178 in 1998) (Luo et al., 2004). The greatest expense (i.e. 75%) is from a small group of LBP cases (i.e. 25%), given the fact that individuals are more likely to have another back pain episode after the first experienced episode (Ferguson et al., 2004). The cost associated with LBP has steadily risen over the years. Webster and Snook (1994) estimated that LBP in 1989 incurred at least $11.4 billion in direct workers' compensation costs. Frymoyer and Cats-Baril (1991) estimated that direct medical cost of back pain in US for 1990 exceeded $24 billion. In 1998, total health care expenditure attributable to LBP exceeded $90 billion with total incremental expenditure of $26 billion (Luo et al., 2004). In Great Britain, rates of incapacity for work because of back problems increased more than sevenfold between 1953 and 1992 (Coggon, 2003). Considering additional costs associated with lost wages, loss of production, cost of recruiting and training replacement workers, cost of rehabilitating the affected workers, and etc, the total cost to national economies becomes even greater. In order to reduce the risk of future back pain, and alleviate pain and suffering we must gain a better understanding of the low back injuries and mechanisms involved.

1.2 Back Pain Risk Factors

1.2.1 Personal and psychosocial risk factors

Back disorders may arise from a variety of origins and can be associated with both occupational and non work-related factors (i.e. personal). The latter may include age, gender, smoking, physical fitness level, obesity, height, pregnancy, anthropometric measures, lumbar mobility, strength, medical history, and structural abnormalities (Garg and Moore, 1992; Pope et al., 1991). A review of 57 industrial-based epidemiological studies (Ferguson and Marras, 1997) indicated that personal factors were the most investigated risk factor for LBP. The tolerance of spine components have been shown to change as a function of individual's factors (Jager et al., 1991; Koeller et al., 1986; Mayer et al., 2001). Spinal loads have also been found to be affected by some of these factors (e.g. pain history, gender). For instance, LBP patients were found to experience 26-75% greater spine loading (Marras et al., 2001), mitigating the reported odd ratio of 9.8 for new episodes of LBP in those having back pain twice a year (van Poppel et al., 1998). Another study found greater spinal loads, up to 20%, in female than male when angular velocity of torso was greater than 45 °/sec (Marras et al., 2002). It should be noted that although the strength of evidence associating individual factors to LBP is mild, most of these risk factors are beyond an individual or society's control (Marras, 2000).

Psychosocial factors, both work and non work – related, have also been associated with back disorders, though they have been suggested to influence the pain behavior rather than pain origination (Adams, 2004). Factors such as job dissatisfaction, monotony of work, limited job control, low job clarity, and lack of social support are the most identified potential risk factors (Davis and Heaney, 2000; NIOSH, 1997). Back injury's odd ratios of 1.3-1.9 for job satisfaction (Bigos et al., 1991; Bigos et al., 1986; Marras et al., 1995; Marras et al., 1993), 2.6 for poor social environment, and 1.6 for coworker supports (Norman et al., 1998) have been reported in the literature. It has, however, been pointed out that it is impossible to separate the contribution of the physical factors from that of psychosocial components of the work in corresponding epidemiological studies (Davis and Heaney, 2000). Nevertheless, consideration of physical factors (i.e. biomechanical influence) can have a significant impact upon the strength of findings related to the psychosocial factors (Marras, 2000). Despite the recognition of the psychosocial factors as important LBP risk factors, the mechanism through which these factors might cause pain is poorly understood. Some speculated that poor psychosocial environment would create an environment where workers are more likely to report injury and illness. Marras et al (2000) showed that spinal compression and lateral shear can respectively increase up to 14 and 27% in subjects while performing lifting under stress (Fig. 1.1). Similarly, spinal loads were found to be affected by the degree of mental processing and pacing required during a physical task (Davis et al., 2002).

1.2.2 Mechanical work related risk factors

Reviewing epidemiological studies, physical (mechanical) work related risk factors can be categorized in five following groups: (1) Manual material handling (i.e. lifting, lowering, pushing, pulling), (2) Motion, repetitive task, and fatigue (i.e. effects of inertial loads and viscoelastic behavior), (3) Whole body vibration and shock, (4) Awkward posture (i.e. combined flexion/torsion/lateral bending), and (5) Static work posture combined with any of foregoing factors (NIOSH, 1997; Pope et al., 2002). Due to their relevance to this study, these factors are further elaborated in the next sections.

1.2.2.1 Manual material handling (MMH)

Association of low back disorders with lifting (i.e. moving or bringing something from a lower level to a higher one) and forceful movement (i.e. movement of objects in other directions such as pulling, pushing, or other efforts) is supported by strong evidence (NIOSH, 1997). 25-70% of back injuries are associated with MMH with the highest rate of injuries due to overexertion being in nursing and personal cares facilities

(NIOSH, 1997). In a large retrospective survey, lifting or bending episodes accounted for 33% of all work related causes of back pain (Damkot et al., 1984). Bigos et al (1986), in the Boeing study, showed that manual handling task was associated with 63% of low back compensation cases. During lifting, spinal loads (i.e. compression and shear) are believed to exceed threshold of the tissue strength resulting in spinal injury (e.g. by irritation of nerve root via vertebral end-plate fracture or disc herniation) (Chaffin et al., 2006).

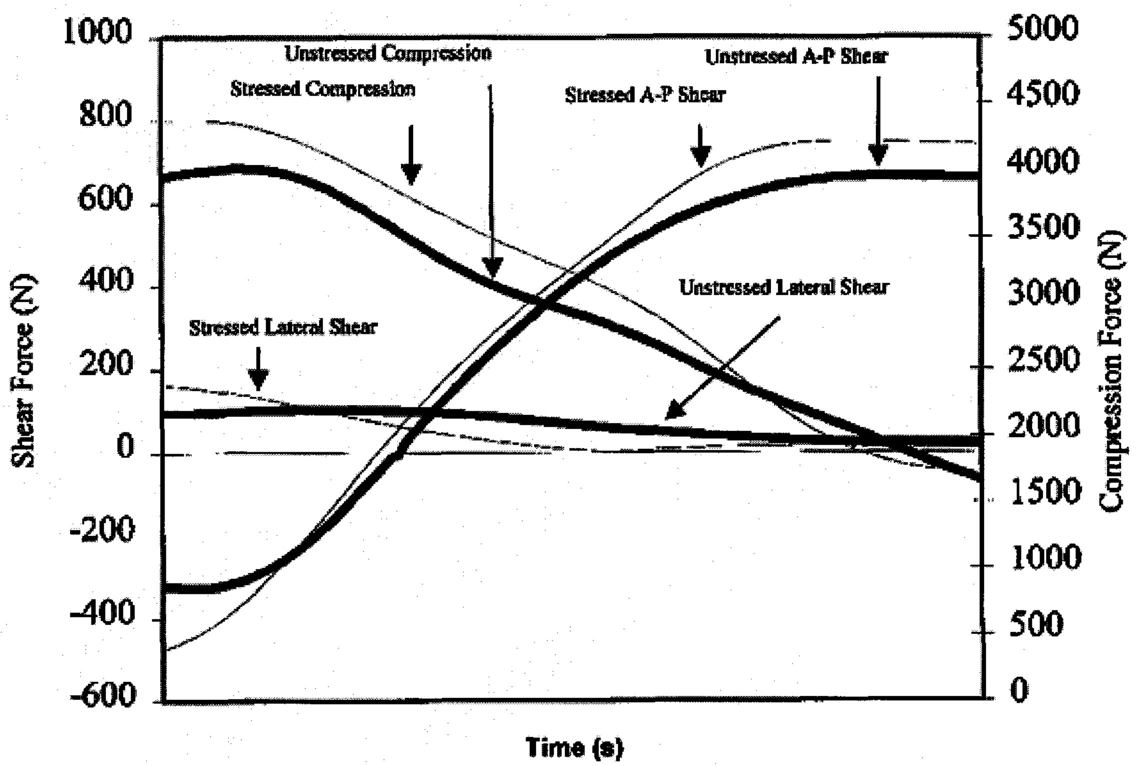

Figure 1.1 Alteration in spinal loads due to change in psychosocial factors (i.e. stressed vs. unstressed). Marras et al 2000

Despite the well-recognized role of lifting in low-back injuries (Andersson, 1981; Burdorf and Sorock, 1997; Damkot et al., 1984; Ferguson and Marras, 1997; Kuiper et al., 1999) the literature on safer lifting techniques remains controversial (Hsiang et al., 1997; van Dieen et al., 1999). In search of optimal lifting methods, squat lift (i.e., knee bent and back straight) is generally considered to be safer than the stoop lift (i.e., knee straight and back bent) in bringing the load closer to the body and, hence, reducing the extra demand on back muscles while counterbalancing the moments of external loads. The importance of the squat versus stoop lifting technique has, however, been downplayed due to the lack of a clear biomechanical rationale for the promotion of either style (Hsiang et al., 1997; McGill, 1997; van Dieen et al., 1999). Many workers, despite instruction to the contrary, prefer the stoop lift due to its easier operation, lower energy consumption in repetitive lifting tasks (Grag and Herrin, 1979; Hagen and Harms-Ringdahl, 1994) and better balance (Toussaint et al., 1997). Arm fatigue has been shown to result in changes in lifting technique (i.e. from squat to stoop), hence imposing larger loads on spine (Chen, 2000). Besides, it is known that squat lift is not always possible due to the lift set up and load size.

The advantages in preservation or flattening (i.e., flexing) of the lumbar lordosis during lifting tasks are even less understood. Lifting has been categorized as either squat or stoop with often no recording of changes in the lumbar lordosis which may influence the risk of injury (McGill et al., 2000; Potvin et al., 1991). The kyphotic lift (i.e., fully flexed lumbar spine) is recommended by some as it utilizes the passive posterior ligamentous system (i.e., posterior ligaments and lumbodorsal fascia) to their maximum thus relieving the active extensor muscles (Gracovetsky, 1988; Gracovetsky et al., 1981). In contrast, however, others advocate lordotic and straight-back postures indicating that posterior ligaments cannot effectively protect the spine and an increase in erector spinae activities is beneficial in increasing stability and reducing segmental shear forces (Delitto et al., 1987; Hart et al., 1987; Holmes et al., 1992; McGill, 1997; Vakos et al., 1994). Moderate flexion has been recommended by model (Arjmand and Shirazi-Adl, 2005a; Shirazi-Adl and Parnianpour, 1999, 2000) as well as experimental studies (Adams et al., 1994) to reduce risk of failure under high compressive forces. As the lumbar posture alters from a lordotic one to a kyphotic one, the effectiveness of erector spinae muscles in supporting the net moment (due to smaller lever arms (Jorgensen et al., 2003; Macintosh et al., 1993; Tveit et al., 1994) and the anterior shear force (due to changes in line of action (McGill et al., 2000) decreases while the passive contribution of both extensor muscles and the ligamentous spine increases (Arjmand and Shirazi-Adl, 2005a, 2006; Macintosh et al., 1993).

Compared to lifting, less attention has been paid to pushing and pulling. It has been estimated that nearly half of MMH tasks consists of pushing and pulling maneuvers (Baril-Gingras and Lortie, 1995; Kumar, 1995). Increased LBP has been found in those whose job involved reaching and pulling (Magora, 1973). NIOSH (1981) has reported that 20% of the injury claims for low back pain are associated with pulling and pushing. In the transport sector, fire fighting, nursing, construction work, and some specific tasks such as floor mopping and garden raking pushing and pulling are frequently performed (Hoozemans et al., 1998). Two type of hazards have been associated with these risk factors (Chaffin, 1987); first overexertion to musculoskeletal system, and second increased risk of accident due to slipping and tripping. Damkot et al (1984) calculated a pushing exposure index by multiplying the weight of pushed object with the number of pushing effort per day and showed that those with severe LBP have five times as much weight-day unit as those without LBP. Spinal loads have also been shown to increase in activities involving pulling (White and Panjabi, 1990). MMH tasks will even result in a higher risk of back injury, if they are associated with faster movement paces and awkward postures (Marras et al., 1995; Marras et al., 1993).

1.2.2.2 Motion, repetitive task and fatigue

The "heavy physical work" category in epidemiological studies is defined as the work that has high energy demands or requires some measure of strength (NIOSH, 1997). This, hence, includes heavy tiring tasks and heavy, dynamic, or intense work and is associated with more moderate risk of LBP than MMH and awkward postures (NIOSH, 1997). Manual material handling of lighter loads at higher velocities is more prevalent in our current industrial environments as compared to earlier un-automated work place. The risk of back injuries in workplace has been identified to significantly increase when tasks are performed at greater trunk velocities (Norman et al., 1998). Previous studies have generally indicated that faster trunk movements reduce trunk strength while imposing greater trunk moments, muscle activities/coactivities and, as a result, spinal loads (Davis and Marras, 2000). A more recent study also reported greater peak moments under faster lifting irrespective of the load and lifting technique (Kingma et al., 2001). In contrast, however, reverse trends (i.e., greater strength whereas smaller moments and muscle activity/coactivity) or no marked differences have also been observed in a number studies when evaluating the effect of movement velocity (see (Davis and Marras, 2000)).

Studies on motor recruitment pattern (Marras and Granata, 1997a; Parnianpour et al., 1988; Sparto and Parnianpour, 1998; Sparto and Parnianpour and Marras et al., 1997; Sparto and Parnianpour and Reinsel et al., 1997) suggest that repetitive lifting may indeed change the motor recruitment pattern due to fatigue and hence influence the loading pattern on spine. Marras et al (2006) found that spinal loads increased after 2 hours of lifting regardless of lifting frequency. They further showed that the greatest spinal loads occurred at frequencies and weight to which subjects are not used. Significant increase in EMG of paravertebral lumbar muscles to a sudden unexpected load applied to the upper trunk has been demonstrated due to fatigue during whole body vibration (Pope et al., 2002). On the other hand, threshold of tissue failure also decreases under repetitive or prolonged loading (Adams and Hutton, 1985; Brinckmann et al., 1989) resulting in further increase in risk of injury (Fig. 1.2).

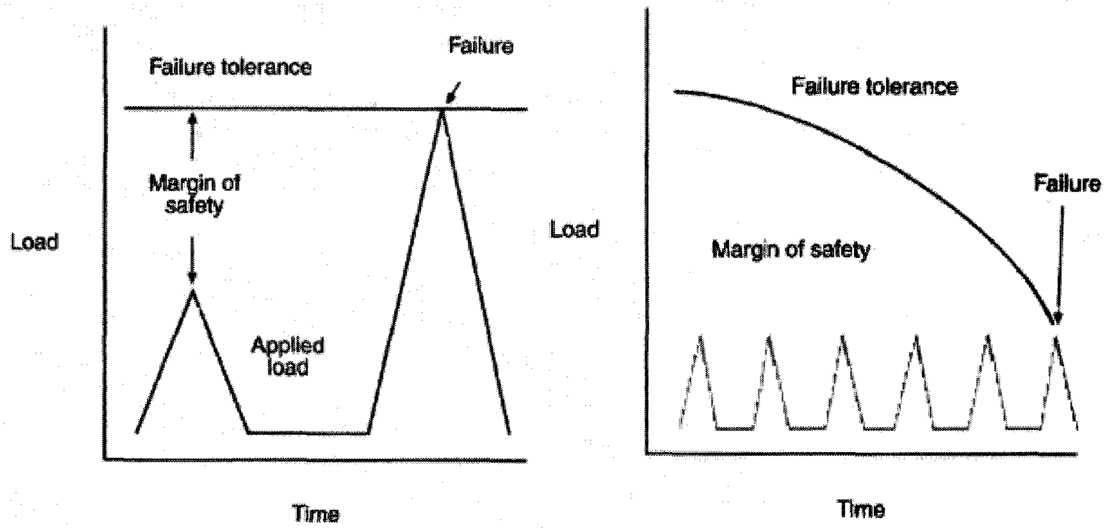

Figure 1.2 (Left) a margin of safety is observed in the first cycle of sub-failure load. In the second loading cycle, the applied load increases in magnitude, simultaneously decreasing the margin of safety to zero, at which point an injury occurs. (Right) repeated sub-failure loads lead to tissue fatigue, reducing the failure tolerance. (McGill, 2007)

1.2.2.3 Whole body vibration and shock

According to NIOSH (1997) WBV refers to mechanical energy oscillations which are transferred to the body as whole (in contrast to specific body regions), usually through a supporting system such as a seat or platform and has been suggested to have strong relation with LBP. Positive relation between LBP and WBV (Bovenzi and Hulshof, 1999; Hulshof and van Zanten, 1987) and/or sufficient reasons to reduce WBV (Lings and Leboeuf-Yde, 2000) has been demonstrated by extensive literature reviews. Those exposed to WBV are 1.4 (Boshuizen et al., 1992) to 9.5 (Bongers et al., 1990) times more prone to back pain than those without WBV exposure. Back injury due to WBV is affected by personal factors such as age (Fritz, 1999) and vertebral strength (Hinz et al., 1994). Moreover, vibration associated with awkward posture increases energy consumption (van Dieen, 1996) and muscular fatigue in the lumbar erector spinae (Hansson et al., 1991), hence, increases the likely risk of back injury. The primary source of vibration in the vehicle is the interaction of the vehicle with the ground surface. Truck driving, tractor driving and heavy equipment operation produce a vibration environment with frequencies from 3.5 to 8.9 with vertical accelerations of up to 2.6g (Frymoyer et al., 1980). Resonant frequencies of the spine occur between 4 and 6 Hz in the vertical direction and between 10 and 14 Hz when there is bending vibration of upper torso with respect to the lumbar spine (Goel et al., 2001; Pope et al., 2002). Experimental studies, when measuring driving point frequency response (impedance, apparent mass) and transmissibility functions between seat input and response at different spinal levels have shown that much of the dynamic response of the spine is due to the combined rotation and vertical compression of the pelvis-buttocks system (Pope et al., 1987). Review of literature showed that the duration of exposure had slightly stronger association with LBP than the magnitude of vibration (Lis et al., 2007). Vehicle seat vibrations with high acceleration content likely cause more back injury than those with low vibration levels that contribute more to the time averaged measures of exposure defined in ISO 2631 (Stayner, 2001). The direct causal association between whole body vibration and low back pain has, however, been questioned by some investigators (Gallais and Griffin, 2006; Lings and Leboeuf-Yde, 2000; Stayner, 2001) suggesting that, on the basis of existing literature, it is not possible to confirm whether whole body vibration exposure alone or in combination with other factors should be considered as a risk factor.

1.2.2.4 Awkward posture

Bending is flexion of the trunk, usually, in forward or lateral direction, and twisting refers to trunk rotation or torsion (NIOSH, 1997). Awkward postures include non-neutral trunk postures (related to bending and twisting) in extreme positions or extreme angles. Positive association of awkward posture and LBP has been reported in the literature (Frymoyer et al., 1980; NIOSH, 1997; Troup et al., 1981). The reported odd ratio of back disorder association with awkward posture ranges from 1.23 (Burdorf et al., 1991) to 8.09 (Punnett et al., 1991). Kelsey et al (1984) found that twisting without lifting had an odd ratio of 3.0 that increased to 3.1 in combination with lifting.

Marras et al (1993, 1995) found that the highest risk of back injury was related to lifting in combination with postural risk factors (e.g. lateral bending, twisting). Antagonistic EMG activity in the deep trunk muscles (Basmajian and De Luca, 1985) as well as higher intradiscal pressure (IDP) (Pope et al., 1986) has been found in axial rotation. Workers on excavating equipment have been reported to have their trunk flexed or twisted for at least 25% of the working time. Accordingly, a linear increase in prevalence of LBP among tractor drivers due to postural stress (i.e. awkward posture) has been suggested by Bovenzi and Betta (1994).

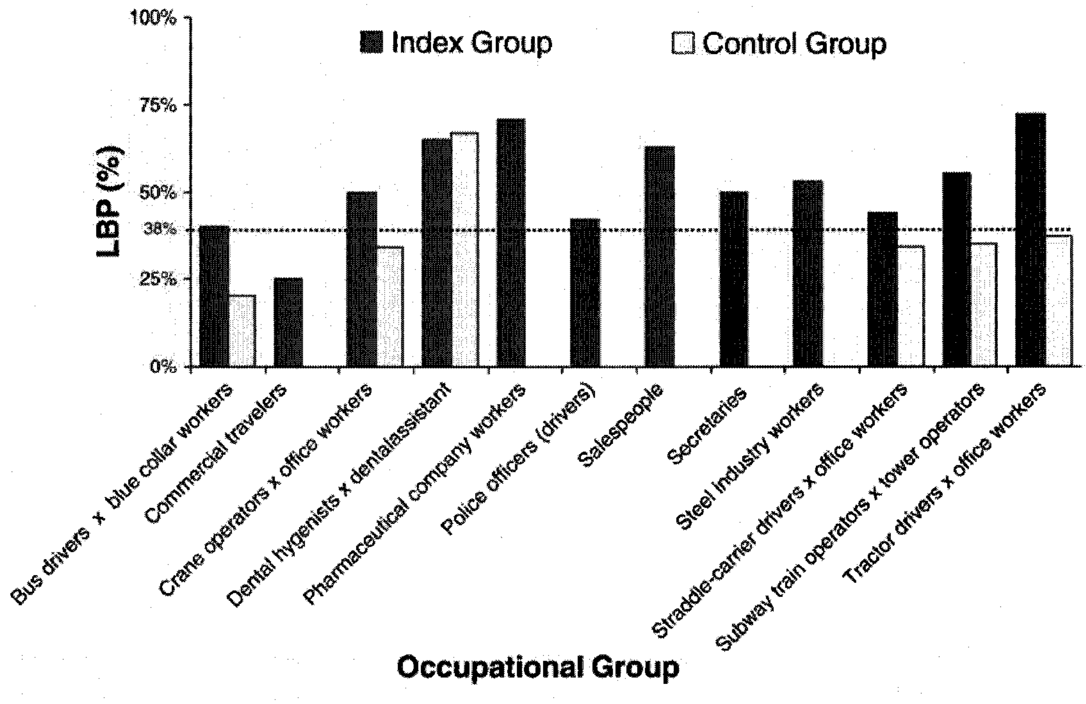

Figure 1.3 Annual prevalence of low back pain found in studies on occupations required to sit for more than half of work-time. (Lis et al., 2007)

1.2.2.5 Static work posture

Static work postures (sitting or standing) include isometric positions where very little movement occurs, along with cramped or inactive postures that cause static loading on the muscles (NIOSH, 1997). Three quarters of all workers in industrialized country have sedentary jobs that require sitting for long periods (Reinecke et al., 2002). There is not enough evidence in the literature to relate LBP with sedentary postures (Lis et al., 2007; NIOSH, 1997). Recent experimental studies showed that IDP can be lower in sitting than in erect posture (Wilke et al., 1999) in contrast to what had been demonstrated earlier (Nachemson and Elfstrom, 1970). Despite, lower risk of LBP in prolonged sitting as compared with other factors, it has been indicated that this group has the highest hospitalization rate for LBP. However, association of sitting with other factors (e.g. WBV and awkward postures) has been shown to increase the risk of LBP significantly (Fig. 1.3) (Lis et al., 2007).

1.3 Functional Biomechanics of the Spine

Detailed functional anatomy and biomechanics of the spine can be found elsewhere (Adams et al., 2006; Ashton-Miller and Schultz, 1997; Bogduk and Twomey, 1991; McGill, 2007), nevertheless for the benefit of readers a cursor description of trunk anatomy and biomechanics along with some pain generating mechanisms will be presented in this section. The trunk musculoskeletal system consists of the spine, rib cage, and pelvis covered with associated soft tissues (i.e. muscles, ligaments, and fascia). The spine is comprised of four regions (Fig. 1.4) containing a total of twenty four distinct vertebrae, separated by intervertebral discs; forming cervical, thoracic, and lumbar, as well as nine other fused vertebrae forming sacral-coccygeal. Each vertebra consists of an anterior structure known as vertebral body connected to a complex posterior structure by pedicles (Fig 1.5). The vertebrae transmit forces and moments and provide attachment sites for muscles. Each set of adjacent vertebrae and intervening disc (i.e. a motion segment), covered by seven intervertebral ligaments, and forms three joints that act to constrain spine motion. These include one symphyses joint, formed between two vertebral bodies and called "intervertebral symphyses", and two synovial joints, formed by the articulation of the superior articular process of one vertebra with the inferior atricular process of the vertebra above and called "zygopophysial" or "facet" joints.

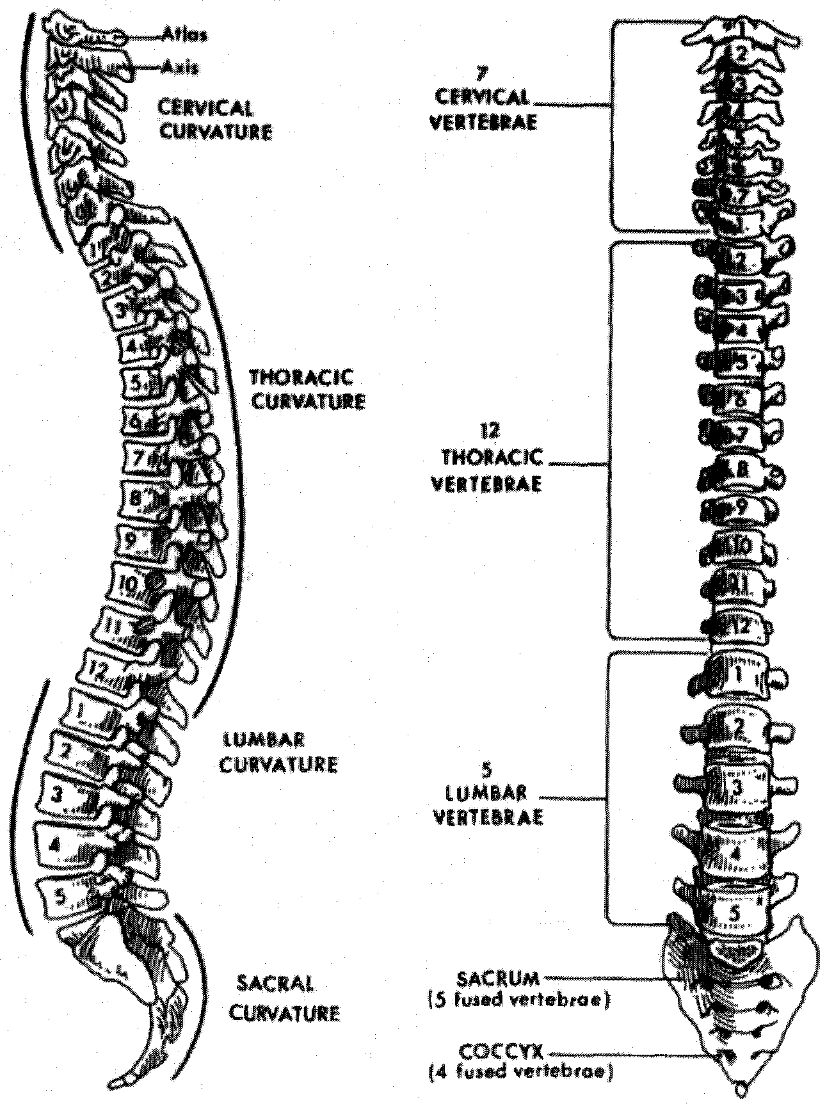

Figure 1.4 Frontal (right) and lateral (left) view of the human spine. (Ashton-Miller and Schultz, 1997)

Each intervertebral disc consists of two parts (Fig. 1.5); a central semi fluid mass of mucoid material called "nucleus pulposus", surrounded by a peripheral collagen fibers arranged in a highly ordered pattern (i.e. "anulus fibrosus"). Two thin layers of cartilage cover the top and bottom of disc that are called "vertebral endplates". During axial compression, both the annulus fibrosus and nucleus pulposus bear the load and transmit it from a vertebra to another. The annulus bulges radially while the end plates tend to bow towards the vertebral bodies. Provided the annulus is healthy and intact, increasing the axial load causes the fracture of end plates prior to any failure of the annulus fibrusus. It is to be noted that the resistance of an end plate to fracture depends on the strength of vertebral body attached to it. Annulus fibrosus is demonstrated to withstand pressures up to 32 MPa while cancellous bone fails at 3.4 MPa compression (Hickey and Hukins, 1980).

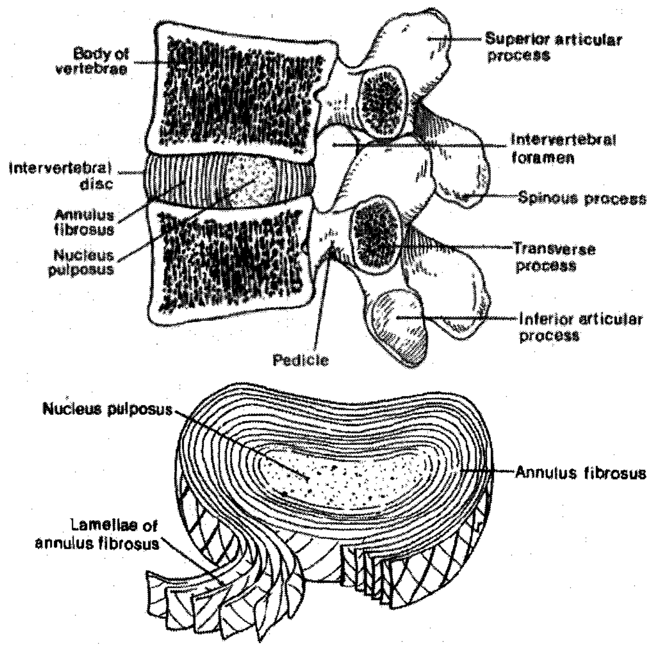

Figure 1.5 Lumbar spine motion segment (top), intervertebral disc (bottom). (Ashton-Miller and Schultz, 1997)

It has been shown that in erect sitting posture zygapophysial joints do not bear any part of the vertical load but in prolonged standing posture with lordotic spine they can bear up to 16% of the axial load (Adams and Hutton, 1980, 1981). Compressive strength of the spine vertebrae has been reported to be about 1.5 kN at C3, 2.0 kN at T1, 2.5 kN at T8, 3.7 T12, reaching 5.7 kN at L5 (White and Panjabi, 1990). There is, however, a considerable variation in ultimate compressive strength between individuals (i.e. personal factor); from 0.8 kN to 15.6 kN reported by Hutton et al (1979).

Furthermore, the fatigue life of vertebrae, indicated by the initial end plate failure, has been shown to be affected by the compressive load (Brinckmann et al., 1987). For instance, the probability of failure in vitro within 1000 cycle increases from 21 % at a loading range 30 to 40% of ultimate compressive strength to 84% at loading range of 60 to 70% of ultimate compressive strength. The human spine is not, however, as much stiff under tension as in compression (Markolf, 1972).

Trunk flexion involves both anterior rotation and anterior translation in spine motion segments. Anterior sagittal translation is mainly borne by direct contact of the articular facets of two adjacent vertebrae while the anterior sagittal rotation is resisted by the disc, zygapophysial joint capsules and posterior ligaments of intervertebral joints. Ultimate strength limit of 70 Nm has been reported for lumbar motion segments in flexion (Miller et al., 1986; Neumann et al., 1992; Osvalder et al., 1993). Experimental studies have shown that sectioning ligaments, zygapophysial joint capsules, and finally whole posterior elements, results respectively in 5, 9, 24 degree increase in flexion range (Bogduk and Twomey, 1991). McGlashen et al (1987) found that removal of the posterior elements resulted in 1.7 fold increase in shear translation, 2.1 fold increase in bending rotation, and a 2.7 fold in axial rotation in response to given loading conditions. Extension, on the other hand, generally involves reverse movements to those of flexion (i.e. posterior translation and rotation). Axial torsion of the spine involves a complex combination of forces and movement; one zygapophysial joint goes into compression while the other one is separated, and the intervertebral disc is strained by torsion and lateral shear. Ligaments and joint capsules were found not to resist considerable amount of torsion (Adams and Hutton, 1981) while the disc and the pressed zygapophysial joint resisted 35 and 65% of the twisting torque respectively (Farfan et al., 1970). Mircoscopic failure in annulus fibers starts as early as 3 degree of axial rotation and continues till the point of overt failure at about 12 degree.

It is plausible to assume that any structure in the spine that has nerve supply is potentially a source of pain. This includes zygapophysial joints, the ligaments of posterior elements, the para-vertebral muscles, the dura mater, the anterior and posterior longitudinal ligaments and the intervertebral discs. Experimentally, stimulation/anaesthetizing of nerve ending to these structures have been shown to cause/relieve back pain (Bogduk and Twomey, 1991). Back pains, caused by stimulation of nerve ending, is different to those due to nerve root compression and are referred as "Somatic Pain". This group of pain may be caused by stimulation of nerve ending due to chemical or mechanical irritation. Somatic pain generated in lumbar spine can also be perceived in the buttocks and lower limbs as well. On the other hand, a pain caused by compression of lumbosacral nerve roots due to spinal disorders is called "Radicular Pain" or "Sciatica". Lumbosacral arthritis and disc herniation may compress nerve roots. It has been shown that only the compression of a previously damaged nerve root can cause sciatica pain (Howe, 1979; Loeser, 1985). It is to be noted that sciatica pain is a well localized pain, radiating below ankle in contrast to perceived somatic pain in other tissues that is poorly localized (McCulloch and Waddell, 1980). On the basis of existing clinical ground only as few as 5% (only due to disc herniation) of back pain patient can be reliably diagnosed. Nevertheless, pathological studies relate a diverse spectrum of lumbar lesions to compressive and torsional injuries (Bogduk and Twomey, 1991). Hence, accurate determination of load distribution among passive and active components of human trunk in various activities is essential in effective prevention, evaluation and treatment of spinal disorders.

1.4 Estimation of Loads on the Spine

Infeasibility of direct measurement of muscle forces and spinal loads in human beings and the unsuitability in extrapolation of such data collected from quadrupled animal studies have led to indirect quantification of loads on spine by measuring representative biomedical indicators (e.g. intra-discal pressure, muscle electromyographic (EMG) activity, forces in spinal implants). However, apart from invasiveness, cost concerns, limitations and difficulties, the validity of such indicators to adequately represent spinal loads has been questioned (van Dieen et al., 1999). Biomechanical models have, thus, been recognized as indispensable tools for estimation of spinal loads during various occupational and recreational activities.

Spinal loads arise from external loads (including gravitational and inertial loads) and compensatory responses in trunk musculature (Fig. 1.6). To calculate unknown muscle forces, net joint moments (due to external loads) are initially determined using inverse dynamics either analytically for a system of rigid body segments or numerically using finite element methods for a system of deformable bodies. In the former approach, a full kinematics description, anthropometric measures of the body segments, external forces in addition to the base reaction force measured by force plates are employed to determine joint reaction moments in different planes (Kingma et al., 1996; Lariviere and Gagnon, 1998; Plamondon et al., 1996). In the latter one, under the same kinematics and external loads along with stiffness and dynamic properties of the ligamentous spine, time variation of net joint forces and moments are calculated. Both approaches take advantage of measurements by dynamometers/load cells/force transducers at different locations as well as captured kinematics data by 3-D camera systems (e.g., Optotrak).

Calculated net joint moment using either of aforementioned approaches should be balanced internally by trunk passive and active components. The equilibrium equations (static or dynamic) are used at a typical spinal joint (say L5-S1 disc level) to determine unknown internal forces (i.e. muscle forces and spinal loads). Such equilibrium problem is however redundant due to the larger number of unknowns (i.e. muscle forces) than existing equilibrium equations (Fig. 1.6). To resolve the redundancy in equilibrium equations, four different approaches (i.e. reduction, optimization, EMG-assisted, and kinematics-based method) have so far been introduced in modeling studies (Fig. 1.7).

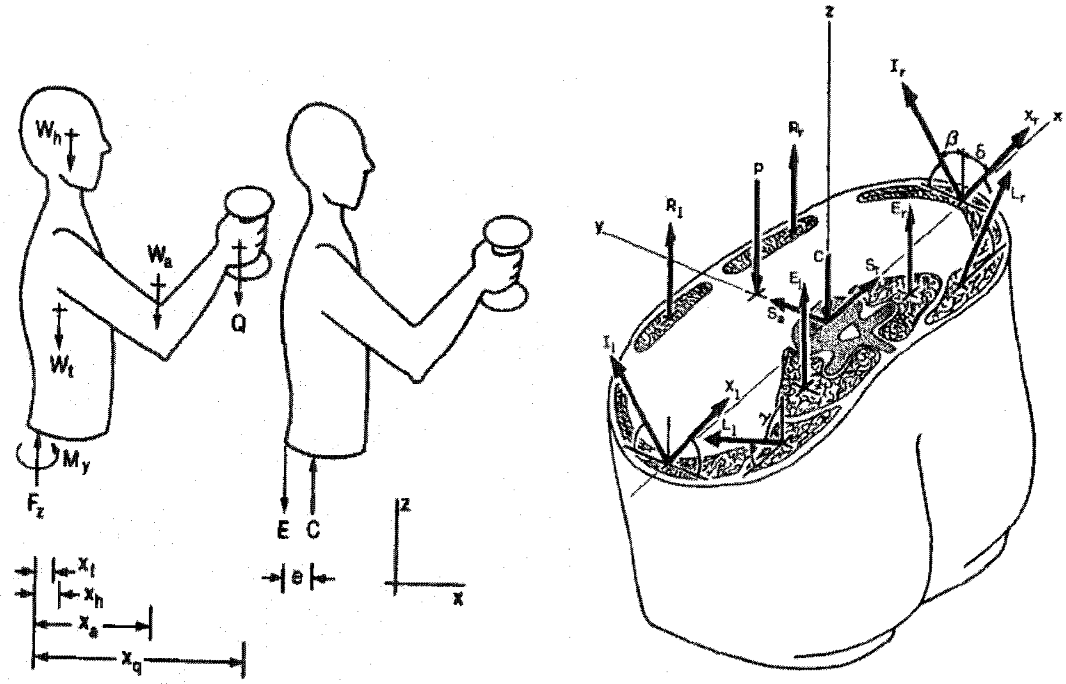

Figure 1.6 Free-body diagram for a simple static weight-holding task in the sagittal plane. (Left) Q is the weight held, W's are the weights of the various body segments, and the x's are the distance anterior to the intervertebral disc center. F_z and M_y are the components of the net reaction. (Middle) E is the contraction force in the back extensor muscles, and C is the compression at the disc level under consideration. E and C together must balance F_z and M_y. (Right) simplified schematic representation of a lumbar trunk cross-sectional model with ten unknown muscle forces. (Ashton-Miller and Schultz, 1997)

In the reduction method, muscles are grouped to form synergistic sets (i.e. having similar function) in order to reduce the number of unknown forces to the number of equations. Its simplest form in sagittal plane results in a single equivalent muscle model (either extensor or flexor) that has been suggested to be used for merely rough estimation of spinal compression forces (Bergmark, 1989; van Dieen and Kingma, 2005). Obviously, this method fails to estimate forces in individual trunk muscles including various fascicles of extensor back muscles each of which attach to a different spinal level and may play a different role during various activities (Bogduk et al., 1992)

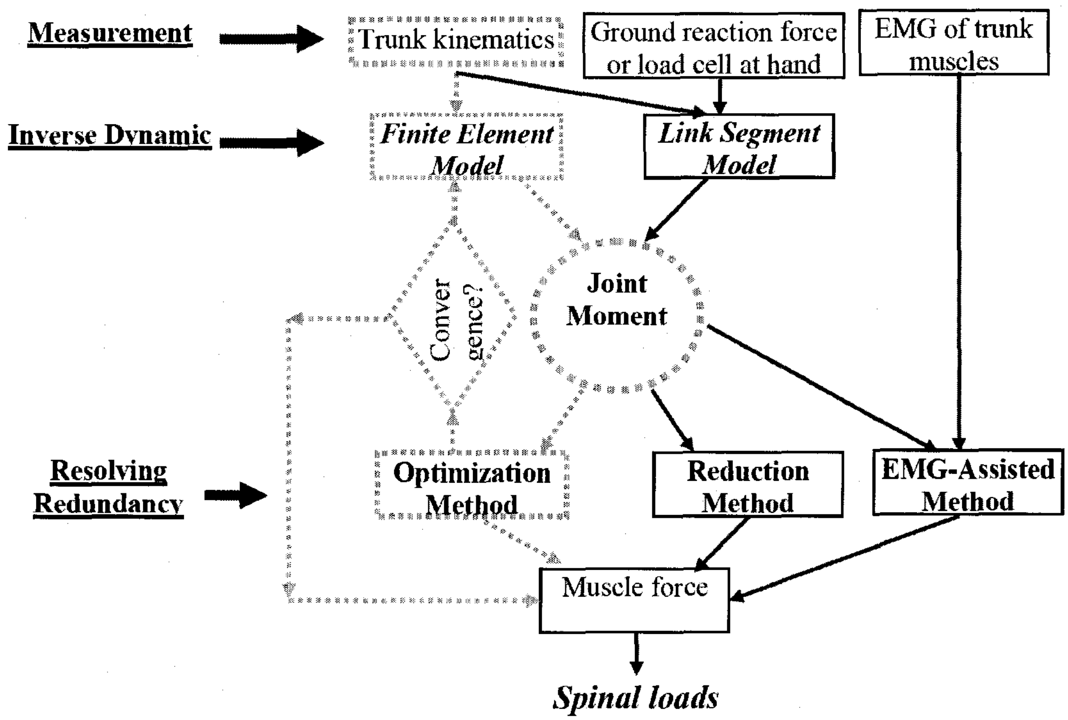

Figure 1.7 Different modeling approaches for calculation of spinal loads. Both link segment and finite element models use measurement data to calculate joint loads. The 3-D joint moment is, then, partitioned among trunk active-passive components using one (or a combination) of methods for resolving trunk redundancy. Flow of kinematics-based approach is depicted by dashed lines.

The premise behind the optimization method is that there may be a cost (objective) function that could be optimized by the central nervous system (CNS) while respecting some equality (i.e. equilibrium equations) and inequality (i.e. limits on muscle stresses) constrains. Various linear and nonlinear cost functions have been used (e.g. minimization of compression and shear forces or minimization of sum of muscle stresses to different powers) to partition the joint moment among active components

(Arjmand and Shirazi-Adl, 2005c; Challis, 1997; Crowninshield and Brand, 1981; Hughes, 2000; Parnianpour et al., 1997; Raikova and Prilutsky, 2001; Tsirakos et al., 1997). So long as optimization methods are subject only to equilibrium constrains, as reduction methods, they are generally unable to predict the co-activation in antagonist muscles. To circumvent this shortcoming, some optimization methods introduce non-zero positive lower bound activity in antagonistic muscles (El-Rich et al., 2004; Gardner-Morse and Stokes, 1998) that have been criticized to yield over-activation of these muscles over the time in dynamic tasks (Zeinali Davarani et al., 2007). Moreover, introduction of stability constraint within the optimization method has been demonstrated to automatically yield co-activity of antagonistic muscles (Stokes and Gardner-Morse, 2001; Zeinali Davarani et al., 2007). Arbitrary selection of cost functions as well as the deterministic nature of optimization method (being unable to predict inter and intra individual variabilities) are some shortcomings of this method. In the light of such criticisms of optimization approach, the use of processed Electromyography (EMG) signals from trunk muscles has been advocated by some investigators to drive "biological-based models" of the trunk (Gagnon et al., 2001; Marras, 2005; McGill, 2007; Sparto and Parnianpour, 1999). Despite the controversy regarding the nature of linear or nonlinear relations between muscle forces and EMG and the difficulty in relating the force and EMG under dynamic conditions, a number of EMG-driven models have been evolved. It is to be noted that calculated muscle forces by EMG method do not necessarily satisfy the equilibrium equation and need to be scaled using some gain factors to assure equilibrium at different planes. Surface EMG electrodes are not suitable for measuring the activity of deeper muscles. Deep muscles have been shown to be activated independently (Moseley et al., 2002), hence, challenging the prediction of deep muscle activity from those of surface trunk muscles. Maximum muscle stress and maximum voluntary contraction are both important parameters and have tremendous consequences in estimating muscle forces based on normalized EMG, but are not measured easily during maximum effort in cardinal planes.

Finally, in kinematics-based approach, kinematics data are prescribed in more than one spinal level and direction, hence, alleviating redundancy of the problem (Kiefer et al., 1997) while satisfying equilibrium equations at all levels and directions instead of only at one level that has been the case in most of earlier studies (Arjmand et al., 2007). This method will be elaborated more in next sections. Different combinations of aforementioned methods have also been developed and applied to exploit the strength of each method toward improvement in kinetic modeling of the spine. This includes a combined EMG-optimization method to avoid arbitrary scaling of calculated muscle forces at different planes (Cholewicki and McGill, 1994; Cholewicki et al., 1995; Gagnon et al., 2001). Others recruited optimization along with kinematics-based approach to resolve the remaining redundancy of equilibrium at various spinal levels (Arjmand and Shirazi-Adl, 2005a, 2006; El-Rich and Shirazi-Adl, 2005; El-Rich et al., 2004). Some models have considered only 4 to 11 pairs of trunk muscles (Granata and Marras, 1995; Schultz and Andersson, 1981) while other models incorporate up to 50 to 180 muscle fascicles (Bazrgari et al., 2007; Cholewicki and McGill, 1996; McGill and Norman, 1986; Stokes and Gardner-Morse, 1995).

1.5 Biodynamic Models of the Spine

Biodynamic models of the spine aiming to evaluate muscle forces and spinal loads can be divided into three distinct groups dealing with manual material handling, whole body vibration, and impact/fall. Further advancement in modeling of trunk kinematics requires a precise understanding of existing biodynamic models of spine.

1.5.1 Manual material handling

Both link segment and finite element methods have been used to evaluate dynamic spinal loads and trunk muscle forces during manual material handling tasks. Body parts are regarded as rigid element connected to each other through joints in link segment models. Their length, mass, and mass moment of inertia are extracted from anthropometric data bases available in the literature. Kinematics and force-plate data are used to estimate joints reaction moments and forces by writing instantaneous dynamic equilibrium equations at a lumbar level (usually at the L4-L5 or L5-S1 disc level) (Gagnon et al., 2001; Kingma et al., 1996; Lariviere and Gagnon, 1998) (Fig. 1.8). These models have been recently reviewed by Reeves and Cholewicki (2003) and are often used to study the effect of movement velocity (McGill, 2007; McGill, 1997) and lifting asymmetry on net spinal moments (Gagnon et al., 1993; Kingma et al., 1998; Plamondon et al., 1995). The validity of link segment results depends on accurate estimation of inputs like the acceleration of the segment, the location of joint center, location of external forces and torques, and anthropometric parameters (Lariviere and Gagnon, 1998). Usually, by comparison of computed and measured vertical ground reaction forces (Freivalds et al., 1984; Kingma et al., 1996; Kromodihardjo and Mital, 1987), computed and measured hand forces (Danz, 1991), or comparing the moment a certain joint from upward and downward analyses (Kingma et al., 1996; Lariviere and Gagnon, 1998; Plamondon et al., 1995) these models are validated. For instance, Plamondon et al (1995) validated their 3-D model by comparing joint moment at L5/S1 joint and showed a correlation of above 0.95 between predictions of lower body and upper body models. Significant sensitivity of the L5/S1 joint moment to measurement error (Lariviere and Gagnon, 1998) and trunk modeling (Lariviere and Gagnon, 1998) have been also reported.

Using top down inverse models, maximum net moment at L5/S1 was found to be significantly greater ($p<0.001$) in dynamic model as compared to static one for both level and sloped (facing uphill and downhill at 10 and 20 degrees) ground conditions (Menzer and Reiser, 2005). McGill and Norman (1985) showed that dynamic model resulted in peak L4/L5 moments 19% higher on average, with a maximum difference of 52% than those determined from static model while subject lifted a load of 18 kg from an extreme reach position of 83 cm. Lavender et al (2003), using a bottom up link segment model reported that the peak dynamic L5/S1 moments were significantly greater when lifting from lower lift heights ($p<0.001$), at faster lifting speed ($p<0.001$), and with heavier loads ($p<0.001$). While faster paces were generally associated with larger net spinal moments (Davis and Marras, 2000), certain pattern of motion were shown to result in inertial forces that can even reduce moments (McGill, 2007; McGill, 1997). Granata and Marras (1995) depicted that trunk moment decreased by increasing lifting velocity mainly because the subject pull the load close to their bodies earlier in excretion than during slower lifts.

To avoid difficulties in capturing kinematics data as well as mathematical complexities associated with link segment methods, Dolan and colleague developed a correlation to calculate spinal moment using only EMG activity of extensor muscles based on a priori knowledge of EMG-extensor moment relation (Dolan and Adams, 1993). While neglecting inertial forces; the method predicted higher spinal moment (up to 25.5%) than link segment model in fast movements (Dolan et al., 1999; Kingma et al., 2001). Other investigators developed special setups to directly calculate spinal moment using one or two force plates and a number of electro-goniometers (Fathallah et al., 1997; Granata et al., 1996). Some of these measurement setups, however, limit the movement due to constraining of the lower body. Granata et al (1996), for instance, while constraining the hip and legs, estimated the applied moments about the lumbo-sacral junction using only a force plate (Fig. 1.8). Consequently, their method was improved by Fathallah et al (1997) that avoided restraining of the hip and legs and calculated moment and loads at the L5/S1 joint using two electro-goniometers and a force plate during free dynamic lifting (Fig. 1.8). The latter method was able to predict the static joint moment with error of about 4% and suggested to provide reliable data under dynamic condition given the fact that in many activities, minimal inertial forces are generated by the lower limbs (Lindbeck and Arborelius, 1991; Plamondon et al., 1996).

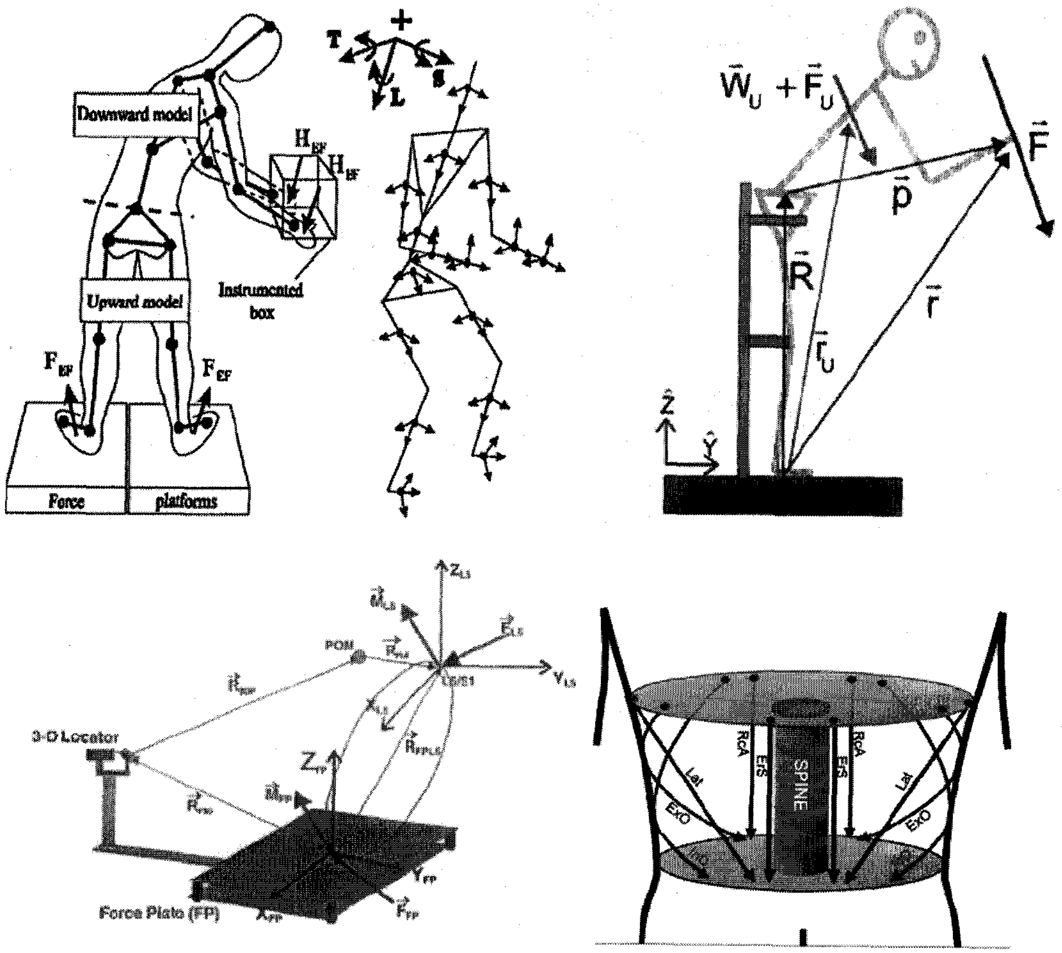

Figure 1.8 (Top-left) link segment model used by Lariviere et al 1998, (Top-right) setup used by Granata et al 1996, (Bottom-left) setup used by Fathallah et al 1997, (Bottom-right) muscle model of Marras et al 2004

After the evaluation of net moments at a spinal level, a biomechanical model is needed to partition the moments among the passive ligamentous spine and active musculature. Some investigators have used single equivalent muscle models with a fixed lever arm of about 5.5 to 6.5 cm and fixed or changing line of action (see (Reeves and Cholewicki, 2003) for references). It is reported that Morris et al (1961) were the first to introduce single muscle model and showed that most of axial loading on the spine is generated by muscles due to their smaller lever arm compared to external loads. Such simple models can provide a rough estimation of spinal loads and have been shown to be sensitive to assumption on muscle moment arm and line of action (Nussbaum et al., 1995; van Dieen and de Looze, 1999). Consideration of a detailed anatomy of trunk active-passive components on the other hand, however, makes the system indeterminate and calls for optimization or EMG assisted methods to resolve the system redundancy.

Marras and colleague have developed an EMG assisted model for partitioning of joint moment among trunk muscles. They demonstrated that spinal loading varied as a function of repetition (Granata et al., 1999; Mirka and Marras, 1993), forward bending (Granata and Marras, 1993, 1995; Marras and Sommerich, 1991), twisting motion (Granata and Marras, 1995), and lateral bending motion (Marras and Granata, 1997b). The strength of their approach in predicting inter and intra individual differences allowed them to assess the changes in spinal loads due to psychosocial factors including stress (Marras et al., 2000) and job complexity (Davis et al., 2002) as well as personal factors including back pain history (Marras et al., 2004) and gender (Marras et al., 2002). Ten pairs of trunk muscles were considered in the model (Fig. 1.8). While accounting for passive muscle force and force-length/velocity factors, the equilibrium was verified at only a single spinal level and the contribution of passive spine in balancing net external moment was neglected. Another biological driven model was developed by McGill and colleagues that consisted of an anatomically detailed three-dimensional representation of the skeleton, muscle, ligament, and nonlinear elastic intervertebral disc to partition net moment from link segment model among trunk active-passive components. While neglecting the translational stiffness of disc, their model was used to analyze spinal loads and stability under different dynamic tasks (Cholewicki and McGill, 1996; McGill, 2007).

Equally, optimization method has been used by investigators, though mostly in static condition, to distribute the net external joint moment among trunk restorative components (Crowninshield and Brand, 1981; Schultz et al., 1982; van Dieen, 1997).

Effect of different optimization criteria on spinal loads and muscle force calculation have also been studied for both single (Buchanan and Shreeve, 1996; Herzog and Leonard, 1991; Hughes, 2000; Hughes et al., 1994; Parnianpour et al., 1997) and multi joint (Arjmand and Shirazi-Adl, 2005c) models. Linear optimization (objective and constraint functions are both linear) has been found to have limited capabilities and require more physiologically constraint to provide acceptable muscle force predictions (Tsirakos et al., 1997). Nonlinear optimization, in contrast, has been suggested to result in more physiologically acceptable predictions (Tsirakos et al., 1997). Although effects of cost functions on spinal compression and shear was found to be statistically significant (Hughes, 2000; Parnianpour et al., 1997), it was shown that the difference in predicted peak spinal compression was only 1.1% between cubed muscle stress and double linear criteria (Hughes, 2000). Consideration of different cost function has been depicted to result in maximum variation of 20% in compression and 14% in shear force at lower lumbar level (Arjmand and Shirazi-Adl, 2005c).

In an effort to integrate the advantages of both EMG assisted method and optimization approach, Cholewicki and McGill (1994) developed a hybrid EMG assisted optimization method to minimize alterations in gain when considering equilibrium in different planes. The method has been suggested to provide more realistic and precise results than pure optimization or EMG assisted methods (Cholewicki and McGill, 1994; Gagnon et al., 2001).

The advantage of the finite element (FE) models over the single level free-body-diagram models rests primarily on the maintenance of equilibrium equations at different spinal levels. These models would also allow both for the direct association between kinematics and kinetics of the entire spine as well as the examination of the trunk stability. A linear FE model of spine and trunk musculature was developed by Stokes and Gardner-Morse (1995) to calculate muscle forces by maximizing moment at the T12 (i.e., maximum exertion task) in standing posture. Kinematics data were treated as extra unknown variables subject to some inequality constrains. Despite the novelty of this study in respecting equilibrium at all spinal levels and simulating a rather complex trunk musculature, it was not suitable for analyzing MMH tasks due mainly to the assumptions of unknown kinematics data and linearity of the model. Hence, the method was used to study the effects of maximum effort (Gardner-Morse et al., 1995; Stokes and Gardner-Morse, 1995), multi-criteria cost function optimization (Stokes and Gardner-Morse, 2001), abdominal co-activity (Gardner-Morse and Stokes, 1998), and axial loads (Stokes and Gardner-Morse, 2003) on trunk muscle recruitment and spinal stability in standing posture. Another detailed FE model developed earlier by Dietrich et al (1991) represented the spinal system consisting of the spinal column (vertebrae and discs), ligaments, muscles, ribcage, and pelvis. Horizontal movement of spine at the T3 was restrained and using an optimization function (minimizing muscle elastic potential energy) muscle forces and spinal loads were calculated. The model, however, was not used to analyze MMH tasks.

A kinematics-based approach was developed by Shirazi-Adl and colleagues that employed FEM strength in satisfying equilibrium at all spinal levels under prescribed kinematics and external loads while using a detailed muscular structure with nonlinear properties of spine segments (Kiefer et al., 1997, 1998). The method accounted for the full penalty of muscle forces through an iterative procedure and to our knowledge is the only FE model to analyze MMH tasks. Since its earlier development, the method has successfully been applied to study spinal loads, trunk muscle force and stability of the spine in isometric standing and flexed (i.e. sagittal forward rotations of 40 and 60 degrees) postures (Arjmand and Shirazi-Adl, 2006; El-Rich et al., 2004). Effects of load height (El-Rich and Shirazi-Adl, 2005), lumbar posture (Arjmand and Shirazi-Adl, 2005a; Shirazi-Adl et al., 2005; Shirazi-Adl et al., 2002), intra abdominal pressure (Arjmand and Shirazi-Adl, 2005b), and optimization cost function (Arjmand and Shirazi-Adl, 2005c) on aforementioned results have also been studied. The method, nevertheless, had been limited to static tasks with symmetry about the sagittal plane.

1.5.2 Whole body vibration

Measured driving point frequency response (impedance, apparent mass) and transmissibility functions between seat input and response at different spinal levels have been used to develop biodynamic models of the spine called either "phenomenological models" (Pankoke et al., 2001) or "input-output models" (Griffin, 2001) (Fig. 1.9). These models do not provide any other information than those they are based on and have no predictive power. Lumped-parameter models consisting of multiple lumped masses interconnected with springs and dampers are suitable for simulation of the seat-driver interaction and optimization of automobile seat suspension system. A chronological review of the lumped parameter models based on the motion direction and spring/damper characteristics (linear or nonlinear) has been made by Goel et al (2001). Physical models or seat-test dummies/anthropodynamic dummies form another type of "input-output models". In fact, these models are physical representations of lumped-parameter models. While lumped parameter models assume linear and frictionless components and can be restricted in uni-axial vibration, practical difficulties in engineering and fabrication of dummies hinder such idealization (Mansfield, 2005). The current applications of anthropodynamic dummies include crash testing and seat testing (Griffin, 2001).

The aforementioned models can predict the overall passive response under various vibration and postural conditions and may be used to improve vehicle suspension design. The muscle forces and spinal loads cannot, however, be estimated using such deterministic models. Another family of models called "mechanistic models" (Griffin, 2001) or "anatomical models" (Pankoke et al., 2001) have hence been developed with the objective to predict data that can not be measured directly (e.g. spinal loads and muscle forces). A two dimensional dynamic finite element model of the lower lumbar vertebrae was developed by Pankoke et al (1998). Under applied static (including a constant extensor muscle force that depended on the posture) and dynamic loads, spinal loads at lower lumbar levels were computed (Fig. 1.9). A head to sacrum finite element model including the entire spinal column and the rib cage was used by Kong and Goel (2003) to study the trunk resonant frequency and transmissibility under base vertical vibration (Fig. 1.9). The muscles were modeled deterministically as tension-only truss elements with a constant elastic modulus of 1.0 MPa. They reported the first vertical natural frequency in the range of 6.8-8.9 Hz depending on the muscle tension and gravity preload. Other 3D dynamic models have evaluated the trunk whole body vibration response under prescribed constant muscle forces that were estimated a-priori using a static analysis with an optimization approach (Buck and Wolfel, 1998; Pankoke et al., 2001).

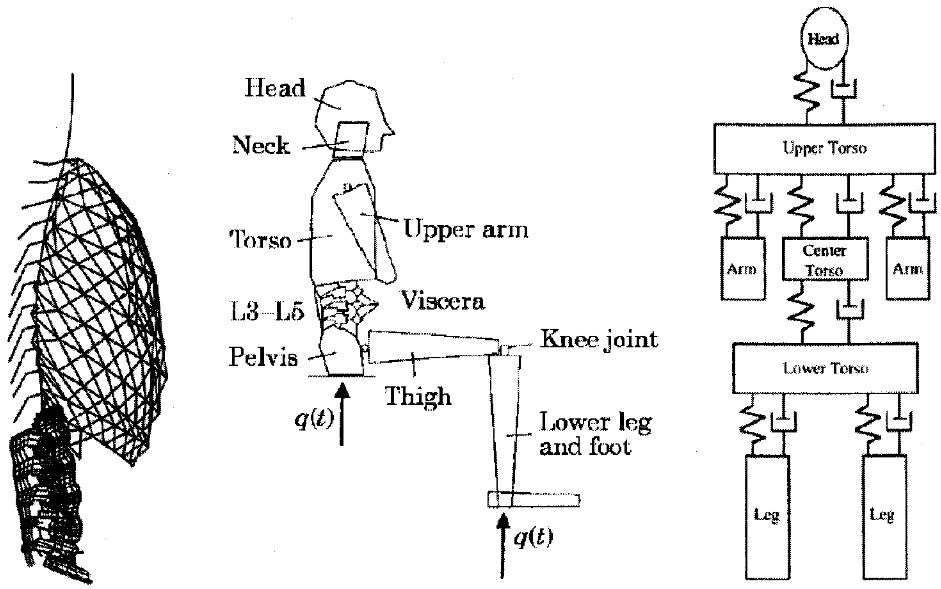

Figure 1.9 (Left) finite element model (Goel et al 2003), (Middle) an anatomic model (Pankoke et al 1998), (Right) a general lumped parameter model (Goel et al 2001)

Foregoing whole body vibration studies of the trunk have either neglected or over-simplified the redundancy in the trunk musculoskeletal system in which the spinal loads depend directly on unknown muscle forces that alter during the whole body vibration period. The importance of a proper estimation of muscle forces in quantification of spinal loads in whole body vibration conditions has been recognized in the literature (Bluthner et al., 2002; Seidel, 2005). The dynamic stability of the spine in whole body vibration studies has also been overlooked. The trunk stability is maintained by activation in muscles as well as passive muscle/spinal stiffness properties and is influenced by changes in the posture, passive properties and load magnitude/height (Arjmand and Shirazi-Adl, 2005a, 2006; Bazrgari and Shirazi Adl, 2007; El-Rich and Shirazi-Adl, 2005; Granata and Orishimo, 2001).

1.5.3 Impact

These models were mainly used to study automobile crash or pilot seat ejection conditions. The response of the body to such motions is much influenced by the inertial properties of the body rather than the stiffness and damping properties that influence the low frequency resonant responses. Discrete model constitute the main body of models dealing with impact problems. One such model for evaluating spinal reaction during car crash was developed by Roberts et al. (1969) and consisted of six rigid elements connected to each other through rotational and translational joints with linear stiffness values. A two dimensional discrete model accounting for axial, bending and shear stiffness of disc and inertial properties of the trunk was developed by Orne and Liu (1971) to analyze pilot ejection problem (i.e. the spine response to Gz acceleration). The model was further employed by McKenzi and Williams (1971) to study the whiplash problem (i.e. the spine response to Gx acceleration). Continuous parameter models (e.g. curved elastic beam-column) have also been developed (Cramer, 1976), with some simplifications to analytically resolve the equations (Fig. 1.10). Liu and Ray (1973) used finite element technique to study the wave propagation in spine. Later, they improved the model by simulating surrounding soft tissues with deformable elements rather than rigid ones and studied the response of spine to Gz acceleration (Liu and Ray, 1975). Despite the significant role of active-passive musculoskeletal system in the spinal response to impact, such contribution has been overlooked in these biomechanical models (Liu, 1982). Pontius et al (1972, 1975) took into account the muscles of the cervical spine (i.e. one flexor and one extensor) in their dynamic model of the whiplash injury, and showed that even with 10 g applied acceleration (i.e. Gx), the active role of muscles was quite manifest. They further showed that decreasing muscular strength as well as increasing the muscle reflex time (neural delay) would increase stress in boundaries of the cervical spine.

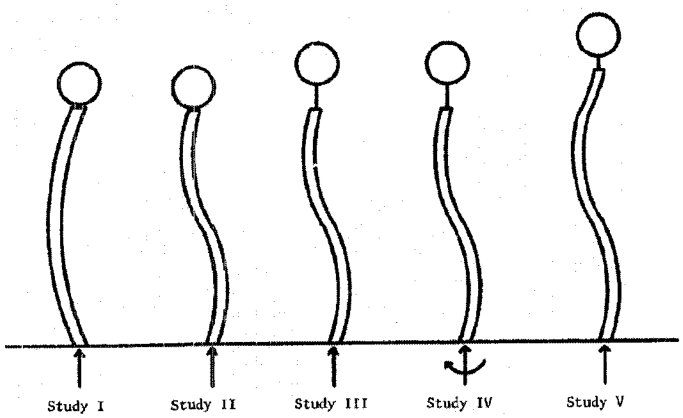

Figure 1.10 Continuous parameter model for the study of impact, (Cramer 1973)

1.5.4 General remarks on biodynamic models

Structural complexity, system nonlinearities, difficulties in measuring movement and forces in body segments, voluntary and involuntary muscular responses, and difficulties in obtaining empirical data on the properties of the biologic tissues are just some of hindrances towards the development of reliable spinal biodynamic models (Griffin, 2001). Since pathological studies attempt to relate mechanical loads to a wide spectrum of spinal lesions (Bogduk and Twomey, 1991), models that predict spinal loads more accurately can provide a better ground for prevention as well as clinical interventions. Link segment models merely provide an image of inertial and gravity loads at one specific spine section with no contribution from internal forces. A distribution model is needed at a subsequent step to calculate trunk muscle forces and spinal loads while accounting for passive spine forces. Such calculated muscle forces are very often based on the consideration of equilibrium at one single level that have been reported to violate the equilibrium at remaining levels (Arjmand et al., 2007). Due to geometrical irregularities, non-homogeneous material properties arrangement, large displacements under rather large loads and movements, the finite element method has been suggested to be the most suitable approach in spine modeling (Liu, 1982; Reeves and Cholewicki, 2003; Seidel, 2005; Shirazi-Adl and Parnianpour, 2001). The application of this method in MMH tasks is, however, scarce and only for static analyses. The other FE models used to study WBV and impact do not properly account for the nonlinear and direction dependent properties of motion segments as well as the redundancy in the trunk musculoskeletal system. Wrapping of back extensor muscles under movements involving moderate to large trunk flexions was often neglected or not properly simulated (Cholewicki and McGill, 1996). In upright (sitting or standing) positions that are often considered in whole body vibration tests, such consideration, however, does not influence predictions.

1.6 Spine Stability

Any biological system in equilibrium should function in a safe range of loading and motion with sufficient margin of stability. Spinal injury due to excessive stress and strain in tissue may be caused by an unstable or equivalently hypermobile configuration of the spine. Consideration of stability becomes hence as important as that of equilibrium especially under transient loads. The buckling loads of throcolumbar and lumbar spines in coronal plane have been shown to be about 20 N and 88 N, respectively (Crisco and Panjabi, 1991; Lucas and Bresler, 1961; Shirazi-Adl and Parnianpour, 1993, 1996a). Certain mechanisms, including the off-center placement of the line of gravity, pelvic tilt, and change in lumbar lordosis, have been shown to stabilize the passive system allowing it to carry larger compression loads with minimal displacement (Kiefer et al., 1996; Shirazi-Adl and Parnianpour, 1993, 1996a, b). Nevertheless, during daily activities the spine experiences much greater loads with no signs of instability (Arjmand and Shirazi-Adl, 2005a, 2006; El-Rich et al., 2004). Generally, three subsystems of trunk are suggested to provide the spinal stability; (1) the passive musculoskeletal subsystem including vertebrae, facet articulations, discs, spinal ligaments, and passive muscles, (2) the active musculoskeletal subsystem including muscles, and (3) the neural and feed-back subsystem (Panjabi, 1992). While muscles have been suggested to contribute to spinal stability via their passive, intrinsic, and reflexive responses (Moorhouse and Granata, 2007; Shadmehr and Arbib, 1992), it has been suggested that no single muscle can be accounted as the most important contributor to the spinal stability (Cholewicki and VanVliet, 2002).

Generally, to study stability of a system (either at rest or in motion), a perturbation is applied to the system and certain characteristics, called norm, are then assessed. If the response to perturbation does not exceed this defined measure, the unperturbed state is called stable (Leipholz, 1970). Clinical definition of spinal instability, on the other hand, is rather ambiguous and includes behavior as diverse as larger-than-normal segmental translations or rotations (i.e., hypermobility) (Frymoyer et al., 1990; Wilder et al., 1988), collapse of the isolated spine due to buckling (Lucas and Bresler, 1961), supine magnetic resonance scan images indicating a possible pscudarthrosis (Lang et al., 1990), and abrupt changes in the symptomatology of back pain patient (Kirkaldy-Willis and Farfan, 1982). White and Panjabi (1990) defined clinical instability of the spine "as the loss of the ability of the spine under physiologic loads to maintain its pattern of displacement so that there is no initial or additional neurological deficit, no major deformity, and no incapacitating pain."

Spine stability was first studied by Bergmark (1989) using a multiple muscle model with five rigid vertebrae, a rigid pelvis and a rigid thoracic cage. A detailed muscle model was considered consisting of both local (attaching to lumbar vertebrae and originating from pelvis) and global (attaching to rib cage and originating from pelvis) muscles. A linear force-stiffness relationship for muscles was assumed in which muscle stiffness was proportional to muscle force and inversely proportional to its length (i.e. $k=qF/L$) with q as a non-dimensional muscle stiffness coefficient taken equal for all muscles. He found that the spine became unstable for q-values smaller than 37 in upright standing posture. The importance of proper consideration of all trunk muscles in stability analyses was further examined by Crisco and Panjabi (1991) in their five-vertebra spine model. Gardner-Morse et al (1995), using a linear finite element model, found a critical q-value in range of 3.7-4.7 in standing posture for maximum trunk extension effort, suggesting that spine becomes more stable as muscle forces increase. They further showed that the stability of spine decreases as the passive stiffness decreases. With similar muscles force-stiffness relation, kinematics-based approach investigated system stability using both nonlinear and linear analyses in upright standing and flexed postures with and without loads in hands (Arjmand and Shirazi-Adl, 2006; El-Rich et al., 2004). The perturbation and buckling analyses were performed at the final stressed configuration of the trunk under loads to determine stability margins under given stiffness coefficients and critical muscle stiffness coefficient. Critical q-values of 75 (20) in upright standing without (with) load (El-Rich et al., 2004), 6 (0) and 10 (0) in 65 and 40 degree trunk flexion without (with) load (Arjmand and Shirazi-Adl, 2006) were predicted by this approach. Finally, recalculating data of earlier investigations (Bahler, 1967; Proske and Morgan, 1984; Rack and Westbury, 1974) Cholewicki and McGill (1995) reported the q-values in range of 36 to 60 for cat soleus, 61 to 170 for frog sartorious, and 80 to 106 for the rat garcilis and frog iliofibularis.

The stability of the spine was found to increase with trunk flexion angle (Arjmand and Shirazi-Adl, 2006; Cholewicki and McGill, 1996), external loads (Arjmand and Shirazi-Adl, 2006; El-Rich and Shirazi-Adl, 2005; El-Rich et al., 2004), coactivity of antagonistic muscles (Cholewicki et al., 1999; El-Rich et al., 2004), lumbar lordosis (Arjmand and Shirazi-Adl, 2005a), and intra abdominal pressure (Arjmand and Shirazi-Adl, 2005a; Cholewicki et al., 1999). Increasing the load height while keeping its moment arm fixed (i.e. diminished spinal stability), Granata and Orishimo (2001) showed that central nervous system responses to stability demands by increasing the extensor and flexor muscles activity. Such changes can not be predicted in optimization-based models unless a stability criterion is accounted in optimization routine. Recently, the introduction of this criterion in a dynamic lifting task was demonstrated to cause antagonistic co-activity in an optimization approach (Zeinali Davarani et al., 2007).

The study of Cholewicki and McGill (1996) was the first modeling study to evaluate spinal stability for a series of dynamic activity. At each instant of time the 18th root of the determinant of the Hessian matrix was taken as the stability index. It is to be noted that such an index only provides an insight about a stable system and requires an earlier confirmation of the positive lowest eigenvalue of the system (Gardner-Morse et al., 2006; Howarth et al., 2004). They found that stability increases with larger net moment or joint compression forces with the lowest level of stability associated with low muscle activity (i.e. standing upright). In contrast to other stability models which determine a critical modulation factor (i.e. dimensionless muscle stiffness coefficient) for the linear muscle stiffness (i.e. $k_{linear} = F/L$) below which the system becomes unstable, a cross bridge bond distribution moment model (DM) was used in their study to obtain muscle force and stiffness from measured EMG activities. The nonlinear force-stiffness relation in reflexive muscle activations has been demonstrated to noticeably enhance the trunk stability (see for example Shadmehr and Arbib 1992, Moorhouse and Granata 2007). In a recent study (Zeinali Davarani et al., 2007), such contribution of muscle spindle has been shown to decrease the error in positioning and velocity of trunk movement. They have further shown that the co-activation of antagonist muscle, in response to stability demands, decreased the response of muscle spindle to perturbation and suggested 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".

No model investigations of the spinal stability in dynamic tasks have examined the likely role of alterations in movement velocity, whole body vibration and impact. The trunk stability has been estimated empirically to deteriorate as the flexion-extension is performed at a faster pace (Granata and England, 2006). The pelvis was nevertheless restrained in this work that likely inversely influences the free trunk movement and behavior (Gupta, 2001). Moreover, the foregoing deterioration in system stability at faster movements appears out of line with the reported associated increase in muscle activities/coactivities (Davis and Marras, 2000) and the consequent increase in stability (El-Rich et al., 2004; Granata and Orishimo, 2001).

1.7 Concluding Remarks

Back pain is a significant socioeconomic burden on our aging society. Effective management of back pain rests on accurate estimation of trunk mechanical parameters (i.e. spinal loads (compression and shear), muscle forces, stress/strain in passive and active trunk components, and spinal stability). Practical difficulties and ethical considerations associated with invasive direct and indirect measurement methods along with limitations in indirect measurement methods support the use of biomechanical models. Existing biomechanical models of the spine have either neglected or oversimplified the nonlinear passive resistance, complex geometry/loading/dynamics of the spine, equilibrium in all spinal levels and directions, and the wrapping of extensor muscles. The Kinematics-based approach has been established as a powerful tool in earlier isometric studies of the human trunk during lifting tasks. The method should be extended to transient conditions in order to estimate muscle forces, spinal loads and trunk stability in dynamic tasks with non-negligible inertial effects.

1.8 Objectives and Thesis Organization

1.8.1 Development of dynamic kinematics-based approach

The kinematics-based approach was first introduced by Shirazi-Adl and his colleagues in 1997. Since then the method has been used to evaluate trunk muscle forces and spinal loads as well as spinal stability in isometric standing and flexed postures (Arjmand and Shirazi-Adl, 2006; El-Rich et al., 2004; Kiefer et al., 1997). Changes in trunk biomechanics due to changes in lumbar lordosis (Arjmand and Shirazi-Adl, 2005a; Shirazi-Adl et al., 2005; Shirazi-Adl et al., 2002), flexion angle (Arjmand and Shirazi-Adl, 2006), co-activity of antagonistic muscles (El-Rich and Shirazi-Adl, 2005), intra-abdominal pressure (Arjmand and Shirazi-Adl, 2005b), optimization cost function (Arjmand and Shirazi-Adl, 2005c), and wrapping of global extensor muscles (Arjmand et al., 2006) in static conditions were studied. In contrast to the static working postures, however, dynamic tasks at greater movement velocities involve a greater risk of injury (NIOSH, 1997). Hence, the principle objective of the current study was set to use the same notion and develop an iterative dynamic kinematics based approach in order to study trunk biomechanics under a series of sagittaly-symmetric dynamic tasks and conditions.

1.8.2 Lifting technique (stoop vs. squat)

Despite the well-recognized role of lifting in back injuries, the relative biomechanical merits of squat versus stoop lifting remains controversial. This is, however, due to the lack of a clear biomechanical rationale for promotion of either style. Such promotion should be based on accurate assessment of load distribution among active-passive trunk components and stability consideration. Hence, analysis of stoop vs. squat lifting technique was set as the first application of the dynamic kinematics-based method toward producing accurate information for the development of ergonomic guidelines for design of safer lifting tasks. In vivo kinematics measurements and model studies were combined to estimate trunk muscle forces and internal spinal loads under dynamic squat and stoop lifts with and without load in hands. Measurements were performed on healthy subjects to collect segmental rotations during lifts, needed as input data in subsequent model studies.

The following specific objectives were also considered and are addressed in chapters three and four:

1- To extend the Kinematics-based approach to dynamic conditions by accounting for the inertia and damping in the solution of the nonlinear transient equations of motion over the lifting period,
2- To perform in vivo measurements of trunk kinematics, needed as input data in subsequent model studies, on subjects performing sagittaly symmetric forward/backward lifting tasks with/without loads in squat and stoop techniques,
3- To evaluate and compare trunk muscle forces (active/passive components) as well as spinal loads at different levels under loading and kinematics considered in vivo
4- To determine, for the first time, spine stability margin under dynamic stoop and squat lifts
5- To assess the effects of the dynamic characteristics of the trunk (inertia and damping) and positioning of the external load on results.
6- To quantify the changes in internal spinal loads, muscle forces and stability margin due to alterations in passive stiffness of spine.
7- To quantify the effect of the curved paths of global muscles on predictions as compared with straight paths.

1.8.3 Trunk motion

Similar to lifting technique, effect of trunk motion on spine biomechanics has not been properly addressed in earlier investigations. Previous studies have generally indicated that faster trunk movements reduce trunk strength while imposing greater trunk moments, muscle activities/coactivities and, as a result, spinal loads. In contrast, reverse trends (i.e., greater strength whereas smaller moments and muscle activity/coactivity) or no marked differences have also been observed in a number studies when evaluating the effect of movement velocity. Changes in stability of spine due to alteration in trunk velocity have neither been studied by biomechanical models.

Moreover, the response of trunk extensor muscles to alteration in flexion/extension velocity at large flexion angles, where flexion relaxation happen in isometric condition for normal subjects, needed to be enlighten as well. Therefore, in vivo measurements of kinematics and ground reaction forces were carried out on young asymptomatic subjects while performing free flexion/extension at different paces. The collected kinematics of three subjects representing maximum, mean and minimum lumbar rotations was subsequently used in the Kinematics-driven model to compute results during the entire movements at three different velocities. Chapter five is presented to address the following specific objective:

1- To evaluate the effect of changes in velocity of movement and lumbar rotation during unconstrained flexion-extension tasks on muscle activations, spinal loads and stability.

1.8.4 Whole body vibration

Whole body vibration has been indicated as a risk factor in back disorders. Proper prevention and treatment managements, however, requires a sound knowledge of associated muscle forces and loads on spine. Previous trunk model studies have either neglected or over-simplified the trunk redundancy with time-varying unknown muscle forces. Trunk stability has neither been addressed. The iterative dynamic kinematics-driven approach was employed to evaluate muscle forces, spinal loads and system stability in a seated subject under a random vertical base excitation with ~ ±1 g acceleration shock contents. The input base excitation is taken from the literature and effect of posture, co-activity in abdominal muscles and changes in buttocks stiffness on spine biomechanics are investigated in chapter six.

1.9 List of Publications

The following articles have been submitted for publication during the course of my PhD study:

1. **Bazrgari B**, Shirazi-Adl A, Arjmand N. Biomechanical analysis of squat and stoop dynamic liftings - muscle forces and internal spinal loads. European Spine Journal, 2007; 16(5):687-99.
2. **Bazrgari B**, Shirazi Adl A. Spinal Stability and Role of Passive Stiffness in Dynamic Squat and Stoop Lifting. Computer Methods in Biomechanics and Biomedical Engineering 2007; 10 (5): 351-60
3. **Bazrgari B**, Shirazi Adl A., Trottier M, Mathieu P. Computation of trunk equilibrium and stability in free flexion-extension movements at different velocities. Journal of Biomechanics (2007- in press)
4. **Bazrgari B**, Shirazi-Adl A, Kasra M. Trunk muscle forces, spinal loads, and spinal stability under whole body vibration: A model study. Submitted to Journal Sound and Vibration (2007)

CHAPTER TWO

(Article-I)

ANALYSIS OF SQUAT AND STOOP DYNAMIC LIFTINGS - MUSCLE FORCES AND INTERNAL SPINAL LOADS

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2.1 Abstract

Despite the well-recognized role of lifting in back injuries, the relative biomechanical merits of squat versus stoop lifting remains controversial. In vivo kinematics measurements and model studies are combined to estimate trunk muscle forces and internal spinal loads under dynamic squat and stoop lifts with and without load in hands. Measurements were performed on healthy subjects to collect segmental rotations during lifts needed as input data in subsequent model studies. The model accounted for nonlinear properties of the ligamentous spine, wrapping of thoracic extensor muscles to take curved paths in flexion, and trunk dynamic characteristics (inertia and damping) while subject to measured kinematics and gravity/external loads. A dynamic Kinematics-driven approach was employed accounting for the spinal synergy by simultaneous consideration of passive structures and muscle forces under given posture and loads.

Results satisfied kinematics and dynamic equilibrium conditions at all levels and directions. Net moments, muscle forces at different levels, passive (muscle or ligamentous) forces and internal compression/shear forces were larger in stoop lifts than in squat ones. For the relatively slow lifting tasks performed in this study with the lowering and lifting phases each lasting ~2s, the effect of inertia and damping was not, in general, important. Moreover, posterior shift in the position of the external load in stoop lift reaching the same lever arm with respect to the S1 as that in squat lift did not influence the conclusion of this study on the merits of squat lifts over stoop ones. Results, for the tasks considered, advocate squat lifting over stoop lifting as the technique of choice in reducing net moments, muscle forces and internal spinal loads (i.e., moment, compression and shear force).

2.2 Introduction

Musculoskeletal impairments occur frequently and have a substantial impact on the health and quality of life of the population as well as on the health care resources. Search for a safer lifting technique has attracted considerable attention due to the high risk of injury and low-back pain (LBP) associated with frequent lifting in industry. Compression force limits have been recommended for safer manual material handling (MMH) maneuvers based on the premise that excessive compression loads could cause injury. Despite the well-recognized role of lifting in low-back injuries, (Andersson, 1981; Burdorf and Sorock, 1997; Damkot et al., 1984; Ferguson and Marras, 1997; Kuiper et al., 1999) the literature on safer lifting techniques remains controversial (Hsiang et al., 1997; van Dieen et al., 1999). In search of optimal lifting methods, squat lift (i.e., knee bent and back straight) is generally considered to be safer than the stoop lift (i.e., knee straight and back bent) in bringing the load closer to the body and, hence, reducing the extra demand on back muscles while counterbalancing the moments of external loads. The importance of the squat versus stoop lifting technique has, however, been downplayed due to the lack of a clear biomechanical rationale for the promotion of either style (Hsiang et al., 1997; McGill, 1997; van Dieen et al., 1999). Many workers, despite instruction to the contrary, prefer the stoop lift due to its easier operation, lower energy consumption in repetitive lifting tasks (Grag and Herrin, 1979; Hagen et al., 1993) and better balance (Toussaint et al., 1997). Besides, it is known that squat lift is not always possible due to the lift set up and load size.

The advantages in preservation or flattening (i.e., flexing) of the lumbar lordosis during lifting tasks are even less understood. Lifting has been categorized as either squat or stoop with often no recording of changes in the lumbar lordosis which may influence the risk of injury (McGill et al., 2000; Potvin and Norman et al., 1991). The kyphotic lift (i.e., fully flexed lumbar spine) is recommended by some as it utilizes the passive posterior ligamentous system (i.e., posterior ligaments and lumbodorsal fascia) to their maximum thus relieving the active extensor muscles (Gracovetsky, 1988; Gracovetsky et al., 1981). In contrast, however, others advocate lordotic and straight-back postures indicating that posterior ligaments cannot effectively protect the spine and an increase in erector spinae activities is beneficial in increasing stability and reducing segmental shear forces (Delitto et al., 1987; Hart et al., 1987; Holmes et al., 1992; McGill, 1997; Vakos et al., 1994). Moderate flexion has been recommended by model (Arjmand and Shirazi-Adl, 2005a; Shirazi-Adl and Parnianpour, 1999, 2000) as well as experimental studies (Adams et al., 1994) to reduce risk of failure under high compressive forces. As the lumbar posture alters from a lordotic one to a kyphotic one, the effectiveness of erector spinae muscles in supporting the net moment (due to smaller lever arms (Jorgensen et al., 2003; Macintosh et al., 1993; Tveit et al., 1994) and the anterior shear force (due to changes in line of action (McGill et al., 2000)) decreases while the passive contribution of both extensor muscles and the ligamentous spine increases (Arjmand and Shirazi-Adl, 2005a, 2006; Macintosh et al., 1993).

Evidently, an improved assessment of various lifting techniques and associated risk of tissue injuries depends directly on a more accurate estimation of the load partitioning in human trunk in dynamic lifting conditions. The spinal loads are influenced not only by the gravity, inertia and external loads but more importantly by trunk muscle forces (due to their smaller moment arms and their compensatory response to stability demands and tissue injuries). Despite conflicting data in the literature, previous studies generally indicated a decrease in trunk strength but an increase in trunk moments, muscle coactivity, muscle forces, and spinal loads as the trunk movement was performed at a faster rate (Davis and Marras, 2000; Dolan et al., 1994; Dolan et al., 1999; Granata and Marras, 1995a, b; Lariviere and Gagnon, 1998; McGill and Norman, 1985). A number of optimization and EMG driven models have been used to estimate the muscle forces in various lifting conditions. Many of these models do not properly account for the nonlinear passive resistance and/or complex geometry/loading/dynamics (i.e., inertia and damping) of the spine (Cholewicki and McGill, 1996; Dietrich et al., 1991; Gardner-Morse et al., 1995; Stokes and Gardner-Morse, 1995). In addition, many are simplified in not considering dynamic equilibrium equations simultaneously in all directions and at all levels.

Towards a more accurate estimation of muscle forces and spinal loads in lifting tasks, an iterative hybrid dynamic Kinematics-based finite element approach is introduced and applied in which a priori measured kinematics of the spine along with nonlinear passive properties are exploited. The Kinematics-based approach that results in a synergistic solution of the active-passive system has already been successfully applied to isometric conditions in upright (El-Rich et al., 2004; Shirazi-Adl et al., 2005) and flexed postures (Arjmand and Shirazi-Adl, 2006). The objectives of this work are, hence, set as follows:

1- To extend Kinematics-based approach to dynamic conditions by accounting for the inertia and damping in the solution of nonlinear transient equations of motion over the lifting period,

2- To perform in vivo measurements of trunk kinematics, needed as input data in subsequent model studies, on subjects performing sagittaly symmetric forward/backward lifting tasks with/without loads in squat and stoop techniques,

3- To evaluate muscle forces (active/passive components) as well as spinal loads at different levels under loading and kinematics considered in vivo while accounting for the curved path (i.e., wrapping) of global extensor muscles, and

4- To assess effects of dynamic characteristics of trunk (inertia and damping) and positioning of the external load on results.

2.3 Materials and Methods

2.3.1 In vivo measurement

Fifteen healthy men with no recent back complications volunteered for the study after signing an informed consent form approved by the Institute de réadaptation de

Montréal. Their mean (±S.D.) age, body height, and mass were 30±6 years, 177±7 cm, and 74±11 kg. While bending slightly forward, light-emitting diode, LED, markers were attached on the skin at the tip of the T1, T5, T10, T12, L1, L3, L5, and S1 spinous processes for evaluation of lumbar and torso flexions. Three extra LED markers were placed on the posterior-superior iliac spine and ilium (left/right iliac crests) for evaluation of pelvic rotation, and one on the load to track the position of weights in hands. A three-camera Optotrak system (NDI International, Waterloo, Ontario) was employed to collect 3D coordinates of LED markers.

Subjects were asked, with no instruction on lumbar posture, to perform sagittaly-symmetric squat (knee bent) and stoop (knee straight) lifts with and without 180 N weight placed on a bar in front at 20 cm height from the floor. Each task lasted 4-5s and started from upright standing with no load in hands and ended again in upright standing with or without load in hands. Two-way analyses of variance (ANOVA) for repeated measure factors were performed to study the effect of different lifting techniques (2 levels: stoop and squat) and load magnitude (2 levels: 0 N and 180 N) on collected kinematics data (Statistica, StaSoft, Tulsa, OK).

2.3.2 Thoracolumbar finite element model

A sagittally symmetric T1-S1 beam-rigid body model made of six deformable beams to represent T12-S1 discs and seven rigid elements to represent T1-T12 (as a single body) and lumbosacral (L1-S1) vertebrae was used (Arjmand and Shirazi-Adl, 2005a, 2006; El-Rich et al., 2004; Shirazi-Adl et al., 2005). The beams modeled the overall nonlinear stiffness of T12-S1 motion segments (i.e., vertebrae, disc, facets and ligaments) at different directions and levels. The nonlinear load-displacement response under single and combined loads along with flexion vs extension differences were represented in this model based on numerical and measured results of previous single- and multi-motion segments studies (Arjmand and Shirazi-Adl, 2005a; Oxland et al., 1992; Pop, 2001; Sadouk, 1998; Shirazi-Adl et al., 2002; Yamamoto et al., 1989). The trunk mass and mass moments of inertia were assigned at gravity centers at different levels along the spine based on published data for trunk segments (Pearsall, 1994; Pearsall et al., 1996) and head/arms (Zatsiorsky and Seluyanov, 1983). Connector elements parallel to deformable beams were added to account for the intersegmental damping using measured values (Kasra et al., 1992; Markolf, 1970); translational damping = 1200 N-s/m and angular damping = 1.2 N-m-s/rad. For the cases with 180 N external load in hands, the inertia of the load was computed using measured kinematics and subsequently added as an external load.

2.3.3 Prescribed postures

Mean measured sagittal rotations at thorax (evaluated by the change in inclination of the line attaching the T1 marker to the T12 one) and pelvic (evaluated by the orientation of normal to the plane passing through the three markers on the pelvis) of one subject were applied at the T12 and S1 vertebrae, respectively. The total lumbar rotation, calculated as the difference between the foregoing two rotations, was subsequently partitioned in between various segments based on values reported in earlier investigations (Anderson et al., 1985; Dvorak et al., 1991; Frobin et al., 1996; Pearcy et al., 1984; Plamondon et al., 1988; Shirazi-Adl and Parnianpour, 1999; Yamamoto et al., 1989). Relative proportions of ~7%, 12%, 15%, 22%, 27%, and 17% were used to partition the lumbar rotation between various motion segments from T12 to L5 levels, respectively.

2.3.4 Muscle model and muscle force calculation

A sagittally symmetric muscle architecture with 46 local (attached to the lumbar vertebrae) and 10 global (attached to the thoracic cage) muscles was used (Fig. 2.1) (Arjmand and Shirazi-Adl, 2005a, 2006; Bogduk et al., 1992; Stokes and Gardner-Morse, 1999). In order to accurately simulate the curved path of global muscles (i.e., longissimus thoracis pars thoracic and iliocostalis lumborum pars thoracic) at flexion angles considered in this study, these muscles were constrained to follow a curved path whenever their distances from T12-L5 vertebral centers at undeformed configuration diminished beyond ~10% (i.e., to reach the limit values of 53, 53, 55, 56, 54 and 48 mm for the global longissimus and 58, 56, 56, 55, 52 and 45 mm for the global iliocostalis at T12 to L5 vertebrae, respectively). This wrapping mechanism, similar to that formulated in our earlier works (Shirazi-Adl, 1989, 2006; Shirazi-Adl and Parnianpour, 2000), was considered in order not to allow the line of action of these muscles reach unrealistically close to the vertebrae resulting in erroneous small lever arms at different levels that occurs in larger flexions when global muscles are simulated as straight lines between their insertion points. During the analysis, when a wrapping is detected at a vertebral level, the contact force between the muscle and the corresponding vertebra is evaluated using the equilibrium equation in the instantaneous configuration assuming a frictionless contact that results in a constant muscle force along its entire length. This wrapping contact force was considered as an additional external force in subsequent iteration.

To evaluate muscle forces, Kinematics-based algorithm was employed to solve the redundant active-passive system subject to prescribed measured kinematics, inertia, damping and external loads. In this manner, calculated muscle forces at each instance of loading were compatible with the prescribed kinematics (i.e., posture) and loads while accounting for the realistic nonlinear stiffness of the passive system as well as trunk dynamic characteristics. Initially, the model calculated the moments at different levels required for the a priori prescribed rotations (i.e., measured posture). To resolve the redundancy problem at each level (i.e., in partitioning the calculated moment among muscles attached to that level), an optimization approach with the cost function of minimum sum of cubed muscle stresses was also needed along with inequality equations of unknown muscle forces remaining positive and greater than their passive force components (calculated based on muscle strain and a tension-length relationship (Davis et al., 2003)) but smaller than the sum of their respective maximum active forces (i.e., 0.6 MPa times muscle's physiological cross-sectional area, PCSA) (Winter, 2005) and the passive force components. Once muscle forces were calculated, the axial compression and horizontal shear penalties of these muscle forces along with wrapping contact loads (if needed) were fed back into the finite element model as additional updated external loads. This iterative approach was continued at each time instance till convergence was reached. The finite element program ABAQUS (V.6.5, 2004) was used to carry out nonlinear transient analyses while the optimization procedure was analytically solved using an in-house program based on Lagrange Multipliers Method (Raikova and Prilutsky, 2001). Implicit algorithm with unconditionally stable Hilber-Hughes-Taylor (1978) integration operator was used to solve the problem. The time step was automatically selected by the solver but was constrained to remain <0.01s.

2.3.5 Parametric studies

In order to determine the relative role of inertia on results, the lifting case with squat technique and 180 N in hands was reanalyzed with both trunk and external load inertias neglected (i.e., quasi-static analysis). To further investigate the effect of inertia, another quasi-static analysis of the same task was performed with the inertia of external load considered as an additional load. As for the effect of damping on results, additional dynamic analyses were performed assuming either a totally undamped system or one with 5-fold increase in damping simulating an over-damped condition.

Subjects carried the 180 N load further away (anteriorly from their S1 level) in stoop lift than in squat lift with the mean difference of <88 mm. In order to examine the effect of such load positioning on our predictions, the stoop lift was re-analyzed with the load position in the horizontal direction shifted closer to the body (i.e., the S1 level) to become identical to that in squat lift.

2.4 Results

Subjects carried stoop lifts, as compared with squat, with significantly larger thorax, pelvis, and lumbar (T12-S1) rotations ($p < 0.05$, 0.05 and 0.03, respectively). The magnitude of load (0 N versus 180 N), did not however have any significant effect on these rotations. The temporal variations of pelvic and thorax rotations of one typical subject measured under four different cases along with the intervening lumbar vertebral rotations were used as input data into the subsequent model studies. Polynomials of 6^{th} order were fitted on these rotations in order to smooth prescribed data (R^2 >98%, Fig. 2.2) into the model. Positions of the external load (180 N in hands) in the model studies of the squat and stoop lifts were based on the mean of measurements; the horizontal location of the load was nearly the same in both lifts when evaluated with respect to the T12 level whereas it was more anterior (by <88 mm) in stoop lifts when calculated with respect to the S1 for each subject.

The maximum net external moment at the L5-S1 disc substantially increased as 180 N load was carried in hands and as the lifting was performed in stoop technique (by ~28% compared to squat lifts) (Fig. 2.3). These moments were carried primarily by muscles with a small contribution (~10-30% depending on the lifting technique) from the passive ligamentous spine. At the T12 level and under both loading conditions, the moments resisted by the global extensor muscles and ligamentous spine were both larger in stoop lifts than in squat lifts (Fig. 2.4). The contribution of active muscle forces, especially in case with 180 N in hands, was greater than that of the passive muscle forces. The relative differences in global muscle forces in squat versus stoop lifts were due primarily to the smaller passive components in the former lifts (Fig. 2.4). Maximum muscle forces at different local and global levels (Fig. 2.5) were larger in stoop lifts than in squat lifts. Internal local compression and shear forces at different intervertebral disc levels were also greater in stoop lifts than in squat lifts with maximum differences reaching ~800 N in compression and ~200 N in shear at the L5-S1 level (Fig. 2.6 and Table 2.1). Calculated shear forces showed a dramatic increase from the L4-L5 level to the L5-S1 in all cases. Due to larger intersegmental lumbar rotations, passive segmental moments were also larger in stoop lifts (Table 2.1).

Except for the time periods at the beginning and end of tasks as well as immediately after lifting the external load, results were almost the same for both static and dynamic analyses over the entire duration of motion (Fig. 2.7). Inclusion of the inertia of the external load in the analysis was also found to have negligible effects on results. Increasing the damping at the motion segments did not change results, while considerable fluctuations in response (±10 N-m on required thorax moment) were noted in the absence of any damping in the model (Fig. 2.7). Closer positioning of the external load in stoop lift (by <88 mm in order to arrive at the same relative lever arm with respect to the S1 as that considered in squat lift) reduced the total muscle forces as well as net moment and internal compression/anterior shear forces at the L5-S1 level to values in between those predicted for squat and stoop lifts; i.e., remaining never-the-less greater than those for the squat lift.

2.5 Discussion

The controversy on a safer lifting technique persists due partly to the complexity of the problem (e.g., dependence on changes in the posture and load positioning) and over-simplifications (assumptions involving kinematics, constraints, geometry, material properties, loading, dynamic characteristics, etc.) in model studies. In this work, the Kinematics-based approach that has previously been applied to isometric lifting conditions was extended to predict muscle forces and internal spinal loads in dynamic stoop and squat lifts. For this purpose, parallel in vivo studies were performed to collect kinematics required as input data into the model. The entire forward-backward movements were carried out over 4-5s with either squat or stoop techniques but no instructions on the lumbar posture.

2.5.1 Methodological issues

Evaluation of the segmental rotations from skin markers is recognized to involve errors in identification of vertebral positions, skin movement relative to the underlying vertebrae, and deformability of vertebrae themselves (Cappozzo et al., 1996; Lee et al., 1995; Lundberg, 1996; Shirazi-Adl, 1994; Zhang and Xiong, 2003). Due to these inherent errors, the measurements were used to evaluate temporal variations of pelvic tilt and thorax rotation while the intervening lumbar segmental rotations were evaluated based on the partitioning of the difference between foregoing measured rotations using the relative values reported in the literature. The assumption of rigid body motion at the T1-T12 segments (upper torso) in the model was justified, in agreement with others (Nussbaum and Chaffin, 1996; Toussaint et al., 1995), by measuring nearly equal rotations for lines attaching either the markers T12 to T5 or markers T12 to T1. Changes in the relative proportions used to partition the total T12-S1 rotation among intervening segments would, as expected, alter to some extent the net moment, passive ligamentous resistance and muscle recruitments at these levels. Moreover, although these proportions were assumed constant during the entire lifting tasks, such may not necessarily be true in vivo as the relative demand at different levels could vary during lifting. These relative ratios were taken from data obtained in static measurements (Dvorak et al., 1991; Frobin et al., 1996; Pearcy et al., 1984; Plamondon et al., 1988) which have also been used in previous dynamic studies (McGill and Norman, 1986; Potvin and McGill et al., 1991; Potvin and Norman et al., 1991) in order to evaluate the contribution of passive tissue in offsetting external load. To prescribe measured rotations in the model, kinematics data of one typical subject rather than the mean of all subjects were considered. This was done due mainly to noticeable variations in duration of lowering/lifting phases in between subjects.

The transverse abdominal, latissimus dorsi, lumbodorsal fascia, intersegmental and multisegmental muscles were neglected, whereas the oblique abdominal muscles were presented by straight single lines rather than curved sheets of muscle. Consideration of several fascicles instead of just one for oblique muscles (EO and IO) has influenced the estimated spinal loads significantly in asymmetric lifting tasks but only slightly in symmetric ones (Davis and Mirka, 2000). Indirect effect of the transverse abdominal and latissimus dorsi muscles in unloading the spine through lumbodorsal fascia have been reported not be sizable during lifting tasks (Bogduk et al., 1998; Cholewicki et al., 1991; Macintosh et al., 1987; McGill and Norman, 1986; McGill et al., 1988; Tesh et al., 1987). Moreover, the likely mechanical effects of the intra-abdominal pressure (IAP), neglected in this study, have been found to depend on the posture and the co-activity level of abdominal muscles (Arjmand and Shirazi-Adl, 2005b). While local muscles were modeled as straight lines between their respective insertion points, realistic muscle paths were considered for global extensor muscles by wrapping them over all T12-S1 vertebrae whenever in the course of lifts their distance to associated vertebral bodies reduced more than 10% of their initial distances. This allowed for a maximum of ~10% reduction in muscle lever arms at different levels during flexion which was chosen in accordance with published data in the literature (Jorgensen et al., 2003; Macintosh et al., 1993; Tveit et al., 1994). The wrapping of global muscles occurred at all levels under larger flexion angles and resulted in curved paths with realistic lever arms at different levels. Had straight lines been assumed for global muscles, much smaller lever arms would have been generated resulting in greater muscle forces and internal loads. The wrapping contact forces (Table 2.2) remained relatively small compared with muscle forces suggesting minor changes in lines of action at wrapping points.

In presence of nonlinearity in equations, numerical integration using an unconditionally stable implicit method was employed in this study. Minimum sum of cubed muscle stresses, as the cost function used in the optimization, has been recognized to agree better with EMG data (Arjmand and Shirazi-Adl, 2005c; Hughes et al., 1994; van Dieen, 1997). The convergence of the nonlinear optimization solution on a global minimum was assured in this study by solving the optimization problem analytically. For the sake of comparison with EMG measurements of earlier studies, the computed muscle forces were partitioned, at the post-processing phase of the analysis, into passive and active components using a passive tension-length relationship for all muscles (Davis et al., 2003). Moreover, the maximum allowable muscle stress of 0.6 MPa was assumed for all muscles neglecting the effect of activation level on this value. It is important to emphasize that the passive load-length relationship considered for muscles in the current study have absolutely no bearing at all on the predicted spinal loads and total muscle forces. The rate-dependent viscoelastic properties of the spinal segments, which could play a role at much higher loading rates, (Neumann et al., 1994; Wang et al., 1998, 2000) were not considered in this study. Finally, in accordance with parallel in vivo measurements, the response was limited to the sagittal plane, thus neglecting out of plane motions.

2.5.2 Effect of dynamic parameters

Generally, faster trunk movements have been associated with a decrease in trunk strength but increases in trunk moments, muscle coactivity, muscle forces, and spinal loads (Davis and Marras, 2000; Dolan et al., 1994; Dolan et al., 1999; Granata and Marras, 1995a, b; Lariviere and Gagnon, 1998; McGill and Norman, 1985). Inertia effects of the trunk and external load have been indicated to play a noticeable role at the onset of a lift with jerky movements (Lariviere and Gagnon, 1998). Our results showed a negligible effect of inertia forces on trunk moment and spinal loads except at three time intervals; the beginning and end of the tasks as well as a short period after picking the load up (Fig. 2.7) which agrees with earlier observations (Holmes et al., 1992). Apart from these periods, a quasi-static analysis would yield sufficiently accurate results with no real need to account for inertia forces which could be due to the slow lifting performed by our subjects (i.e., lowering and lifting periods each lasting ~2s). Our results also demonstrated that the inertia of the trunk, and not that of the load, was the major factor for the observed differences in these three time periods. In a different lifting condition, however, the latter has been estimated to be more important than the former (McGill and Norman, 1985). The computed net moment at the L5-S1 is noted to be in good agreement with values reported in previous dynamic studies (Davis et al., 1998; Fathallah et al., 1998; Kingma et al., 2001; Toussaint et al., 1995).

Although recognized as an important parameter, damping has been neglected in earlier biomechanical model studies of dynamic lifting (Kingma et al., 2001). A five-fold increase in the segmental damping value which was used in the model based on earlier measurements did not markedly alter predictions of this work, especially away from the three time intervals indicated earlier (Fig. 2.7). Introduction of damping appeared to primarily smooth the temporal response by removing high frequency fluctuations (i.e., noise).

2.5.3 Effect of lifting techniques

The relative lumbar/pelvic rotations during lowering/lifting phases showed greater contributions in all cases from the pelvis than the lumbar spine (by as much as two-fold) and remained within the range of data reported in the literature (Esola et al., 1996; Granata and Sanford, 2000; McClure et al., 1997; Porter and Wilkinson, 1997). Thorax and pelvis rotations were both larger in stoop lifts compared to those in squat lifts (Fig. 2.2) resulting in greater lumbar (T12-S1) rotations in stoop lifts by 10.5° and 5.9° in cases with and without 180 N load in hands, respectively. These additional flattenings of the lumbar spine in stoop lifts increased the wrapping contact forces (Table 2.2) and moment-carrying contribution of passive ligamentous spine and trunk muscles. Moreover, despite identical lever arms considered (based on measurements) for the external load of 180 N at the T12 level, the net moments and hence muscle forces and internal loads were all greater in stoop lifts than in squat ones; e.g., maximum net moments of 200 N-m and 160 N-m were predicted at the L5-S1 level for stoop and squat lifting, respectively. Same trends were also found in the absence of external loads or even when the external load was shifted by <88 mm closer to the body in the stoop lift in order to reach the same lever arm with respect to the S1 as that considered in the squat lift.

Therefore, results of this study appear to suggest the squat lift as the safer lifting technique in reducing the net moment, muscle forces and internal ligamentous loads at all levels. It should be emphasized that the relative merits of these lifting techniques depend not only on the relative rotations at the thorax, pelvis and lumbar spine but also on other factors such as position of external loads, voluntary alterations in the lumbar curvature and speed of movement. These could partly be the reason why the literature remains yet inconclusive as some report smaller net moment and trunk load in squat lifting (Buseck et al., 1988; de Looze et al., 1994; Hagen and Harms-Ringdahl, 1994; Potvin and McGill et al., 1991) while others indicate otherwise (de Looze et al., 1998; Dolan et al., 1999; Lindbeck and Arborelius, 1991; Troup et al., 1983; van Dieen et al., 1994). The reduction in net moment in squat lifts, under all cases with and without external load, is due primarily to smaller pelvic and lumbar (and hence thorax) rotations in this technique resulting in much reduced net moments from the mass of the upper body and the external load about the L5-S1 (Fig. 2.8). Variations in location of external loads and rotations of pelvis and lumbar spine from a lift to another, as expected in different studies, are important and could substantially influence the results and subsequent comparison of lifting techniques towards identification of the optimal one. The biomechanical advantages for the squat lifts in our study would become even more apparent had a smaller lever arm for the external load been considered in these lifts (Bendix and Eid, 1983; Troup et al., 1983). In an earlier combined in vivo-model study on the effect of changes in the lumbar curvature on trunk response in isometric lifts with identical thorax rotations (Arjmand and Shirazi-Adl, 2005a), the maximum segmental shear/compression forces and activity in extensor muscles occurred in the lordotic posture while the maximum segmental flexion moment occurred in the kyphotic posture. The kyphotic postures exploited primarily the passive ligamentous/muscle force components while the active muscle forces played more important role in lordotic postures. The study advocated the free style posture or a posture with moderate flexion as the posture of choice in static lifting tasks when considering both internal spinal loads and active/passive muscle forces. One must note that in that study the thorax rotation remained nearly the same irrespective of changes in the lumbar curvature. In the current study, however, the thorax rotations of 66.9° and 70° in stoop lifts respectively with and without 180 N in hands were much greater than corresponding rotations of 38.4° and 49.7° in squat lifts (Fig. 2.8). Although we did not investigate the effect of changes in lumbar curvature in dynamic stoop and/or squat lifts, the conclusions of the previous isometric study advocating a flattened lumbar spine and current dynamic one advocating a squat lift (involving more lordotic lumbar curvature) do not contradict each other due to the crucial effect of posture (i.e., thorax and pelvic rotations) on results. Earlier studies on the effect of posture in lifting have suggested a lordotic posture in increasing the extensor activity during the early phases of the lift (Delitto et al., 1987; Hart et al., 1987; Vakos et al., 1994).

Results of previous works on extensor muscle activities in stoop lifts usually demonstrate two peaks; the first and smaller one occurring in lowering phase while the second and larger one in lifting phase of the tasks (Haig et al., 1993; Lariviere et al., 2000; McGill et al., 1999; Paquet et al., 1994; Peach et al., 1998). Our predictions on active extensor muscle forces also show similar variations during the tasks (Fig. 2.4). Due to the relatively small flattening of the lumbar spine (T12-S1) considered in the model (remaining <26°), no flexion relaxation was observed which would otherwise have influenced the results in the final periods of the lowering phase of the study.

In conclusion, the current work while accounting for nonlinear properties of the ligamentous spine, wrapping of global extensor muscles, trunk dynamic characteristics (inertia and damping) and in vivo measured postures, calculated muscle forces and internal spinal loads during squat and stoop lifts using a novel dynamic Kinematics-based approach. The model accounted for the spinal synergy by simultaneous consideration of passive ligamentous structure and muscle forces under given posture and loads. The predictions, therefore, satisfied kinematics and dynamic equilibrium conditions at all levels and directions. Results, for the tasks considered, advocate squat lifting over stoop lifting as the technique of choice in reducing net moments, muscle forces and internal spinal loads. These values remained greater, though to a lesser extent, even when the lever arm of the external load in stoop lift was reduced to become equal to that in squat lift. Furthermore, for the relatively slow lifts performed and modeled in this work, dynamic characteristics of trunk did not demonstrate significant effects on results.

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Table 2.1 Maximum internal loads* in passive ligamentous spine for different cases at various levels; passive segmental moment, M (N-m), local compression force, C (N), and local anterior shear force, S (N).

Disc Level	0 N						180 N					
	Stoop			Squat			Stoop			Squat		
	M	C	S	M	C	S	M	C	S	M	C	S
T12-L1	20	926	226	17	902	187	18	2416	384	14	2315	222
L1-L2	24	1155	244	19	1121	196	21	2921	381	14	2660	192
L2-L3	24	1445	184	18	1374	135	19	3383	244	12	2922	85
L3-L4	20	1793	300	14	1675	249	15	3903	536	7	3274	376
L4-L5	20	2162	258	14	1989	227	19	4518	502	9	3704	425
L5-S1	23	2355	800	16	2159	737	33	4831	1635	16	4023	1416

* occurring nearly at the time of maximum trunk flexion

Table 2.2 Maximum wrapping contact forces* for different cases at various levels (N).

Vertebra Level	0 N		180 N	
	Stoop	Squat	Stoop	Squat
T12	0	0	0	0
L1	40	27	60	32
L2	55	39	113	66
L3	62	43	97	44
L4	80	55	168	89
L5	99	70	251	138

* occurring nearly at the time of maximum trunk flexion

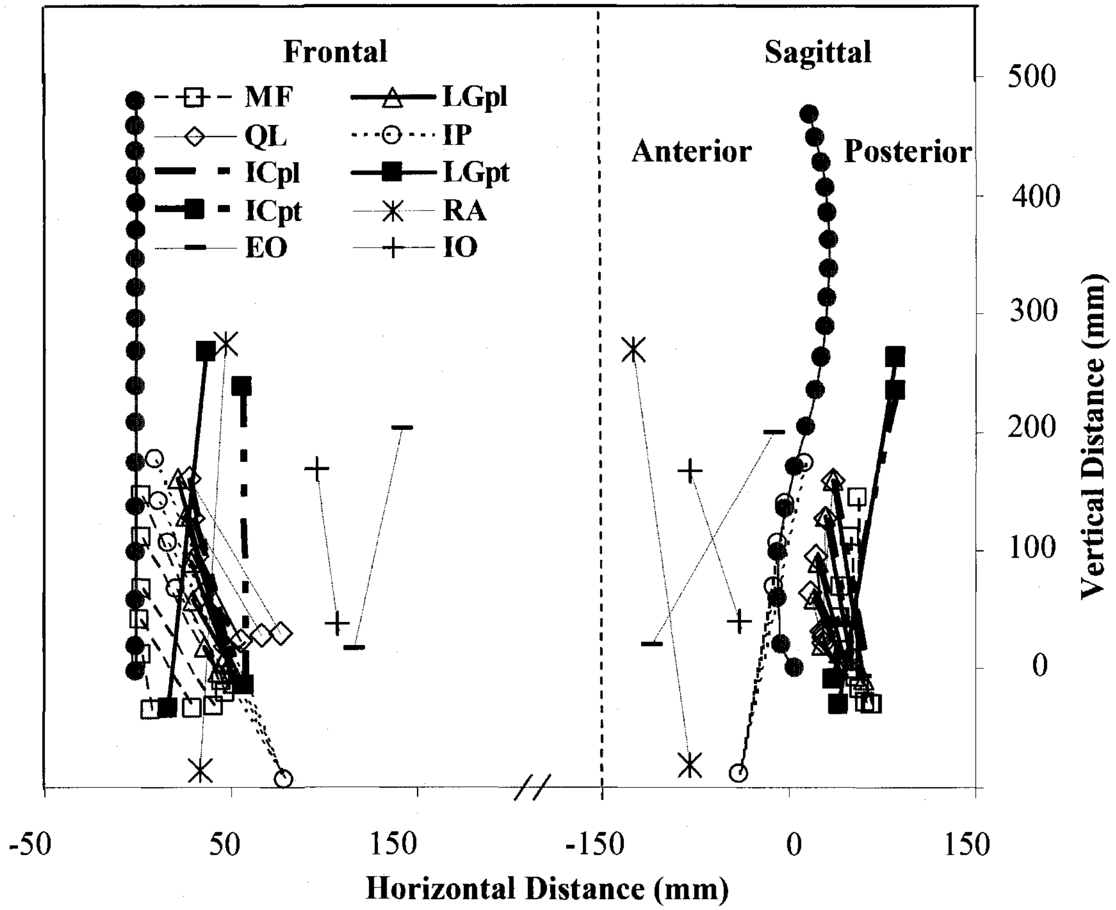

Figure 2.1 Representation of the model as well as global and local musculatures in the sagittal and frontal planes. Fascicles on one side are shown; ICpl: Iliocostalis Lumborum pars lumborum, ICpt: Iliocostalis Lumborum pars thoracic, IP: Iliopsoas, LGpl: Longissimus Thoracis pars lumborum, LGpt: Longissimus Thoracis pars thoracic, MF: Multifidus, QL: Quadratus Lumborum, IO: Internal Oblique, EO: External oblique, and RA: Rectus Abdominus.

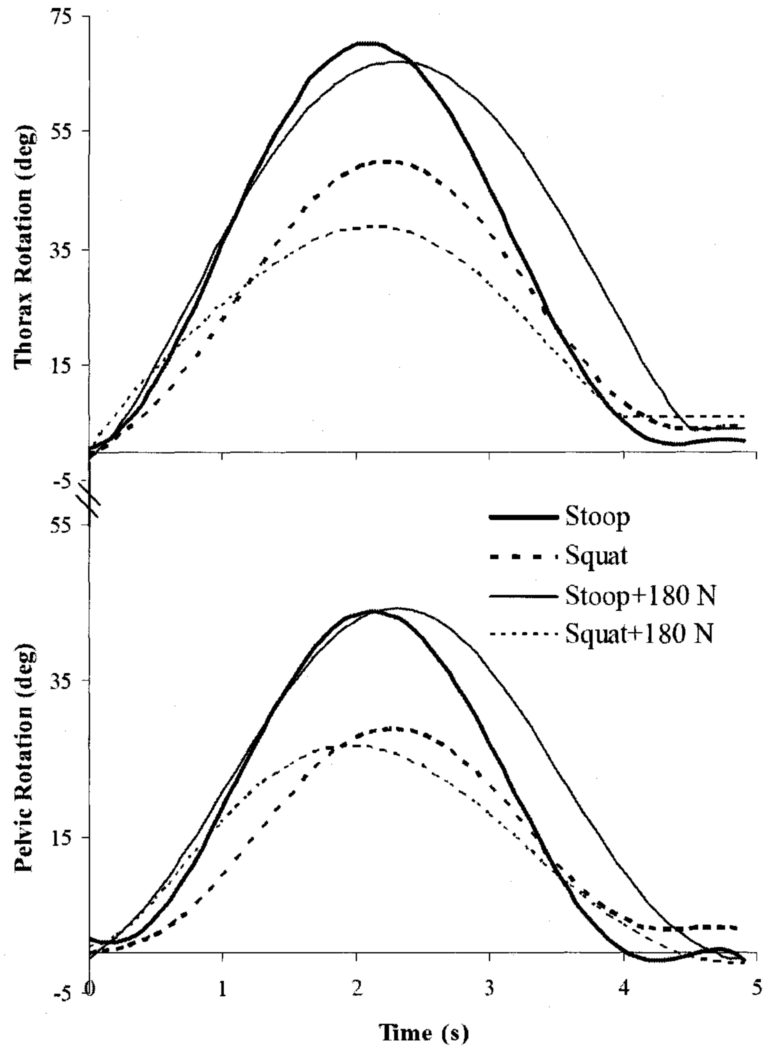

Figure 2.2 Prescribed thorax (top) and pelvis (bottom) rotations in the model for various cases based on in vivo measurements of a typical subject (smoothed by 6^{th} order polynomials, R^2 >98%). The T12-S1 rotations are subsequently prescribed in the model based on the difference between these two rotations and proportions given in the text.

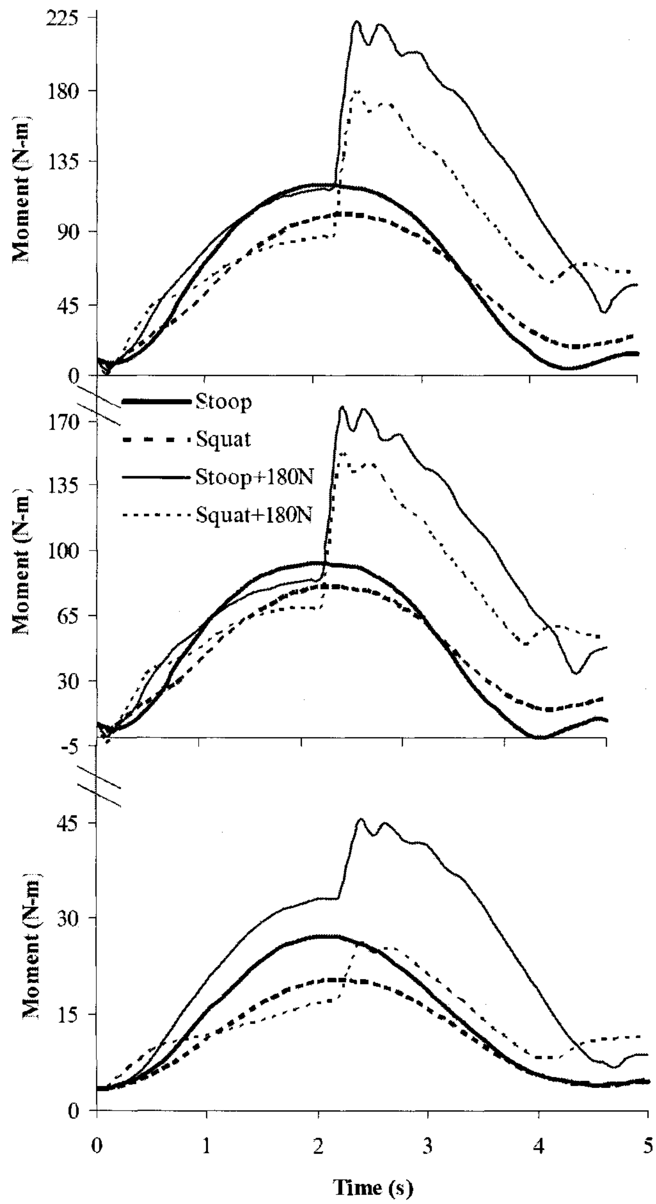

Figure 2.3 Predicted temporal variation of sagittal moments at the L5-S1 level for different cases (N-m); net external moment (top), portion resisted by muscle forces (middle), and portion resisted by passive ligamentous spine (bottom). For the cases with load in hands, the sharp increase in moments is noted as the load reaches its maximum value of 180 N in 0.2s duration.

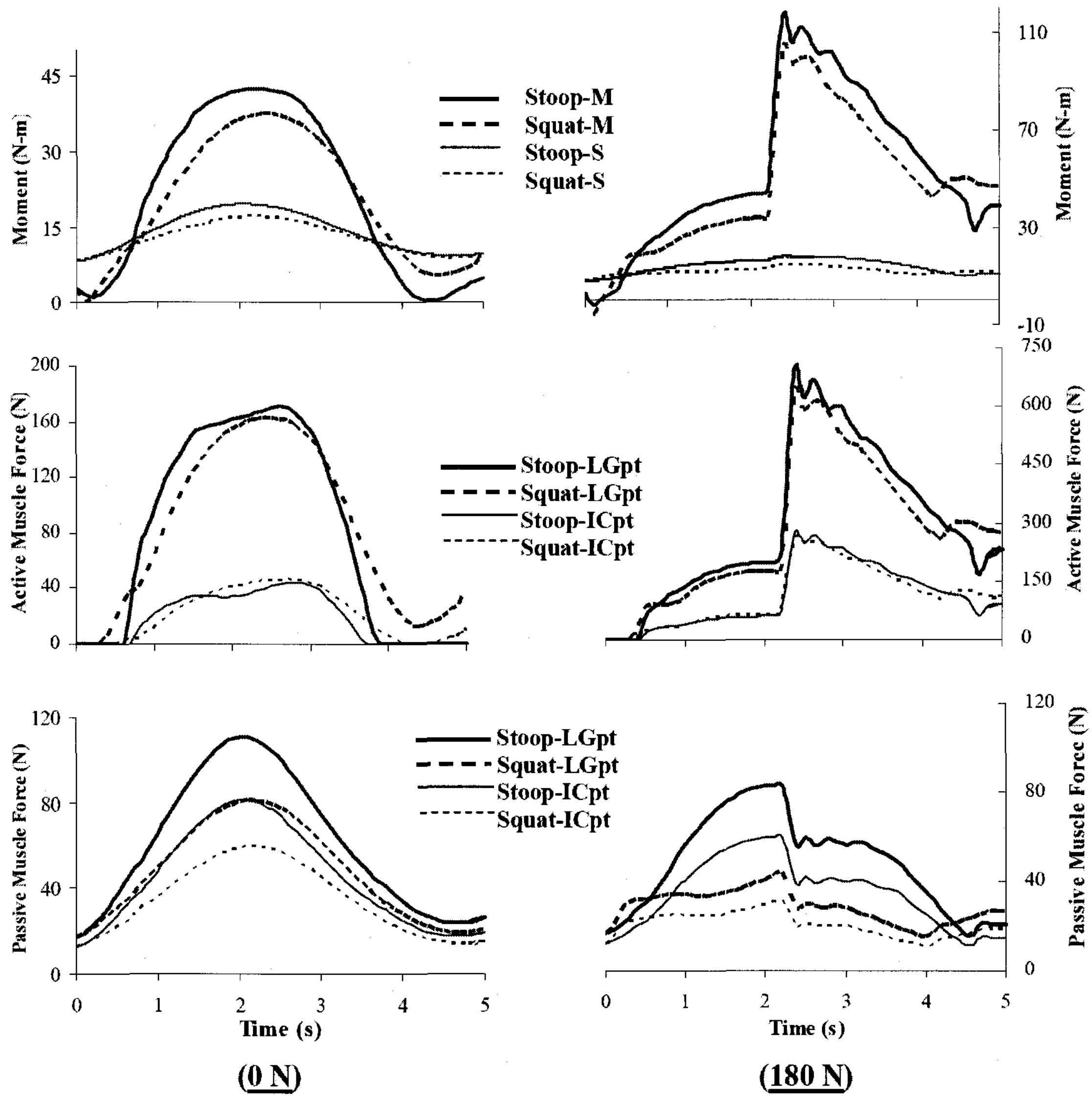

Figure 2.4 Predicted temporal variation of net external moment at the T12 level (top) and associated active (middle) and passive (bottom) global muscle (longissimus, LGpt, and illiocostalis, ICpt) forces for different lifting techniques without any load in hands (left side) and 180 N load in hands (right side). The rising time of 180 N external load applied in hands is shown by lines on the right. S: moment resisted by passive spine, M: moment resisted by muscles.

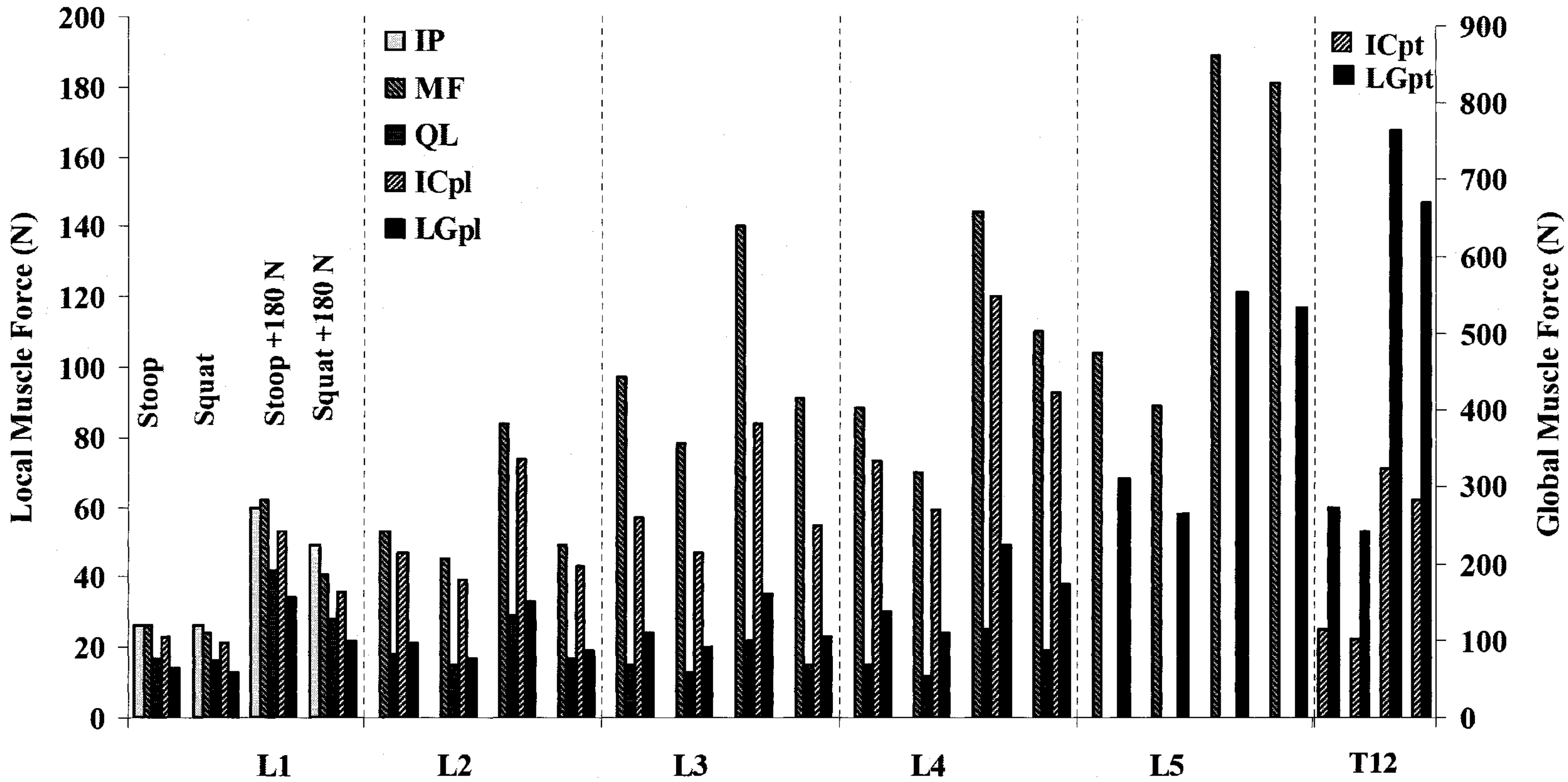

Figure 2.5 Maximum predicted total local and global muscle forces at various levels for different cases

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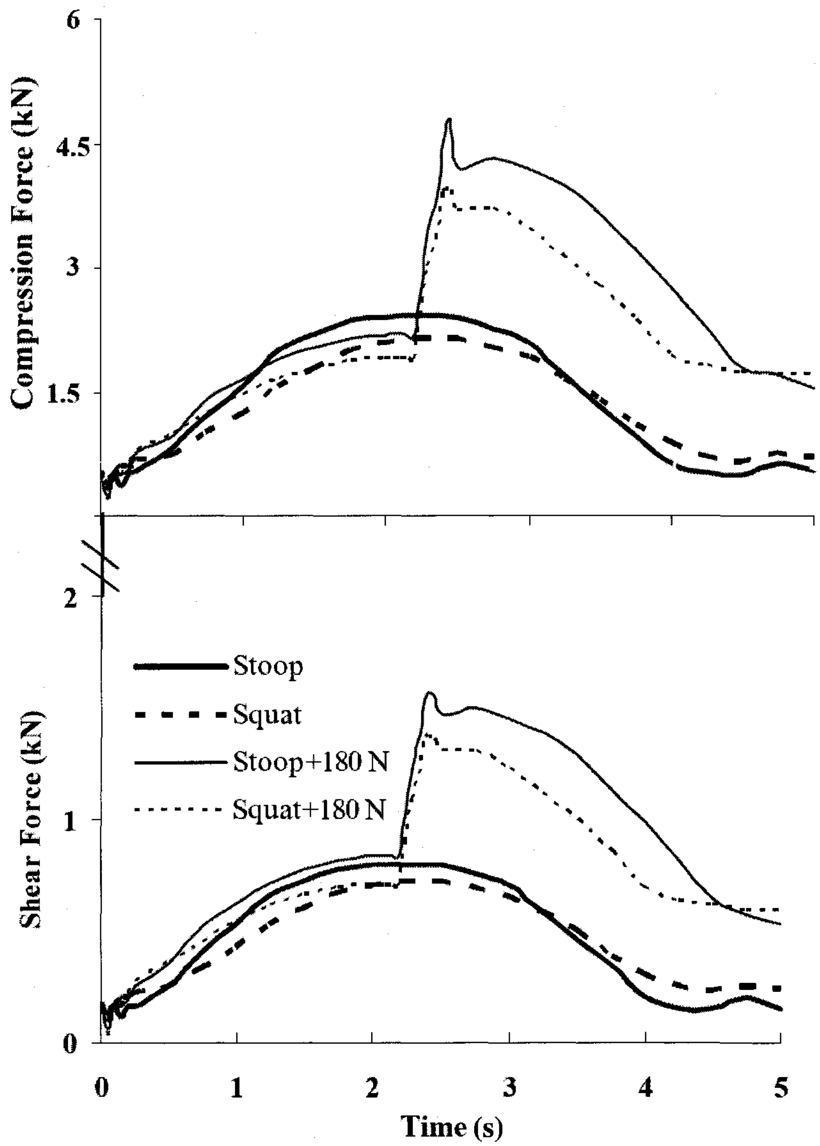

Figure 2.6 Computed temporal variation of local compression (top) and anterior shear (bottom) forces at the L5-S1 level for different cases. These forces are normal and tangential to the disc mid-height planes.

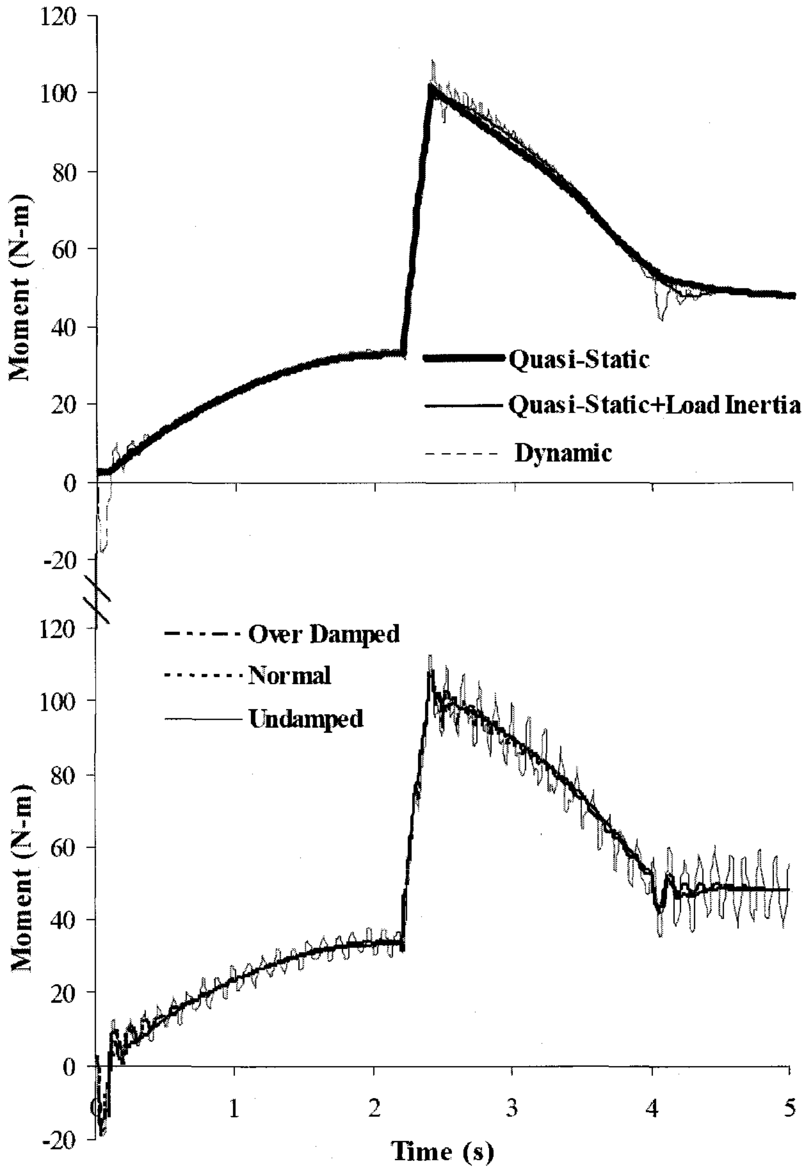

Figure 2.7 Predicted effect of changes in system dynamics characteristics on the net moment at the T12 level for the squat lift with 180 N in hands; effect of consideration of trunk and load inertias (top) and of damping (bottom).

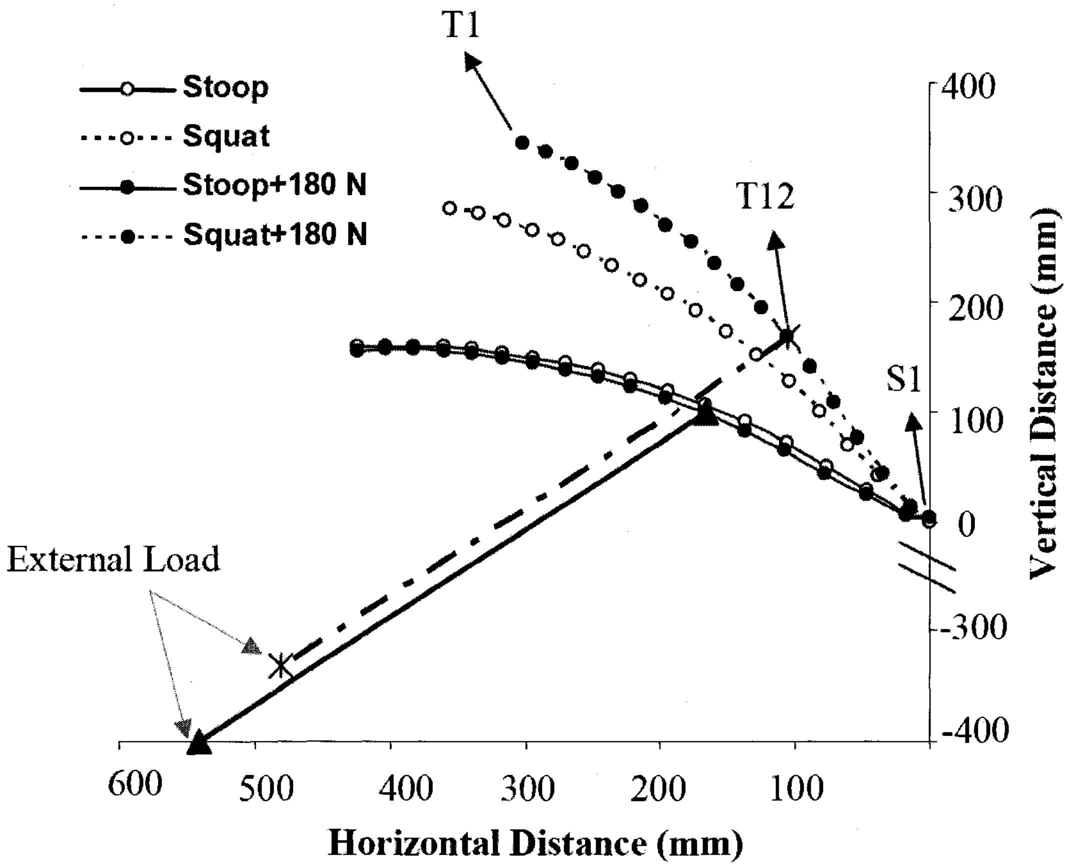

Figure 2.8 Deformed configurations of the model at the beginning of lifting phase (i.e., end of lowering phase) for various cases. The position of the external load held in hands, also shown, has identical horizontal lever arms with respect to the T12 in both squat and stoop configurations. The deformed configurations have been shifted in both horizontal and vertical directions to place the S1 at the origin of axes. The thorax rotation is much larger in stoop lifts (70° and 66.9° without and with load, respectively) than in squat lifts (49.7° and 38.4°).

CHAPTER THREE

(Article-II)

SPINAL STABILITY AND ROLE OF PASSIVE STIFFNESS IN DYNAMIC SQUAT AND STOOP LIFTS

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3.1 Abstract

The spinal stability and passive-active load partitioning under dynamic squat and stoop lifts were investigated as the ligamentous stiffness in flexion was altered. Measured in vivo kinematics of subjects lifting 180 N at either squat or stoop technique was prescribed in a nonlinear transient finite element model of the spine. The Kinematics-driven approach was utilized for temporal estimation of muscle forces, internal spinal loads, and system stability. The finite element model accounted for nonlinear properties of the ligamentous spine, wrapping of thoracic extensor muscles and trunk dynamic characteristics while subject to measured kinematics and gravity/external loads. Alterations in passive properties of spine substantially influenced muscle forces, spinal loads and system stability in both lifting techniques, though more so in stoop than in squat. The squat technique is advocated for resulting in smaller spinal loads. Stability of spine substantially improved with greater passive properties, trunk flexion and load. Simulation of global extensor muscles with curved rather than straight courses considerably diminished loads on the spine and increased stability throughout the task.

3.2 Introduction

Despite the recognized role of lifting in industrial low-back injuries (Burdorf and Sorock, 1997; Ferguson and Marras, 1997; Kuiper et al., 1999), the literature on safer lifting techniques remains controversial (Hsiang et al., 1997; van Dieen et al., 1999). In search of optimal lifting methods, squat lift (i.e., knee bent and back straight) is generally recommended over the stoop lift (i.e., knee straight and back bent) in bringing the load closer to the body and, hence, reducing the demand on back muscles in supporting external loads. The preference of the squat lifting technique over the stoop one has, however, been downplayed due to the lack of a clear biomechanical rationale for the promotion of either style (Hsiang et al., 1997; McGill, 1997; van Dieen et al., 1999). It should, nevertheless, be emphasized that an improved biomechanical assessment of various lifting techniques depends partly on a more accurate estimation of the associated load partitioning and stability in the human trunk.

Loads on the human spine are influenced not only by the gravity, inertia and external loads but also by the stiffness of the ligamentous spine and forces in trunk muscles. Changes in passive properties of the ligamentous spine have been demonstrated to affect equilibrium and stability of the spine in isometric (Arjmand and Shirazi-Adl, 2006; Gardner-Morse et al., 1995; Stokes and Gardner-Morse, 2003) and dynamic (Cholewicki and McGill, 1996) tasks. Greater passive stiffness would reduce muscle forces resulting in smaller compression forces on the spine while larger compression forces would occur under greater muscle forces following a decrease in the passive resistance. Passive properties of the human spine may alter with ageing, degeneration, injury, surgical intervention or even during daily activities as a function of single/combined loads and time (Shirazi-Adl, 2006; Wang et al., 1996). Likely influence of such changes on spine biomechanics in dynamic squat and stoop lifts remains to be investigated.

The stability of the spine is another important consideration that could play a role in muscle recruitments and back injuries. The system stability has been quantified under a number of postures and loading conditions (Arjmand and Shirazi-Adl, 2005, 2006; Cholewicki and McGill, 1996; Crisco and Panjabi, 1991; El-Rich and Shirazi-Adl, 2005; El-Rich et al., 2004; Gardner-Morse et al., 1995; Granata and Wilson, 2001). The relative effect of lifting technique, stoop versus squat, on system stability in dynamic lifts remains yet to be investigated. Previous biomechanical model studies on equilibrium and stability of the spine have often not properly accounted for the nonlinear passive resistance and/or complex geometry/loading/dynamics of the spine (Cholewicki and McGill, 1996; Crisco and Panjabi, 1991; Dietrich et al., 1991; Gardner-Morse et al., 1995; Stokes and Gardner-Morse, 2003). In addition, some earlier works attempting to estimate load partitioning and stability margin have not considered equations of motion simultaneously in all directions and/or at all levels.

To overcome shortcomings in earlier biomechanical model studies estimating muscle forces and spinal loads in lifting tasks, an iterative hybrid dynamic Kinematics-driven finite element approach is introduced and applied in which a priori measured kinematics of the spine and external loads along with nonlinear passive properties are considered. The estimated muscle forces, internal spinal loads and external/gravity/inertia loads satisfy equilibrium equations in deformed configurations at all directions and levels. The stability of the spine is investigated using nonlinear and linear analyses. The Kinematics-driven approach that results in a synergistic solution of the active-passive system has already been successfully applied to isometric conditions in upright (El-Rich and Shirazi-Adl, 2005; El-Rich et al., 2004; Shirazi-Adl et al., 2005) and flexed postures (Arjmand and Shirazi-Adl, 2005, 2006). In lifting, the curved path (wrapping) for global extensor muscles have also been considered; a representation that have been demonstrated to substantially affect estimated spinal loads in larger trunk flexion angles under isometric conditions (Arjmand et al., 2006).

In this study, based on measured trunk motions in dynamic lifts, the Kinematics-driven approach is employed to quantify the effect of changes in spinal passive properties on load partitioning and system stability under both squat and stoop lifts. The system stability margin is determined using nonlinear analyses as the gold standard as well as frequency and perturbation analyses at instantaneous deformed configurations during the course of motion assuming various muscle stiffness coefficients. The specific objectives of the current investigation are as follows; (1) to determine spine stability margin under dynamic stoop and squat lifts of a 180 N weight from near the floor; (2) to quantify the changes in internal spinal loads, muscle forces and stability margin due to alterations in passive stiffness of spine in foregoing lifting tasks; and (3) to quantify the effect of curved paths of global muscles on predictions as compared with straight paths. It is hypothesized that the passive resistance of both the ligamentous spine and trunk musculature plays a crucial role in equilibrium and stability of the system and that the extent of this role alters depending on the lifting technique and the relative changes in ligamentous tissue.

3.3 Methods

3.3.1 In vivo measurements

Fifteen healthy men with no recent back complications volunteered and signed consent forms for the study. Their mean (±S.D.) age, body height, and mass were 30±6 years, 177±7 cm, and 74±11 kg. While bending slightly forward, infrared light emitted markers, LED, were attached on the skin at the tip of the T1, T5, T10, T12, L1, L3, L5, and S1 spinous processes for evaluation of lumbar and torso flexions. Three extra LED markers were placed on the posterior-superior iliac spine and ilium (left/right iliac crests) for evaluation of pelvic rotation, and one on the load to track the position of weights in hands. A three-camera Optotrak system (NDI International, Waterloo, Ontario, Canada) was employed to collect 3D coordinates of LED markers. Subjects were asked, with no instruction on lumbar posture, to perform sagittaly-symmetric squat (knee bent) and stoop (knee straight) lifts with 180 N weight placed on a bar in front at

20 cm height from the floor. Each task lasted 4-5s and started from upright standing with no load in hands and ended in upright standing with the 180N load in hands.

3.3.2 Model studies

A sagittally symmetric T1-S1 beam-rigid body model made of six deformable beams to represent T12-S1 discs and seven rigid elements to represent T1-T12 (as a single body) and lumbosacral (L1-S1) vertebrae was used (Arjmand and Shirazi-Adl, 2005, 2006; El-Rich and Shirazi-Adl, 2005; El-Rich et al., 2004). The beams represented the overall nonlinear stiffness of T12-S1 motion segments (i.e., vertebrae, disc, facets and ligaments) at different levels with nonlinear axial compression-strain and sagittal moment-curvature relations. These nonlinear responses were based on numerical and measured results of previous single- and multi-motion segment studies [see (Arjmand and Shirazi-Adl, 2005; Shirazi-Adl, 2006; Shirazi-Adl et al., 2005; Yamamoto et al., 1989)]. The finite element beam elements were shear deformable with quadratic translations and rotation fields. For the current nonlinear dynamic analyses; the trunk mass and mass moments of inertia were assigned at different levels along the spine at their respective gravity centers based on published data (Pearsall et al., 1996; Zatsiorsky and Seluyanov, 1983). Connector elements parallel to deformable beams were added to account for the intersegmental damping using measured values (Kasra et al., 1992; Markolf, 1970) where translational damping =1200 Ns/m and angular damping = 1.2 Nms/rad.

The temporal variation of pelvic and thorax rotations of a typical subject measured under two lifting techniques along with the intervening lumbar vertebral rotations were prescribed as input data into the model studies. Mean data of all subjects was not considered to avoid associated fluctuations in temporal variation of input kinematics. These input data were verified to be within the range of mean ± standard deviation of all subjects and to preserve the overall differences between squat and stoop lifts. Temporal variation in measured lever arm of the external load (180 N in hands) with respect to the T12 was noted to be statically insignificant in squat and stoop lifts among subjects; 399±87 mm and 198±45 mm in squat at the beginning and end of lifting compared to 353±109 mm and 224±74mm in stoop, respectively. To circumvent likely inter-subject differences in this lever arm, the model studies of both squat and stoop lifts used identical values calculated as the average of these measurements. The total lumbar rotation, calculated as the difference between the foregoing two rotations, was subsequently partitioned in between various segments based on values reported in earlier investigation (Dvorak et al., 1991; Frobin et al., 1996; Pearcy et al., 1984; Plamondon et al., 1988; Shirazi-Adl and Parnianpour, 1999; Yamamoto et al., 1989). Relative proportions of ~7%, 12%, 15%, 22%, 27%, and 17% were used for T12-L1 to L5-S1 levels, respectively.

A sagittally symmetric muscle architecture with 46 local (attached to the lumbar vertebrae) and 10 global (attached to the thoracic cage) muscles was used (Fig. 3.1) (Arjmand and Shirazi-Adl, 2005, 2006).To evaluate muscle forces, Kinematics-based algorithm was employed to solve the redundant active-passive system subject to prescribed measured kinematics, inertia, damping and external loads. To resolve the redundancy problem at each level, an optimization approach with the cost function of minimum sum of cubed muscle stresses was also needed along with inequality equations of unknown muscle forces remaining positive and greater than their passive force components (calculated based on instantaneous muscle length and a tension-length relationship (Davis et al., 2003)) but smaller than the sum of their respective maximum active forces (i.e., 0.6 PCSA (Winter, 2005)) and the passive force components. To simulate wrapping of global muscles (i.e., longissimus thoracis pars thoracic and iliocostalis lumborum pars thoracic), they were constrained to follow curved paths whenever their distances from T12-L5 vertebral centers decreased below 90% of their respective values at undeformed configuration (i.e., to reach the limit values of 53, 53, 55, 56, 54 and 48 mm for the global longissimus and 58, 56, 56, 55, 52 and 45 mm for the global iliocostalis at T12 to L5 vertebrae, respectively). This wrapping mechanism, similar to that formulated in our earlier static simulations (Shirazi-Adl, 1989, 2006; Shirazi-Adl and Parnianpour, 2000), was considered in order not to allow the line of action of these muscles approach unrealistically close to the vertebrae while at the same time simulating a maximum of 10% reduction in their lever arms as observed during forward flexion tasks (Arjmand et al., 2006; Jorgensen et al., 2003; Macintosh et al., 1993; Tveit et al., 1994).

The wrapping contact mechanism acts to enforce kinematics constraints on deformations whereby the penetration of global muscles into underlying muscles and vertebrae are prevented as the spine flexes forward. Such constraints change the orientation of global muscles and result in contact forces in between global muscles and the spine at different levels. During the analysis, when wrapping of a muscle was detected at a vertebral level (i.e. muscle lever arm reached below the pre-defined minimum values given earlier), muscle course was modified to pass over the pre-defined point. Subsequently, a contact force between the muscle and the corresponding vertebra was evaluated using the equilibrium equation in the instantaneous configuration assuming a frictionless contact that results in a constant muscle force along its entire length. The axial compression and horizontal shear penalties of calculated muscle forces along with wrapping contact loads (if needed) were then fed back into the finite element model as additional updated external loads. This iterative approach was continued at each time instance till convergence was reached.

Once the muscle forces were calculated, the model was modified with single (or multiple) uniaxial elements introduced to directly represent each muscle between their insertion points (and wrapping points if existed). The nonlinear analysis was repeated under the same external loads but with no prescribed rotations (with the exception of the pelvic tilt). The stiffness of each uniaxial element, k, was assigned using linear stiffness-force relation (i.e. $k=q\ F(t)\ /\ L(t)$) in which the muscle stiffness is proportional to the instantaneous muscle force, $F(t)$, and inversely proportional to its current length, $L(t)$, with q as a dimensionless muscle stiffness coefficient that is taken to be the same for all muscles. At each instance of time, non-linear analysis were performed with different q values to identify the minimum (critical) q value to maintain system stability. The stability margin under each q value was further investigated at the loaded deformed configurations by natural frequency and linear perturbation analyses. In general, to assess the stability of a nonlinear system using the static stability criterion (i.e., divergence type), one can use linear buckling, perturbation or free vibration analyses at a deformed stressed configuration evaluated based on a prior nonlinear analysis. In perturbation analysis, the translation of T1 vertebra under application of a unit load was used for different q values to identify the associated stability margin (i.e., stability and stiffness decreased as q approached the critical value). On the other hand, smallest natural frequency of structure, determined using free vibration analysis, was also used as an indication of structural stability (i.e., system became unstable as the smallest natural frequency approached zero).

The finite element program ABAQUS was used to carry out nonlinear transient analyses while the optimization procedure was analytically solved using an in-house program based on Lagrange Multipliers Method (Raikova and Prilutsky, 2001). Implicit algorithm with unconditionally stable Hilber-Hughes-Taylor (Hilber et al., 1978) integration operator was used to solve the nonlinear transient problem. The time step was automatically selected by the solver but was constrained to remain <0.01s.

In order to determine the effect of alterations in ligamentous spine properties on muscle forces, spinal loads and stability margin, analyses were repeated for both squat and stoop lifts with segmental nonlinear flexural rigidities at all levels in the sagittal plane varied by +20%, -20% , and -40%. Role of wrapping of global muscles and their interaction with spine on results was also investigated by reanalyzing squat lifting using straight global muscles rather than curved ones.

3.4 Results

Subjects carried stoop lifts, as compared with squat, with significantly larger thorax (T12), pelvis (S1), and lumbar (T12-S1) rotations ($p<0.01$, 0.05 and 0.03, respectively) (see Fig. 3.1). The temporal variation of results demonstrate a sudden rise right after ~2s as subjects picked the 180 N load from the floor while extending backward to upright position. The internal spinal loads reached their maximum at the lowermost L5-S1 level. Alterations in passive stiffness of the ligamentous spine substantially influenced results in both stoop and squat lifts, though more so in former than in latter (Table 3.1, Figs. 3.2 and 3.3). The reduction in passive stiffness increased both compression and shear forces while reverse trends were computed when greater stiffness for the ligamentous spine was assumed. The sagittal moment at the L5-S1 varied by as much as 32.2 Nm and 12.9 Nm in stoop and squat lifts, respectively (Table 3.1). These changes in moments resulted in compensatory changes in muscle forces and subsequent variations in spinal compression and shear loads (Figs. 3.2 and 3.3). The 40% decrease in flexural rigidity did not noticeably alter shear forces at upper disc levels while it increased those at the lowermost L5-S1 level by 89 N and 106 N in stoop and squat lifts, respectively (Table 3.1). In this case, compression forces were increased by maximum values of 602 N and 271 N at the L5-S1 level in stoop and squat lifts, respectively. Similarly due to larger lumbar rotations, the influence of changes in ligamentous spine properties on wrapping contact forces was much more pronounced in stoop lifting (Table 3.2).

The critical muscle stiffness coefficient, q, identifies the minimum value required to maintain the system stability. The spinal stability margin would hence increase as the muscle stiffness coefficient exceeds such critical value. The critical muscle stiffness coefficient was found to significantly vary as a function of trunk flexion angle and external load (i.e., forward bending with no load in hands and backward extension with the load in hands) (Fig. 3.4). Relative stability margin of stoop lift with respect to squat lift varied in the forward flexion phase of the task with no load in hands, being smaller in the earlier periods of flexion (i.e. larger critical q values) whereas greater as trunk flexion increased (i.e. smaller critical q values) (Fig. 3.4, top). Alteration in passive properties of the spine influenced the stability primarily during the early stages of lifting when the subjects had no load in hands and the trunk flexion was relatively small (Fig. 3.4, middle).

Consideration of wrapping of global extensor muscles as compared with the model with straight global muscles markedly influenced results; the compression force decreased at all levels by as much as 543 N at the L2-L3 level and the shear force increased at lowermost levels (Table 3.1 and Fig. 3.5). Neglecting curved paths for global muscles and their respective contact forces (due to interaction with the spine) resulted in considerably less stable system throughout trunk flexion and that despite greater muscle forces (Fig. 3.5).

3.5 Discussion

The objectives of this work were to estimate the stability of spine for different lifting techniques and to determine changes in spinal loads and stability margin due to alterations in ligamentous passive properties. Despite earlier works on biomechanics of spine in search of safer lifting techniques (Hsiang et al., 1997; van Dieen et al., 1999), the spinal stability as well as the effects of alterations in ligamentous passive stiffness on spinal response remain to be investigated in squat versus stoop dynamic lifts. Spinal loads and muscle forces were significantly affected not only by trunk flexion and external loads but also by the lifting technique as well as alterations in ligamentous passive properties. Contribution of passive components of the spine and musculature in counterbalancing net moments of gravity, inertia and external loads reached the maximum values of 56.4%, 24.5% in stoop lift right before and after picking the load, respectively. These values increased to 67.6% and 31.2% as passive stiffness increased by 20% whereas they decreased to 46.9% and 18.6% and further to 37.9% and 14.1% as the passive stiffness reduced by 20% and 40%. In squat lift, due to smaller lumbar rotations (Fig. 3.1), similar trends but at smaller magnitudes were predicted. These results support the hypothesis on the crucial role of passive structures in lifting.

Safer lifting techniques could be established based on the premise that excessive compression forces, shear forces or flexion moments in the ligamentous spine could cause injury. The compression strength of Lumbar motion segments is reported to be in the range of 2-10 kN (Brinckmann et al., 1989; Jager and Luttmann, 1997; Ortoft et al., 1993). Jager and Luttmann (1991) reported values of 5.81±2.58 kN for males and 3.97±1.5 kN for females based on relatively large sample populations. These values may decrease when accounting for the deteriorating effect of fatigue and micro-failures (Dolan et al., 1994). The strength in shear force has been reported to be >1 kN (Cyron et al., 1976; Miller et al., 1986) while that in flexion moment exceeds 70 Nm (Miller et al., 1986; Neumann et al., 1992; Osvalder et al., 1993). Notwithstanding the effect of strain rate, existing injuries/degeneration and combined loading on these strength values, a lower risk of injury could be associated with the lifting technique that yields much smaller loads on spine without generating excessively greater muscle forces and smaller stability margin. In this respect, results of this study advocate a squat lift over stoop lift in reducing the risk of fatigue and injury to passive and active components without necessarily deteriorating the spinal stability.

The relative stability of stoop lift versus squat lift was not consistent over time in forward flexion phase of the task (Fig. 3.4). Initially, the stability margin was larger in squat lift due mainly to greater activity of abdominal muscles at the beginning of the task (Figs. 3.2 and 3.3). This was, however, reversed as the abdominal activity disappeared and flexion angle increased towards picking the load (Fig. 3.4) thereby generating greater stiffness in both active and passive sub-systems in stoop lift. The effect of changes in the ligamentous passive properties on stability was also more evident in the earlier forward flexion phase of the task in which higher passive properties demanded lower muscle stiffness coefficients in order to maintain stability. As the load was lifted, the critical muscle stiffness coefficient dropped to nil irrespective of the lifting technique and alterations in passive properties. These predictions highlight the crucial role of passive and active components in stabilization of the spine. Considerable decrease in critical muscle stiffness coefficient during the activation of abdominal muscles in earlier stages of forward flexion phase was due to activation of abdominal muscles along with passive activity of global and local extensor muscles when negative net moment was required (Fig. 3.2). This demonstrates the positive role of such antagonistic activities in enhancement of spine stability (El-Rich et al., 2004; Gardner-Morse et al., 1995; Granata and Orishimo, 2001; Potvin and O'Brien, 1998; Shirazi-Adl et al., 2005).

Consideration of curved global muscles and their interaction with spine was found to have significant effects on predicted spinal loads and muscle forces at larger flexion angles under the load when the wrapping contact forces reached their maximum values. In this case, the maximum muscle forces and consequently spinal compression forces diminished. The improvement in spinal stability was, however, evident throughout the range of flexion and that despite larger muscle forces in the model with straight global muscles. This deterioration in system stability in presence of straight muscles is due to the generated larger compression forces on the spine and smaller lever arms of global muscles. On the other hand, the contact forces between wrapping global muscles and the spine along the lumbar spine could increase the system stability. Simulation of wrapping without the proper consideration of these contact forces in equilibrium and stability at deformed configurations of the spine is not, hence, reliable adversely affecting the accuracy of simulations. A limited number of dynamic studies (Cholewicki and McGill, 1996; McGill and Norman, 1986) considered also curved paths for global muscles but with no indication of the associated wrapping contact forces and their effect as additional loads.

In the current study, the stability of the spine at each instance of time was investigated using nonlinear analyses at different muscle stiffness coefficient values, q. As q decreased, the loss of stability at a critical q value was also confirmed by parallel perturbation and free vibration analyses at deformed stressed configurations. At each time instance, the critical q value in a specific case was identified as the lowest eigen value in free vibration analyses approached zero and the displacement under unit force perturbation analyses increased substantially, These latter analyses should necessarily be performed at the instantaneous deformed states of the system in order to avoid over-estimation of stability margin in such nonlinear and imperfect structures. The stability of spine in dynamic lifting has also been studied by Cholewicki and McGill (1996) while the subjects lifted/lowered weights off/to the floor with apparently a squat technique. This latter study, however, defined the stability index based on the determinant of Hessian matrix of the structure rather than the lowest eigen value (Howarth et al., 2004). While neglecting also the translational degrees-of-freedom at different levels and the pelvic tilt, they also reported an increase in system stability at larger flexion angles, with load and as the passive properties were increased (Cholewicki and McGill, 1996).

The smallest stiffness coefficient to maintain the system stability varied from as high as ~100 in upright posture at the beginning with no load in hands and assuming a 40% reduction in passive stiffness to as low as nil found in all cases under larger flexion angles and load in hands except when the global extensor muscles were taken straight rather than curved. These critical q values fall in the range of 0.5 to 42 (Crisco and Panjabi, 1991) and 36 to 170 (Cholewicki and McGill, 1995) reported in the literature. It should be noted that much smaller critical q values, especially at the beginning of flexion task, would have been estimated in our study, had some coactivity level in abdominal muscles been considered. Our earlier study in upright posture (El-Rich et al., 2004) demonstrated a substantial drop in the critical q value in presence of relatively low abdominal coactivity. It should be emphasized that the choice of linear force-stiffness relation taken in this study for the muscles (i.e., $k = q\ F(t)\ /L(t)$), rather than a nonlinear relation and of identical q for all active muscles have absolutely no influence on computed muscle forces and internal loads. The stiffness of a muscle, however, influences the system stability margin. Moreover, the prediction of a critical muscle stiffness coefficient of $q=0$ under larger flexion angles with load in hands indicates that the system stability may be maintained with no stiffness contribution from muscles. Any activation in muscles, in such cases with no need for a muscle stiffness, would never-the-less improve the system stability via its contribution through the stress stiffness matrix.

To prescribe measured rotations in the model, kinematics data of one typical subject rather than the mean of all subjects were considered. This was done due mainly to noticeable variations in duration of lowering/lifting phases in between subjects. Relative values of intersegmental rotations defined as a fixed proportion of the total lumbar rotation was considered constant throughout the task. This, however, may not hold true in vivo since the relative demand at different segmental levels varies during the lifting. These relative ratios were taken from data obtained in static measurements (Dvorak et al., 1991; Frobin et al., 1996; Pearcy et al., 1984; Plamondon et al., 1988; Shirazi-Adl and Parnianpour, 1999; Yamamoto et al., 1989) which have also been used in previous dynamic studies (McGill and Norman, 1986; Potvin and McGill et al., 1991; Potvin and Norman et al., 1991) in order to evaluate the contribution of passive tissues in offsetting external loads. A maximum of 10% reduction in the lever arm of global muscles during flexion was considered based on earlier studies (Jorgensen et al., 2003; Macintosh et al., 1993; Tveit et al., 1994). This value may however vary depending on the posture (Arjmand et al., 2006). The rate-dependent viscoelastic properties of spinal segments, which have been shown to play an important role at much higher loading rates (Neumann et al., 1994; Wang et al., 1998, 2000) were not considered in this study. The accuracy of results was found unchanged with refinement of mesh or much smaller time increments (i.e. 0.0001 sec.). Since the damping matrix is symmetric and positive definite, at a deformed stressed configuration under loading, the static stability criterion applies in which the spinal system ceases to be stable when its lowest natural frequency reaches zero or equivalently as the second variation of the potential energy vanishes (Komarakul-na-nakorn and Arora, 1990). Static perturbation and frequency analyses, in this case and as expected, yielded identical critical muscle stiffness coefficient values when instantaneous stressed configurations were used. Effects of discs damping were found to be negligible in transient analyses and were omitted subsequently in stability analyses.

In conclusion, alterations in passive properties of spine similar to changes in trunk flexion and external loads significantly influenced spinal loads during dynamic stoop and squat lifting. With relatively similar stability margin while bearing significantly lesser spinal loads (i.e., compression force, shear force and moment); our results advocate squat lifting over stoop lifting. Squat lift also demonstrated less sensitivity to changes in passive properties of spine. Finally, neglecting curved path of global extensor muscles in modeling lifting tasks would, at larger flexion angles, result in estimation of greater spinal loads whereas smaller stability margins.

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3.7 References

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Table 3.1 Maximum internal loads (occurring nearly at the time of maximum trunk flexion) in passive ligamentous spine for different Lifting techniques and segmental levels as spinal passive stiffness was altered.

		Stoop				Squat				Squat†
		-40%	-20%	Intact	+20%	-40%	-20%	Intact	+20%	Intact
T12-L1	S‡	379	383	383	384	218	221	220	221	370
	C‡	2624	2534	2410	2236	2502	2409	2297	2171	2578
	M‡	8.6	12.9	18.7	26.4	6.3	10	15.2	20.3	16
L1-2	S	383	382	382	386	186	189	190	192	287
	C	3188	3072	2911	2716	2865	2751	2643	2492	3100
	M	9.6	14.6	21.4	29.5	7.5	10.5	15.5	21.3	16.7
L2-3	S	231	241	245	253	72	77.6	77	76.5	61
	C	3628	3517	3372	3192	3105	3012	2927	2825	3470
	M	8.6	13.6	19.8	27.4	5.2	8.3	12.3	16.5	12.3
L3-4	S	565	549	535	514	412	387	364	348	357
	C	4076	4001	3891	3766	3363	3306	3262	3184	3768
	M	6.3	10.4	15	20.7	3.6	5.3	7.5	10.6	8.3
L4-5	S	530	508	501	492	473	431	409	393	331
	C	4806	4692	4511	4292	3895	3809	3738	3624	4227
	M	7.1	12.4	19	27.1	3.8	6	8.4	12	8
L5-S1	S	1723	1677	1634	1577	1517	1450	1410	1362	1360
	C	5423	5172	4821	4435	4347	4226	4076	3891	4491
	M	13.3	21.7	33	45.5	7.4	10.2	14.8	20.3	14.9

†Squat- with straight global muscles
‡ M (N.m): sagittal moment, C (N): local compression force, and S (N): local anterior shear force

Table 3.2 Maximum wrapping contact forces (N, occurring nearly at the time of maximum trunk flexion) for different lifting techniques and segmental levels as spinal passive stiffness was altered.

	Stoop				Squat			
	-40%	-20%	Intact	+20%	-40%	-20%	Intact	+20%
T12-L1	0	0	0	0	0	0	0	0
L1-2	91	72	66	61	49	38	39	34
L2-3	151	135	124	112	85	75	71	64
L3-4	108	109	102	98	46	47	47	45
L4-5	186	189	180	167	92	95	88	84
L5-S1	258	266	262	246	138	139	129	123

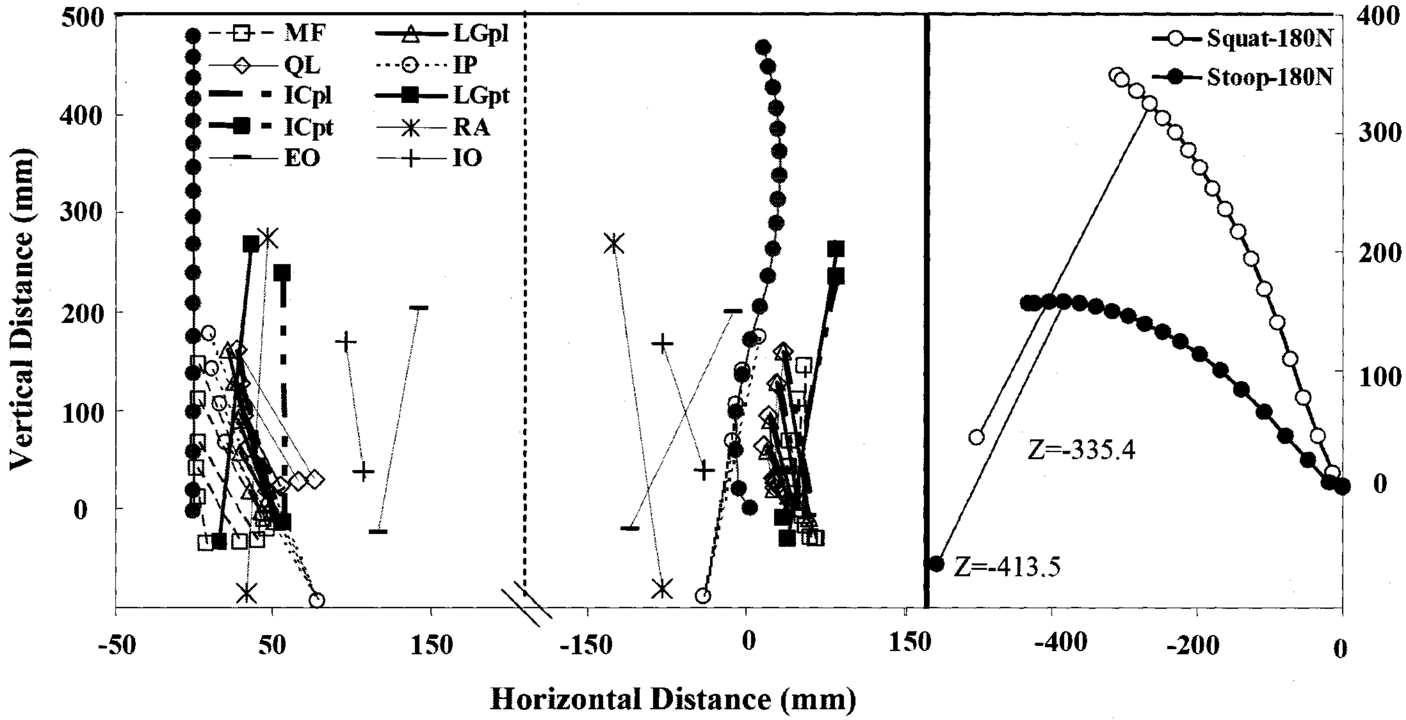

Figure 3.1 Representation of the model as well as global and local musculatures in the sagittal (on the right) and frontal (at the middle, fascicles on one side are shown) planes. ICpl: Iliocostalis Lumborum pars lumborum, ICpt: Iliocostalis Lumborum pars thoracic, IP: Iliopsoas, LGpl: Longissimus Thoracis pars lumborum, LGpt: Longissimus Thoracis pars thoracic, MF: Multifidus, QL: Quadratus Lumborum, IO: Internal Oblique, EO: External oblique, and RA: Rectus Abdominus. Deformed configurations of the model at the instance of lifting the 180 N load for stoop and squat lifts along with their respective load position (shifted to coincide S1 levels) are shown on the right

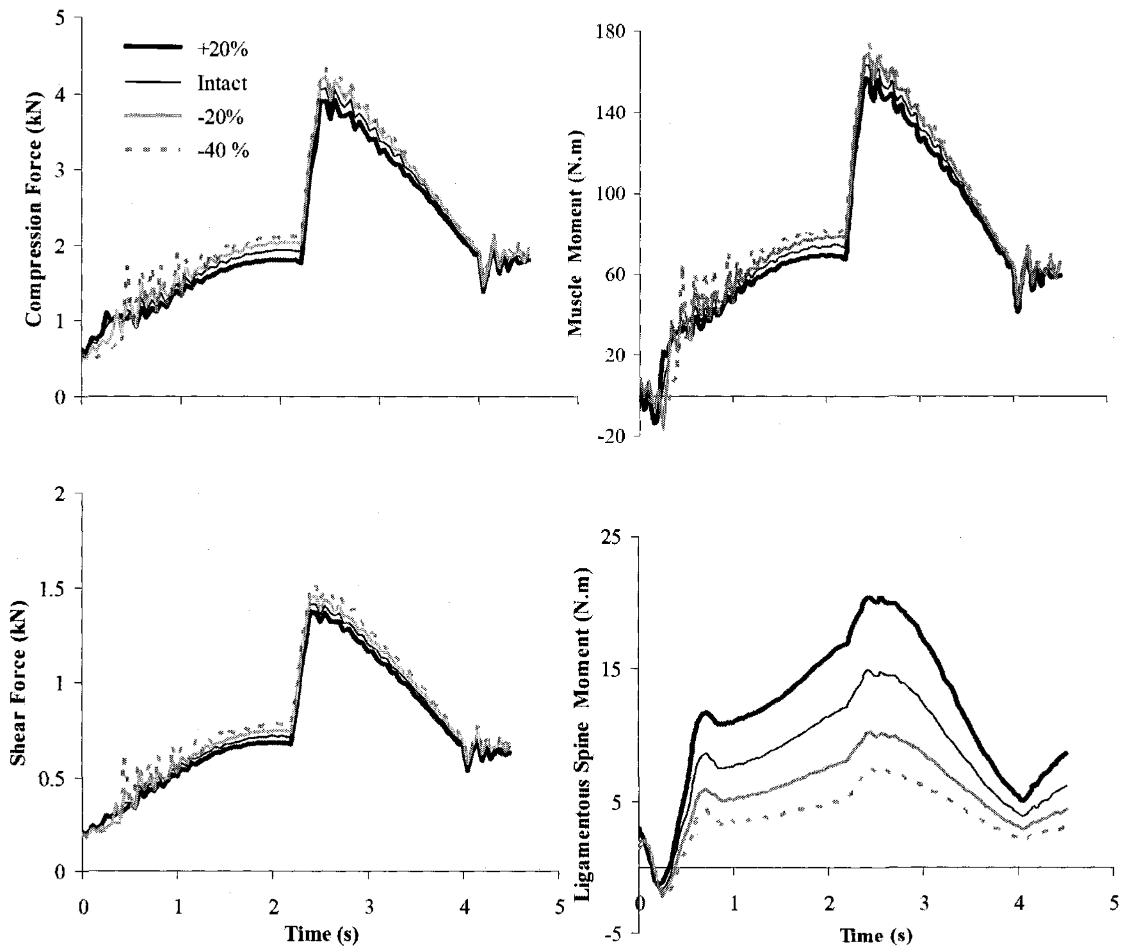

Figure 3.2 Predicted temporal variations of internal forces at the L5-S1 as well as muscles and ligamentous spine contributions in carrying the net moment at the L5-S1 level for different passive properties in squat lifting.

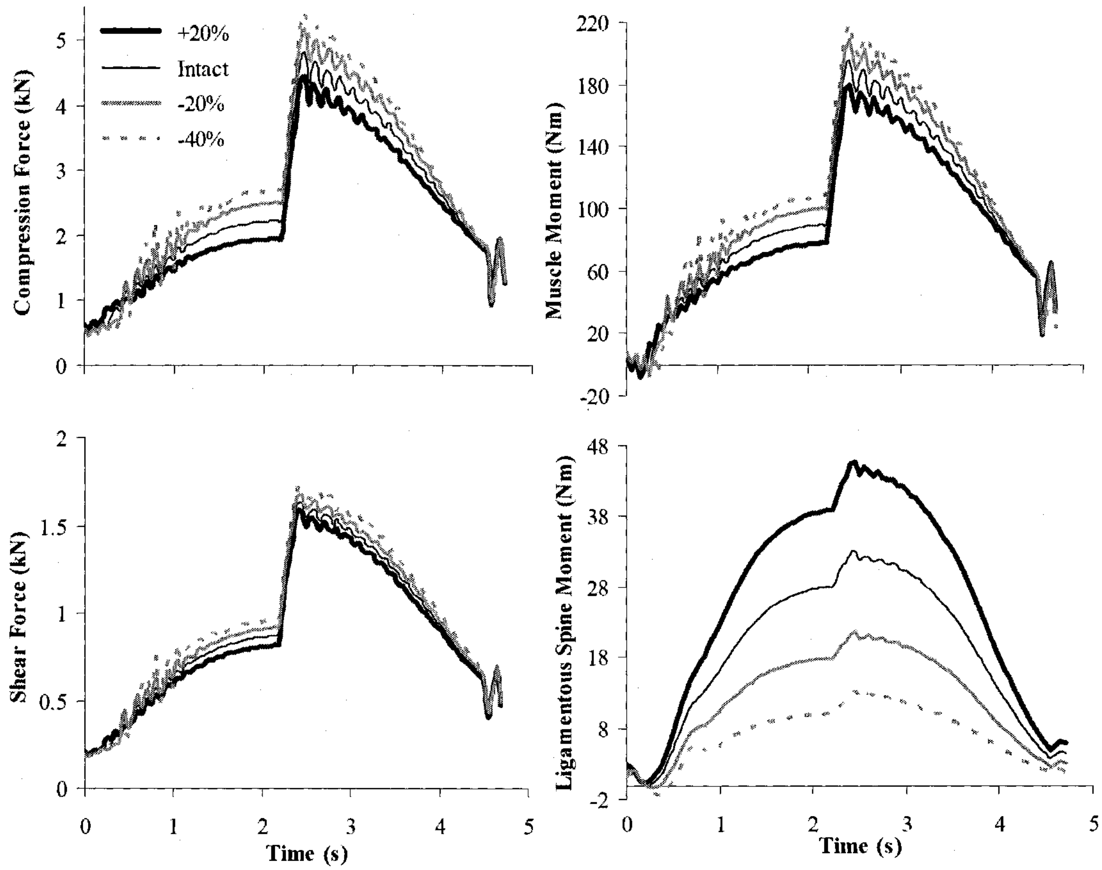

Figure 3.3 Predicted temporal variation of internal forces at the L5-S1 as well as muscles and ligamentous spine contributions in carrying the net moment at the L5-S1 level for different passive properties in stoop lifting.

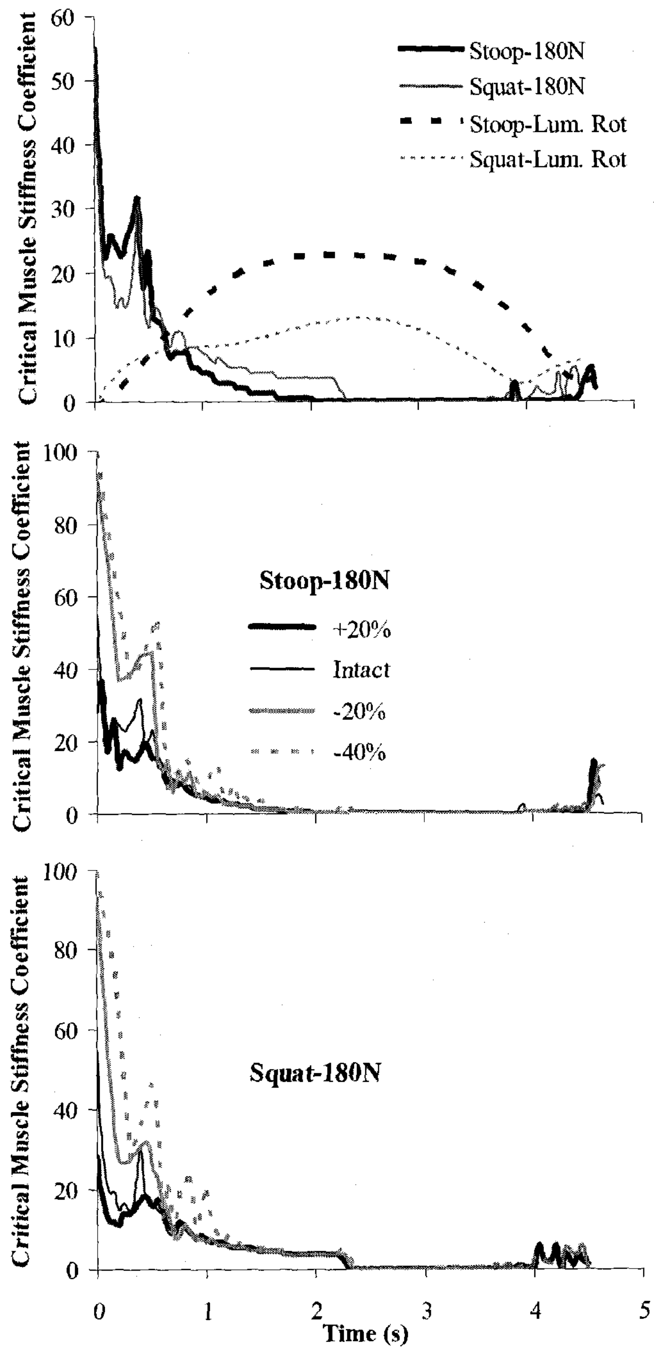

Figure 3.4 Predicted temporal variation of minimum (critical) muscle stiffness coefficient, q, for different lifting techniques and spinal passive properties. The lumbar rotations are also shown in degrees in the top figure.

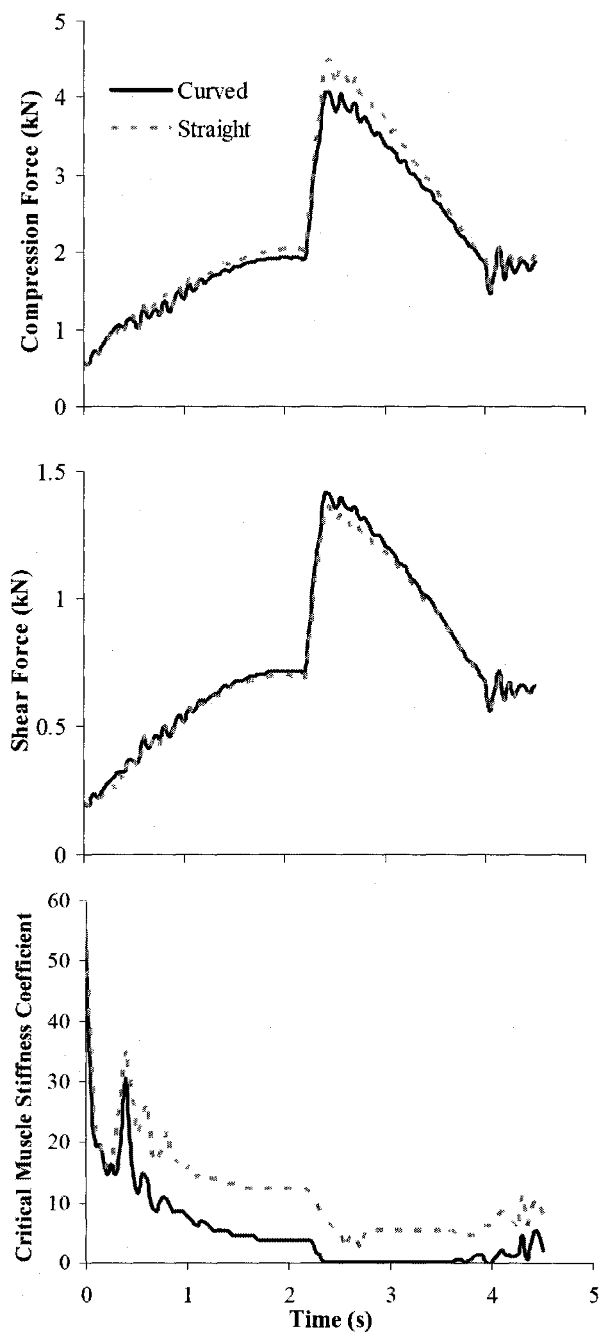

Figure 3.5 Predicted effect of global muscles simulated either as curved or straight on the L5-S1 compression (top) and anterior shear (middle) forces as well as minimum (critical) muscle stiffness coefficient, q, (bottom) for the squat lift.

CHAPTER FOUR

(Article-III)

COMPUTATION OF TRUNK EQUILIBRIUM AND STABILITY IN FREE FLEXION-EXTENSION MOVEMENTS AT DIFFERENT VELOCITIES

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4.1 Abstract

Velocity of movement has been suggested as a risk factor for low-back disorders. The effect of changes in velocity during unconstrained flexion-extension movements on muscle activations, spinal loads, base reaction forces and system stability was computed. In vivo measurements of kinematics and ground reaction forces were initially carried out on young asymptomatic subjects. The collected kinematics of three subjects representing maximum, mean and minimum lumbar rotations were subsequently used into the Kinematics-driven model to compute results during the entire movements at three different velocities.

Estimated spinal loads and muscle forces were significantly larger in fastest pace as compared to slower ones indicating the effect of inertial forces. Spinal stability was improved in larger trunk flexion angles and fastest movement. Partial or full flexion relaxation of global extensor muscles occurred only in slower movements. Some local lumbar muscles, especially in subjects with larger lumbar flexion and at slower paces, also demonstrated flexion relaxation. Results confirmed the crucial role of movement velocity on spinal biomechanics. Predictions also demonstrated the important role on response of the magnitude of peak lumbar rotation and its temporal variation.

4.2 Introduction

Manual material handling of lighter loads at higher velocities is prevalent in current industrial environments. The risk of back injuries in workplace has been identified to significantly increase when tasks are performed at greater trunk velocities (Norman et al., 1998). Previous studies have generally indicated that faster trunk movements reduce trunk strength while imposing greater trunk moments, muscle activities/coactivities and, as a result, spinal loads (Davis and Marras, 2000). A more recent study also reported greater peak moments under faster lifting irrespective of the load and lifting technique (Kingma et al., 2001). In contrast, however, reverse trends (i.e., greater strength whereas smaller moments and muscle activity/coactivity) or no marked differences have also been observed in a number studies when evaluating the effect of movement velocity (Davis and Marras, 2000).

The trunk stability has been estimated empirically to deteriorate as the flexion-extension is performed at a faster pace (Granata and England, 2006). The pelvis was nevertheless restrained in this work that likely inversely influences the free trunk movement and behavior (Gupta, 2001). Moreover, the foregoing deterioration in system stability at faster movements appears out of line with the reported associated increase in muscle activities/coactivities (Davis and Marras, 2000) and the consequent increase in stability (El-Rich et al., 2004; Granata and Orishimo, 2001). Other model investigations of the spinal stability in dynamic tasks have not examined the likely role of alterations in movement velocity.

The effect of changes in velocity of trunk flexion-extension movements on the flexion-relaxation (FR) phenomenon that is recognized as the partial or complete silence in superficial extensor muscles at larger flexion angles (Floyd and Silver, 1951) has also been investigated. Greater flexion-extension movement velocity has been reported either to reduce the frequency of FR observation in repetitive flexion-extension (Mathieu and Fortin, 2000) or to delay its occurrence (Sarti et al., 2001). No effect on the trunk angle at which the FR occurs was, however, found in other studies (Mathieu and Fortin, 2000; Steventon and Ng, 1995).

We have recently developed an iterative active-passive dynamic Kinematics-driven approach that verifies spinal stability and maintains equilibrium equations at all levels and directions (Bazrgari et al., 2007; Bazrgari and Shirazi Adl, 2007). This iterative approach accounts for the nonlinear passive resistance and complex geometry/loading/dynamics of the spine. Crucial effects of proper consideration of global extensor muscles as wrapping elements and of equilibrium equations at all levels rather than a single level have been demonstrated in isometric (Arjmand et al., 2006; Arjmand et al., 2007) and dynamic lifting tasks (Bazrgari and Shirazi Adl, 2007). The goal of current study is set to evaluate the effect of changes in velocity of movement and lumbar rotation during unconstrained flexion-extension tasks on muscle activations, spinal loads and stability. In vivo kinematics of fourteen subjects are initially recorded. The collected data in three of these subjects (with extreme and mean lumbar rotations) are subsequently fed back into the Kinematics-driven model to compute results during the entire cycle of flexion-extension at three different velocities. We hypothesize that alterations in velocity and lumbar rotation during flexion-extension movements substantially influence trunk equilibrium and stability.

4.3 Methods

4.3.1 In vivo measurements

Fourteen healthy males with no recent back complications volunteered for the study after signing an informed consent form approved by relevant committees. The mean (S.D.) age, height, and body mass of the participants were 26 (2.1) years, 180 (7) cm, and 75.3 (10.2) Kg. While bending slightly forward, infrared Light Emitting Diodes (LED) were placed on the skin at the tip of the spinous processes at T5, T7, T10, T12, L1, L5, and S1 levels. Three extra markers were placed on the ilium (left/right iliac crests) and posterior- superior iliac spine for the evaluation of pelvic rotation. A three camera Optotrak system (Northern Digital Inc International, Waterloo, Canada) was used to collect 3D coordinates of skin surface markers. Ground reaction forces were simultaneously recorded using a force-plate (AMTI, Newton, MA, USA).

Each task started from and ended at the standing upright position with flexion or extension phases lasting ~6 sec, 3 sec, and as fast as possible for slow, intermediate and fast paces, respectively. Subjects were also instructed to remain in full flexion for a period of ~3 seconds during slow movement and ~1 second in intermediate one. Each task, hence, lasted in total ~15, 7, and 2-3 seconds for slow, intermediate, and fast trunk movements and was repeated five times by participants. A metronome set at one beat per second was used to assist subjects to control the speed of movement during slow and intermediate velocities. During tests, subjects kept arms extended in gravity direction and knees straight.

4.3.2 Model studies

A sagittaly symmetric T1-S1 spine model with 46 local and 10 global muscles (Fig. 4.1) was used to estimate muscle forces, spinal loads and stability (Arjmand and Shirazi-Adl, 2005, 2006; Bazrgari et al., 2007; Bazrgari and Shirazi Adl, 2007). The nonlinear and direction-dependent mechanical properties of T12–S1 motion segments were represented by deformable beams. The trunk/head/arms mass and mass moments of inertia were assigned at different levels along the spine at their respective gravity centers based on published data (de Leva, 1996; Pearsall et al., 1996; Zatsiorsky and Seluyanov, 1983). Connector elements parallel to deformable beams accounted at each level for the intersegmental damping; translational =1200 Ns/m and angular = 1.2 Nms/rad (Kasra et al., 1992; Markolf, 1970). Measured temporal variation of rotation at pelvis and thorax along with the intervening lumbar vertebral rotation were prescribed as input data into each model. The total lumbar rotation was partitioned in accordance with proportions reported in earlier investigations; ~11%, 15%, 14%, 18%, 22%, and 20% for T12-L1 to L5-S1 levels, respectively (Bazrgari et al., 2007). Temporal variations of pelvic vertical and horizontal translations were applied along with the pelvic tilt into the model at the S1 as base excitations.

To evaluate muscle forces, Kinematics-driven algorithm was employed to solve redundant active-passive system subject to prescribed kinematics, inertia and damping (Bazrgari et al., 2007; Bazrgari and Shirazi Adl, 2007). To resolve the redundancy problem at each level, an optimization approach with the cost function of minimum sum of cubed muscle stresses was used along with inequality equations of muscle forces remaining positive and greater than their passive force components (calculated based on instantaneous muscle length and a tension-length relationship (Davis et al., 2003)) but smaller than the sum of their respective maximum active forces (i.e., 0.6 MPa times muscle's physiological cross-sectional area, PCSA (Winter, 2005)) and passive force components.

Global extensor muscles were constrained to follow a curved path whenever their distances from T12-L5 vertebral centers decreased below 90% of their respective values at undeformed configuration (i.e. reaching 53, 53, 55, 56, 54 and 48 mm for the global longissimus and 58, 56, 56, 55, 52 and 45 mm for the global iliocostalis at T12 to L5 vertebrae, respectively). This wrapping contact mechanism acts to enforce kinematics constraints on deformations preventing the penetration of global muscles into underlying tissues during movements (Arjmand et al., 2006; Shirazi-Adl, 2006; Shirazi-Adl and Parnianpour, 2000).

For the stability analyses, the stiffness of each muscle element, k, was assigned using linear stiffness-force relation (i.e. k=qF/L) in which the muscle stiffness is proportional to the muscle force, F, and inversely proportional to its current length, L, with q as a dimensionless muscle stiffness coefficient taken the same for all muscles. At each instance of time, non-linear analyses were performed to identify the minimum (critical) q value to maintain stability. The stability margin under each q value was further investigated at the loaded deformed configurations by free vibration and linear perturbation analyses. The finite element program ABAQUS (V6.5, ABAQUS Inc., Providence, RI, USA) was used to carry out analyses while the optimization procedure was analytically solved using an in-house program based on Lagrange Multipliers Method (Raikova and Prilutsky, 2001). Implicit algorithm with unconditionally stable Hilber-Hughes-Taylor (Hilber et al., 1978) integration operator was used to solve the nonlinear transient problem. The time step was automatically selected by the solver but was constrained to remain <0.01s.

4.4 Results

Kinematics of three subjects representing minimum ('Min'), maximum ('Max') and almost mean ('Mean') lumbar rotations of all 14 subjects were prescribed into the model. The mean maximum rotations (SD) of trunk, pelvis, and lumbar (in degrees) for all subjects were respectively 113.5 (7), 57.5 (10.2), 56 (6.5) in slow, 113.5 (6.2), 59.1 (10.3), 54.4 (7.3) in intermediate, and 122.6 (5.2), 65.5 (8.7), 57.1 (7.7) in fast movement velocities. The fastest movements lasted ~1.9, 2.2, and 3 seconds in subjects 'Min', 'Max', and 'Mean', respectively, compared to mean of ~2.4 (SD = 0.7) seconds for all subjects. To avoid fluctuations in calculated accelerations, only one representative trial rather than the mean of five trials was considered for each case (Fig. 4.2). Maximum thorax/pelvis angular velocities (°/s) of 257/192, 207/140, and 206/128 were reached in subjects 'Min', 'Mean', and 'Max' respectively (Fig. 4.3). Accelerations on the other hand reached their maximums at full flexion and mid flexion/extension periods (Fig. 4.3).

The significance of inertial forces in fastest movement compared with slower ones was apparent in both computed and measured reaction forces in all three subjects (Fig. 4.3-bottom). Maximum inertial forces of 317 (405), 230 (309) and 104 (118) N were detected in the vertical component of reaction forces by the model at the S1 (and by the force-plate at the ground) for the fastest speed of subjects 'Min', 'Max' and 'Mean', respectively.

To initiate forward flexion at the fastest pace, a burst of abdominal activity at the beginning of the forward flexion was predicted (Fig. 4.4). In contrast, nearly no abdominal activity was estimated at the start of slower tasks (Fig. 4.5). All subjects demonstrated abdominal activity at larger flexion angles in slower movements (Fig. 4.5). At larger flexion angles and in all subjects, global extensor muscles exhibited greater activity in fastest movement whereas FR in slower movements. Forces in local lumbar muscles followed trends similar to those for global muscles. Complete FR in local lumbar muscles (i.e., Fa=0 in Table 4.1) was estimated only at some levels and that in subjects with larger lumbar rotations (i.e., 'Mean' and 'Max').

The computed net external moment at the S1 substantially altered between the fastest movement and the two slower ones (Fig. 4.6). In contrast to the fastest movement, slower movements demonstrated even a decrease in sagittal moment at peak flexion angles. The peak net moment substantially increased by ~ 83%, 22% and 65% from the intermediate velocity to the fastest one in subjects 'Min', 'Mean' and 'Max', respectively. The differences between two slower rates remained, however, negligible.

Spinal loads and net moments increased caudally reaching their maximum at the lowermost L5-S1 level. Spinal loads at different levels were all greater in fastest movement than slower ones (Fig. 4.6 and Table 4.2) with negligible differences in latter slower movements. Extreme flexion angles considered in this study resulted in considerable change in lines of action of global extensor muscles as depicted for the subject 'Max' in Fig. 4.1. The wrapping of global extensor muscles yielded large contact forces at different spinal levels that increased as muscle forces and lumbar rotation increased (Table 4.3).

Stability analysis for subjects 'Min' and 'Mean' under the fastest and intermediate velocities demonstrated that the spine was quite stable in deep flexion (Fig. 4.7); no muscle stiffness was required when the trunk reached forward flexion angles of 19.5° (38°) and 52° (62°) for subjects 'Min' (and 'Mean') under fastest and intermediate paces, respectively.

4.5 Discussion

The iterative transient Kinematics-driven method that accounts for measured kinematics, nonlinear properties of the ligamentous spine, wrapping of global extensor muscles and trunk dynamic characteristics was used to estimate muscle forces, spinal loads and stability in flexion-extension movements at three different velocities. Results confirmed the hypothesis on the crucial role of movement velocity and lumbar rotation on response dynamics, muscle activation, FR, internal spinal loads and trunk stability.

4.5.1 Methodological issues

Due to errors involved in evaluation of segmental rotations (Lundberg, 1996; Shirazi-Adl, 1994), recorded data were used to calculate pelvic tilt and trunk rotation with the difference partitioned between intervening lumbar segments using proportions reported in the literature (Bazrgari et al., 2007). The measurement of linear variation of segmental rotations in flexion-extension movements (Wong et al., 2004) supports the constant proportion taken throughout the motion. Speed of movement has neither been found to significantly influence vertebral rotations (Zhang et al., 2003).

In this study, rather than simulating one single subject on one hand or all 14 subjects on the other, it was decided to take three subjects with extreme and mean peak lumbar rotations. In this manner in accordance with our stated objectives and with a reasonable amount of effort, effects of variations in velocity of movement and lumbar rotation on results were considered. Changes in the lumbar rotation are crucial in terms of the relative contribution of passive ligamentous spine and musculature as well as the wrapping of global extensor muscles. The geometry and upper trunk mass were not, however, changed from a subject to another as the T5-S1 height and total mass of three subjects matched relatively well the values taken in the model; 389 mm, 80 kg for the subject 'Max', 387 mm, 69 kg for the subject 'Mean' and 385 mm, 74 kg for the subject 'Min' as compared with 384 mm, 74 kg used in the model.

A maximum of 10% reduction in the lever arm of global extensor muscles during forward flexion was based on earlier studies (Jorgensen et al., 2003; Macintosh et al., 1993). The extent of this reduction would likely depend on the posture (Arjmand et al., 2006). The computed muscle forces were partitioned, at the post-processing phase, into passive and active components using a passive tension-length relationship (Davis et al., 2003). For normalization, the maximum allowable muscle stress of 0.6 MPa was assumed for all muscles. The predicted activity in muscles (Figs. 4.4 and 4.5) could substantially be influenced by assumptions both on the maximum muscle stress (in terms of its absolute magnitude and likely variations with changes in muscle length and velocity) and on the passive force-length relationship. The damping values at inter-segmental levels assigned based on measurements of disc-body-disc units was not sufficient to attenuate the high frequency fluctuations (i.e. noise content) in required moments during fastest movements. They were, hence, increased by five-fold while accounting also for additional damping of other abdominal tissues surrounding the spine. Another subsequent four-fold increase was verified not to influence predictions.

Changes in ligamentous properties under various movement rates are not expected to be substantial in view of earlier studies on the effect of changes in loading rate on viscoelastic properties of the spine (Neumann et al., 1994; Wang et al., 2000). The computed results were found unchanged with continuous refinement of mesh and much smaller time increments (i.e. 0.1 milliseconds). Since the damping matrix was symmetric and positive definite, the static criterion of stability at stressed-deformed configurations holds (Leipholz, 1970). Static perturbation and frequency analyses, as expected, yielded identical critical muscle stiffness coefficient values.

4.5.2 Implications

Velocity of trunk motion has been reported to have considerable effects on net moments, muscle activity and spinal loads (Davis and Marras, 2000) though the increase in net moments during faster movements is less obvious than that in internal spinal loads. Our predictions indicate that the peak compression/shear forces and net moments substantially increase from the slower movements to the fastest one. The temporal variation of spinal loads and net moments clearly indicate that the computed peak values, irrespective of the velocity of movement, are subject dependent and may not occur at the time of peak trunk flexion. In other words, the foregoing relative effects of velocity of movement on internal loads and net moments could substantially diminish or even reverse when considering the subject 'Mean' at the instance of peak trunk flexion angle. This is due to the kinematics profile of this subject in the fastest movement (Fig. 4.2) that rapidly attained his peak lumbar rotation and then preserved it while the pelvic and trunk rotations reached their maximum values and began to reverse for the extension phase of the movement both together. This observation may in part help understand the existing controversy in the published literature.

The temporal variation of the computed axial reaction force at the base (S1) demonstrates the dominant effect of inertia in the fastest movement (Fig. 4.3). Sharp initial downward acceleration followed by a deceleration when reaching the peak trunk flexion in the forward flexion phase plus the initial upward acceleration and subsequent deceleration to reach back to the upright position in the extension phase results in the pattern of reaction force at the base. The predicted temporal variation of reaction forces in all three subjects at three velocities agree well with measurements bearing in mind that the latter values account also for the inertia of lower body. Due to the extended knee position during movements, the measured inertial effects should nevertheless be attributed primarily to the upper body. The computed results are also in good agreement with reported ground reaction forces measured while lifting weights at different speeds (Dolan et al., 1999).

Considering the reported strength of the lumbar motion segments of ~4-6 kN in compression (Jager and Luttmann, 1991) that may reduce due to fatigue (Dolan et al., 1994), ~1-2 kN in shear (Cyron et al., 1976; Miller et al., 1986) and ~70 Nm in flexion moment (Miller et al., 1986; Osvalder et al., 1993), the results of this study highlights the risk of flexion-extension movements when performed at the fastest pace to full voluntary trunk flexion. Although the presence of external loads in hands, not considered in this work, could further increase the internal loads and hence the risk of injury but it is highly unlikely that subjects voluntarily carry out the flexion-extension movement to the maximum flexion angle as fast when they carry loads in hands.

Velocity of movement influences the spinal stability depending on the trunk flexion angle. Under the same trunk flexion angle, the fastest movement yields overall the most stable configuration (Fig. 4.7). At larger trunk angles and in agreement with our earlier simulations (Arjmand and Shirazi-Adl, 2006), the spine is much more stable with no need for a muscle stiffness. The spinal stability improves under the fastest movement at the beginning of the task that could be partly due to activity in the agonist abdominal muscles and antagonist passive extensor muscles. In order to further increase spinal stability and decrease critical q values in the neighborhoods of neutral standing position, one should introduce (or increase) coactivity in the model (El-Rich et al., 2004; Granata and Orishimo, 2001).

Partial or full relaxation in global extensor muscles is estimated in slower movements in deeper flexion angles. Due to large lumbar rotations on the other hand, abdominal activity is also computed in deep flexion. The reason for this shift of activity between global muscles lies in the growing passive resistance of global extensor muscles in deeper flexion that exceeds that required by external moments thereby activating abdominal muscles. The activity of abdominal muscles during FR has been reported (Gupta, 2001; Kippers and Parker, 1984; Olson et al., 2006). The reduction in global extensor muscle activity in deep flexion during slower movements is in agreement with some earlier works (Gupta, 2001; Mathieu and Fortin, 2000; Olson et al., 2006; Sarti et al., 2001) but in contrast to few others that suggest no reduction in thoracic extensor muscle activity (Toussaint et al., 1995). As for local lumbar muscles, although a general reduction in activity was estimated, the complete relaxation was predicted at few levels and only in subjects 'Mean' and 'Max' (Table 4.1). Using superficial EMG data, a number of studies have confirmed the reduction in lumbar extensor activity in deep flexion (Shin et al., 2004; Toussaint et al., 1995). Andersson et al (1996), using deep wire electrodes, reported silence only in superficial lumbar extensor muscles with activity remaining in deeper ones.

Finally, results demonstrated the marked effect of alterations in movement velocity on muscle activity, spinal loads, base reaction force and stability. Moreover, results were equally influenced by changes in the peak lumbar rotation and its temporal variation during movements.

4.6 Acknowledgements: The work is supported by grants from the NSERC-Canada and Aga Khan Foundation.

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Table 4.1 Active (Fa), passive (Fp) forces in local lumbar muscles at peak flexion (N)

Muscle	Att.		Min			Mean			Max		
			S	M	F	S	M	F	S	M	F
IP	L1	Fp	0	0	0	0	0	0	0	0	0
		Fa	***93***	***119***	***151***	***46***	***15***	***151***	***13***	***19***	***140***
LGpl	L1	Fp	12	10	9	12	12	14	6	8	20
		Fa	***10***	***11***	***33***	***0***	***0***	***0***	***0***	***0***	***22***
	L2	Fp	13	12	9	16	15	17	8	10	24
		Fa	***17***	***19***	***35***	***0***	***0***	***0***	***0***	***0***	***30***
	L3	Fp	18	16	12	21	21	24	16	17	35
		Fa	***13***	***16***	***34***	***0***	***0***	***0***	***0***	***0***	***27***
	L4	Fp	22	20	15	23	22	27	17	19	44
		Fa	***12***	***15***	***45***	***0***	***0***	***0***	***0***	***0***	***22***
	L5	Fp	18	16	11	27	28	34	45	42	34
		Fa	***58***	***61***	***151***	***35***	***35***	***43***	***24***	***27***	***112***
ICpl	L1	Fp	19	17	14	20	19	23	10	13	33
		Fa	***16***	***18***	***53***	***0***	***0***	***0***	***0***	***0***	***34***
	L2	Fp	32	30	24	36	34	39	18	23	63
		Fa	***34***	***40***	***77***	***0***	***0***	***0***	***0***	***0***	***61***
	L3	Fp	42	38	30	49	48	55	37	41	84
		Fa	***32***	***39***	***78***	***0***	***0***	***0***	***0***	***0***	***61***
	L4	Fp	46	41	33	54	53	64	40	45	95
		Fa	***36***	***44***	***113***	***0***	***0***	***0***	***0***	***0***	***62***
MF	L1	Fp	24	22	17	23	21	26	11	14	44
		Fa	***15***	***18***	***57***	***0***	***0***	***0***	***0***	***0***	***30***
	L2	Fp	33	30	24	41	39	45	20	27	61
		Fa	***45***	***51***	***83***	***0***	***0***	***0***	***0***	***0***	***79***
	L3	Fp	46	41	32	70	72	86	63	70	90
		Fa	***80***	***88***	***150***	***14***	***11***	***8***	***0***	***0***	***127***
	L4	Fp	32	29	22	49	50	60	51	57	64
		Fa	***69***	***76***	***158***	***18***	***15***	***19***	***0***	***0***	***134***
	L5	Fp	13	11	9	20	21	25	33	31	26
		Fa	***104***	***107***	***244***	***77***	***76***	***95***	***73***	***77***	***201***
QL	L1	Fp	9	8	6	14	14	17	7	9	18
		Fa	***18***	***19***	***45***	***2***	***0.7***	***1***	***0***	***0***	***33***
	L2	Fp	10	9	6	14	13	15	7	8	21
		Fa	***15***	***18***	***32***	***0***	***0***	***0***	***0***	***0***	***26***
	L3	Fp	12	11	7	13	12	15	10	10	25
		Fa	***7***	***9***	***21***	***0***	***0***	***0***	***0***	***0***	***13***
	L4	Fp	13	12	8	17	11	13	11	9	27
		Fa	***4***	***5***	***22***	***0***	***0***	***0***	***0***	***0***	***6***

Table 4.2 Computed maximum internal spinal loads at the L5-S1 disc mid-height plane in local directions (N) (These peak values could occur at different times; C: Compression (N), S: Shear (N), M: Moment (Nm))

	Slow			Intermediate			Fast		
	C	S	M	C	S	M	C	S	M
Subject 'Min'	2755	881	26.8	2891	933	25.6	4599	1515	28
Subject 'Mean'	2899	828	31.9	2974	846	32.7	3870	1066	34.8
Subject 'Max'	3017	752	38.3	3065	783	38.8	4950	1247	40.9

Table 4.3 Computed maximum wrapping contact forces at different lumbar levels (N) (These peak values could occur at different times)

Wrapping Forces		T12	L1	L2	L3	L4	L5
Subject 'Min'	Slow	15	91	80	75	95	140
	Intermediate	16	82	72	67	85	126
	Fast	30	211	184	161	214	317
Subject 'Mean'	Slow	13	147	129	123	151	220
	Intermediate	13	153	136	130	162	233
	Fast	12	189	167	159	196	283
Subject 'Max'	Slow	12	256	231	220	268	380
	Intermediate	12	242	218	208	255	362
	Fast	11	218	200	190	251	347

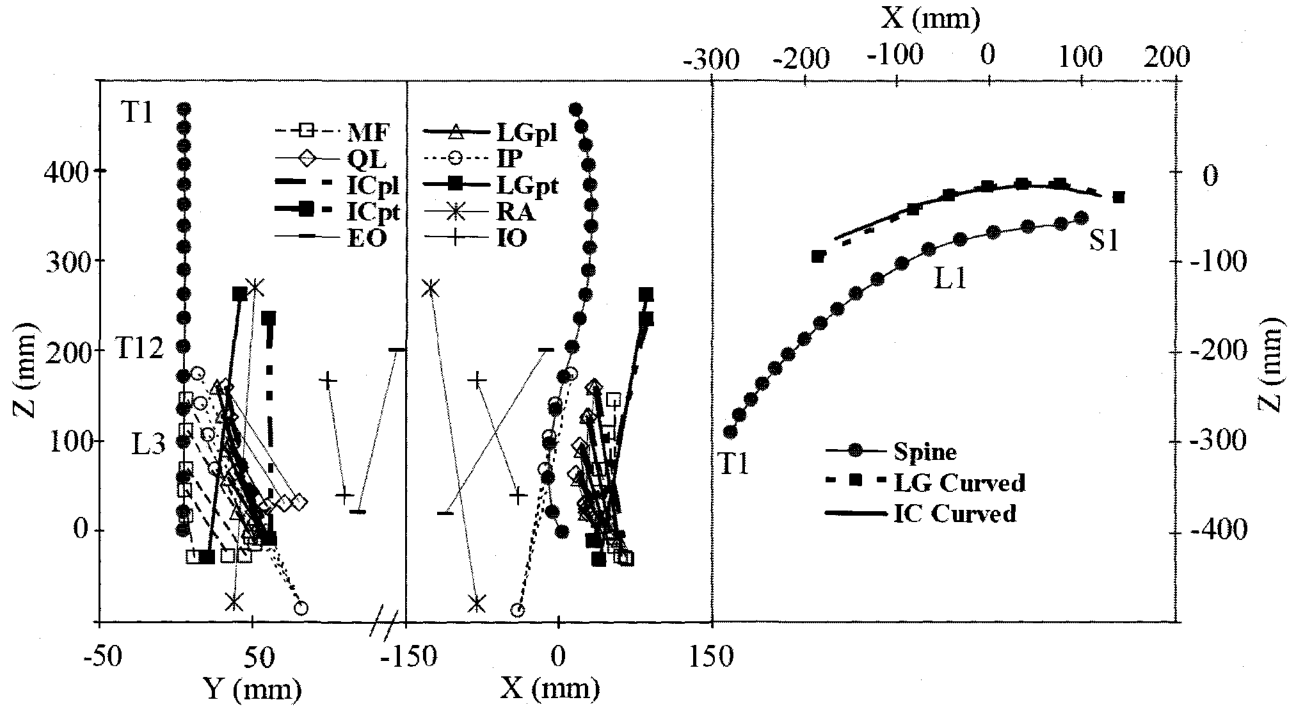

Figure 4.1 Representation of the spine including global and local musculatures in the sagittal (on the right) and frontal (at the middle, fascicles on one side are shown) planes. Deformed configuration (subject 'Max') at the instance of full flexion along with wrapping (curved) global extensor muscles are depicted on the right. ICpl: Iliocostalis lumborum pars lumborum, ICpt: Iliocostalis lumborum pars thoracic, IP: Iliopsoas, LGpl: Longissimus thoracis pars lumborum, LGpt: Longissimus thoracis pars thoracic, MF: Multifidus, QL: Quadratus lumborum, IO: Internal oblique, EO: External oblique, and RA: Rectus abdominus.

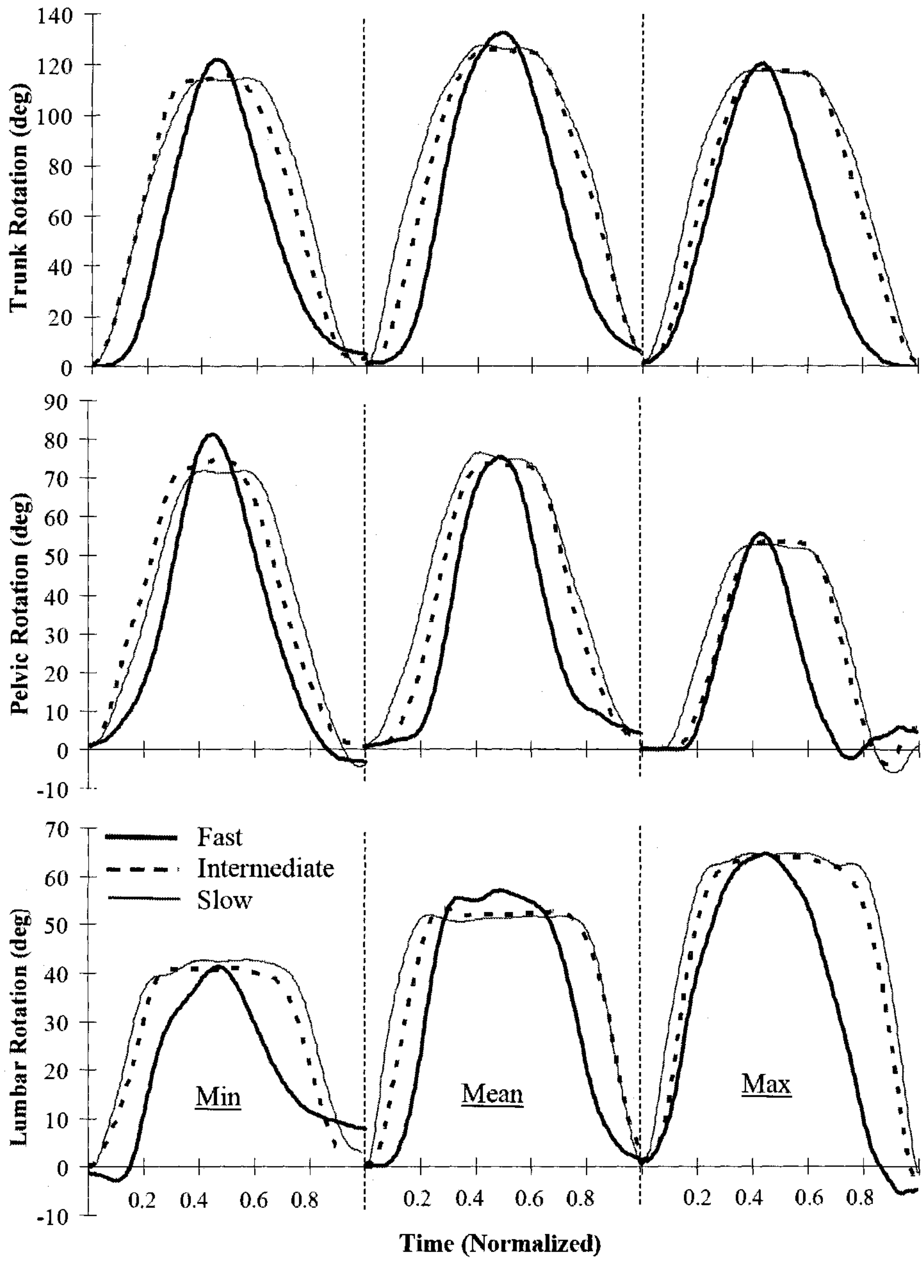

Figure 4.2 Prescribed thorax (top), pelvis (middle), and lumbar (bottom) rotations in the model for three subjects 'Min' (left), 'Mean' (middle), and 'Max' (right) at three movement velocities based on in vivo measurements. The lumbar rotations are subsequently partitioned in between different levels based on proportions given in the text.

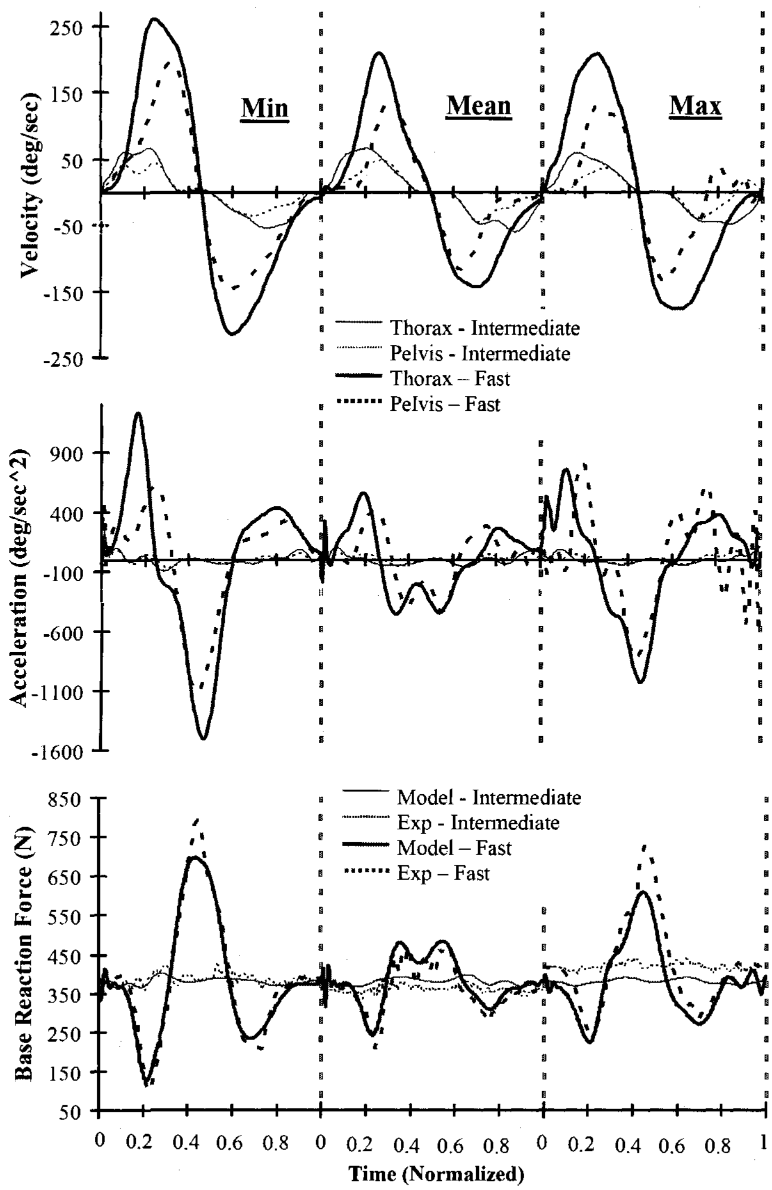

Figure 4.3 Temporal variation of Thorax/pelvis angular velocity (top) and angular acceleration (middle) along with the time history of predicted base reaction force and measured ground reaction force in the gravity direction (bottom) for three subjects 'Min' (left), 'Mean' (middle), and 'Max' (right) at two velocities. Results at the slowest movement velocity shows even less temporal variations than those at intermediate velocity and are, hence, not shown. It should be noted that the base reaction force is completely different from the internal spinal compression force presented in Fig. 4.6.

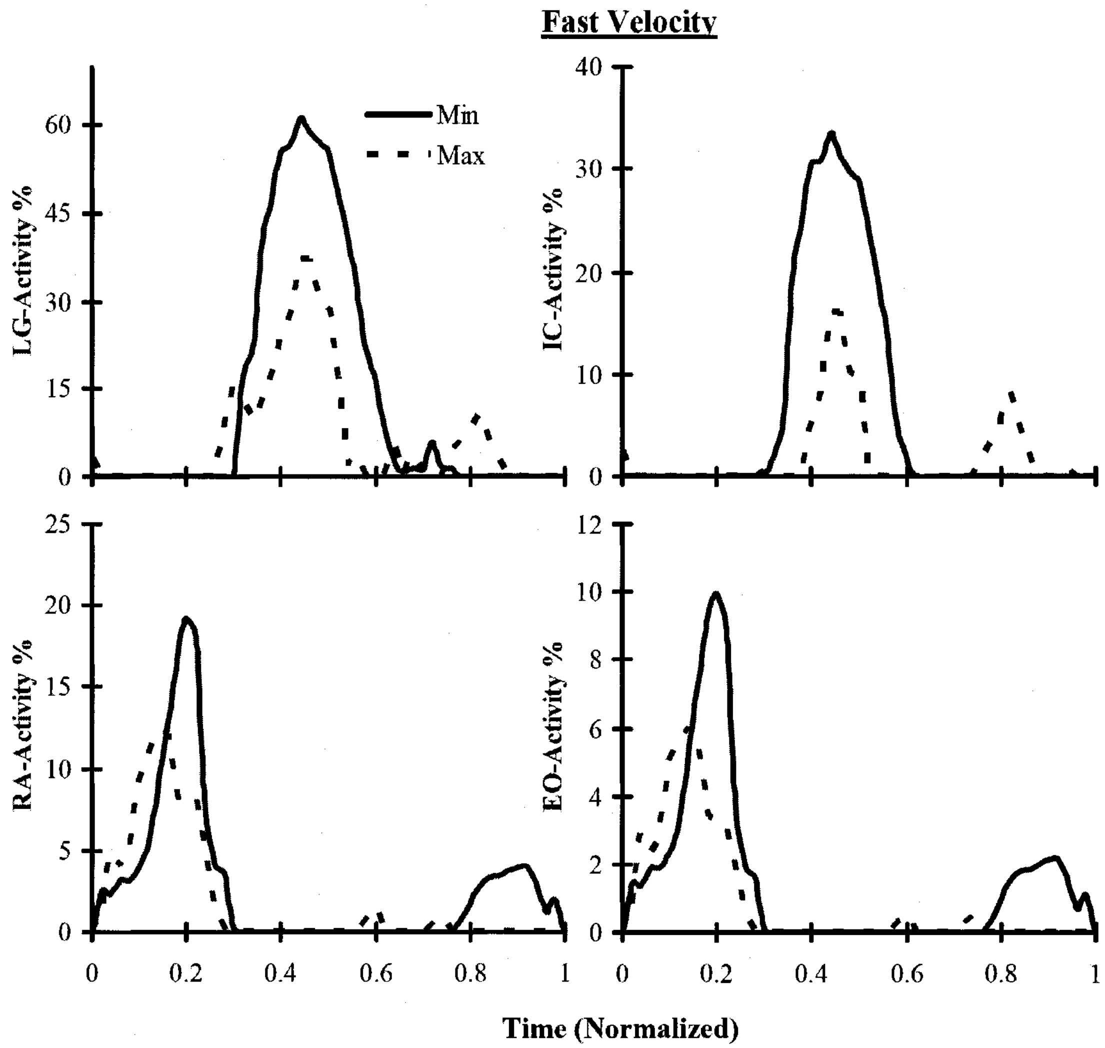

Figure 4.4 Predicted temporal variation of muscle activity for the subjects 'Min' and 'Max' at the fastest movement velocity. LG: Longissimus thoracis pars thoracic, IC: Iliocostalis lumborum pars thoracic, EO: External oblique, and RA: Rectus abdominus. Normalization in estimated muscle forces is done here just for the sake of presentation as it does not account for length and velocity effects.

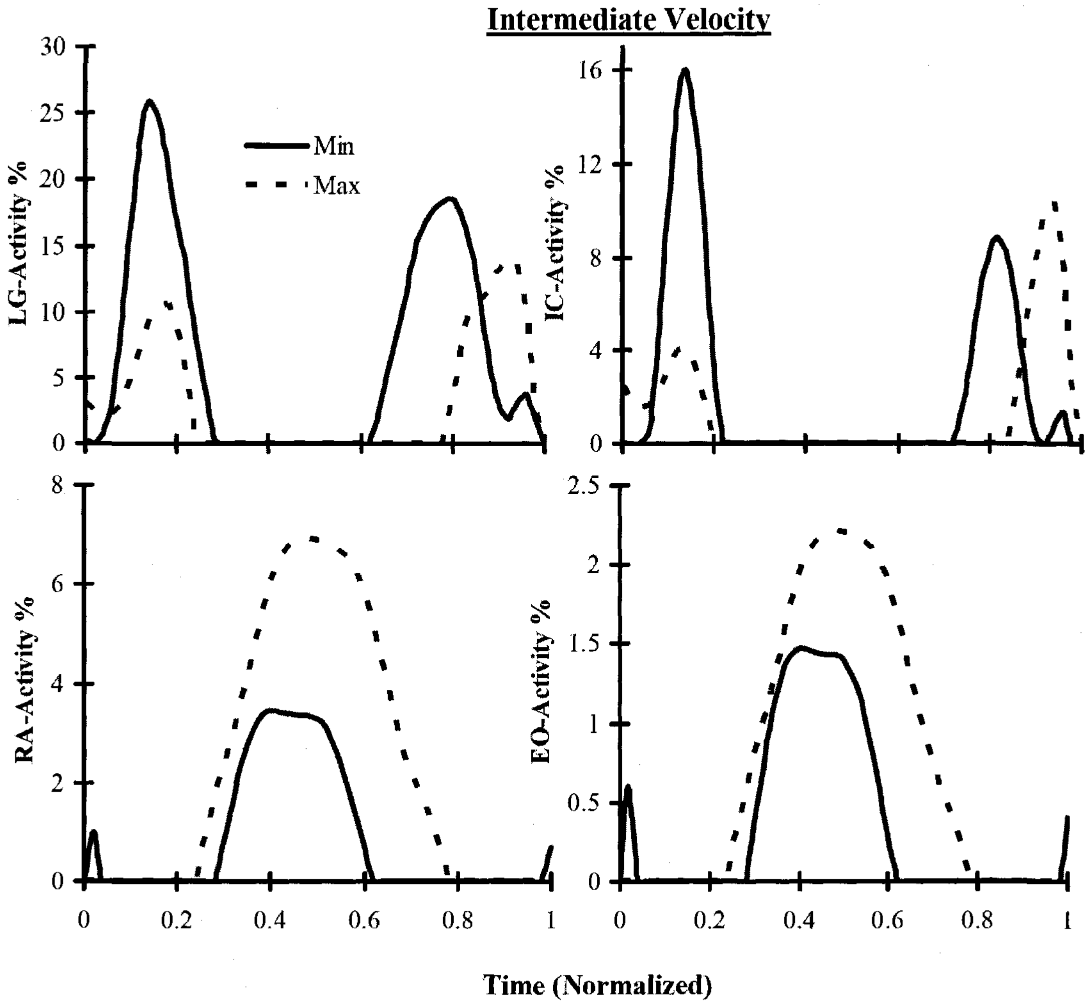

Figure 4.5 Predicted temporal variation of muscle activity for the subjects 'Min' and 'Max' at the intermediate movement velocity. LG: Longissimus thoracis pars thoracic, IC: Iliocostalis lumborum pars thoracic, ES: Erector spinae, EO: External oblique, and RA: Rectus abdominus. Normalization in estimated muscle forces is done here just for the sake of presentation as it does not account for length and velocity effects.

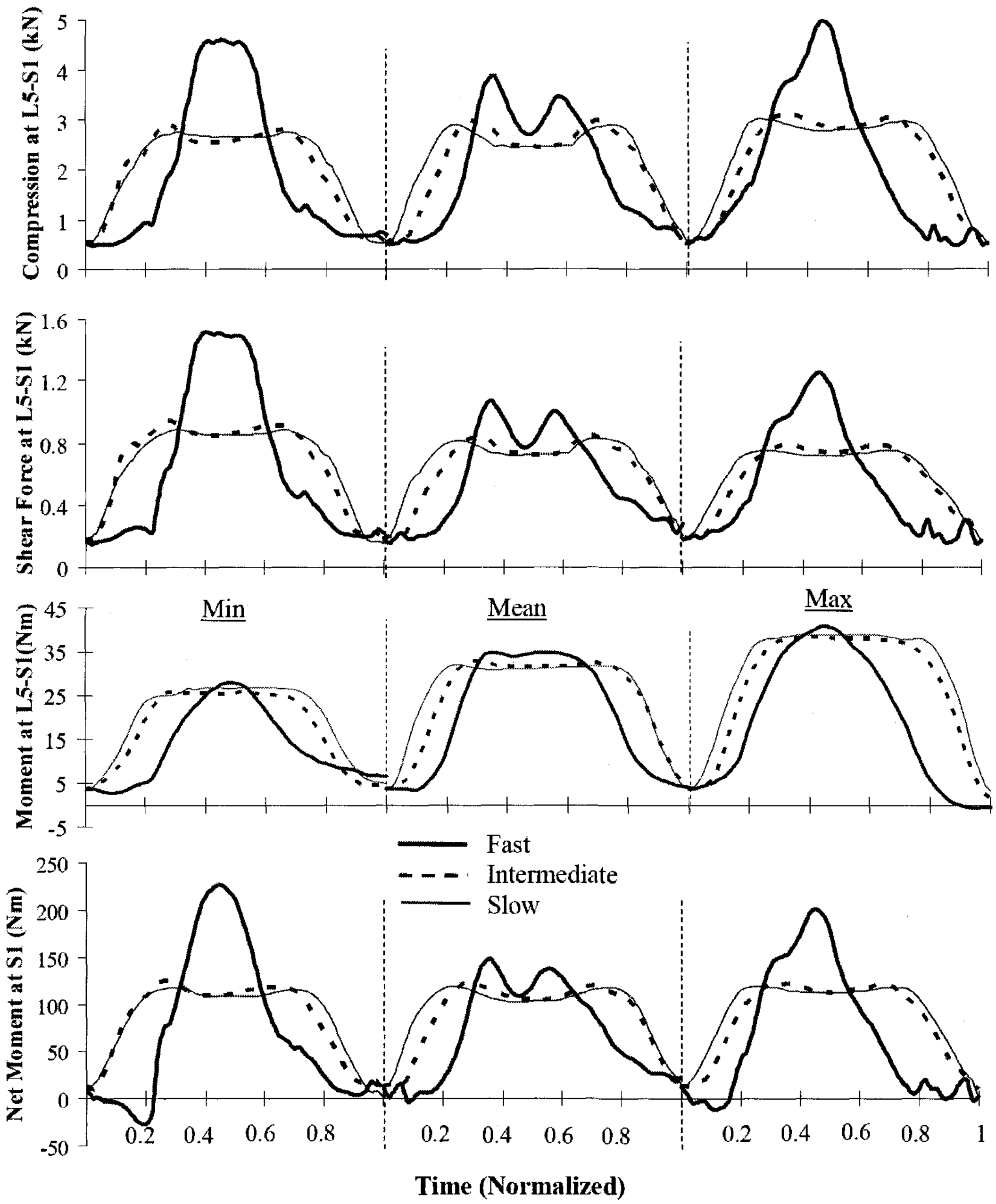

Figure 4.6 Predicted temporal variation of internal spinal forces at the L5-S1 disc mid-height in local directions (compression force, shear force and sagittal moment) and the net moment at the S1 (to be resisted by all muscles and passive ligamentous spine) at the bottom for three subjects 'Min' (left), 'Mean' (middle), and 'Max' (right) at different movement velocities.

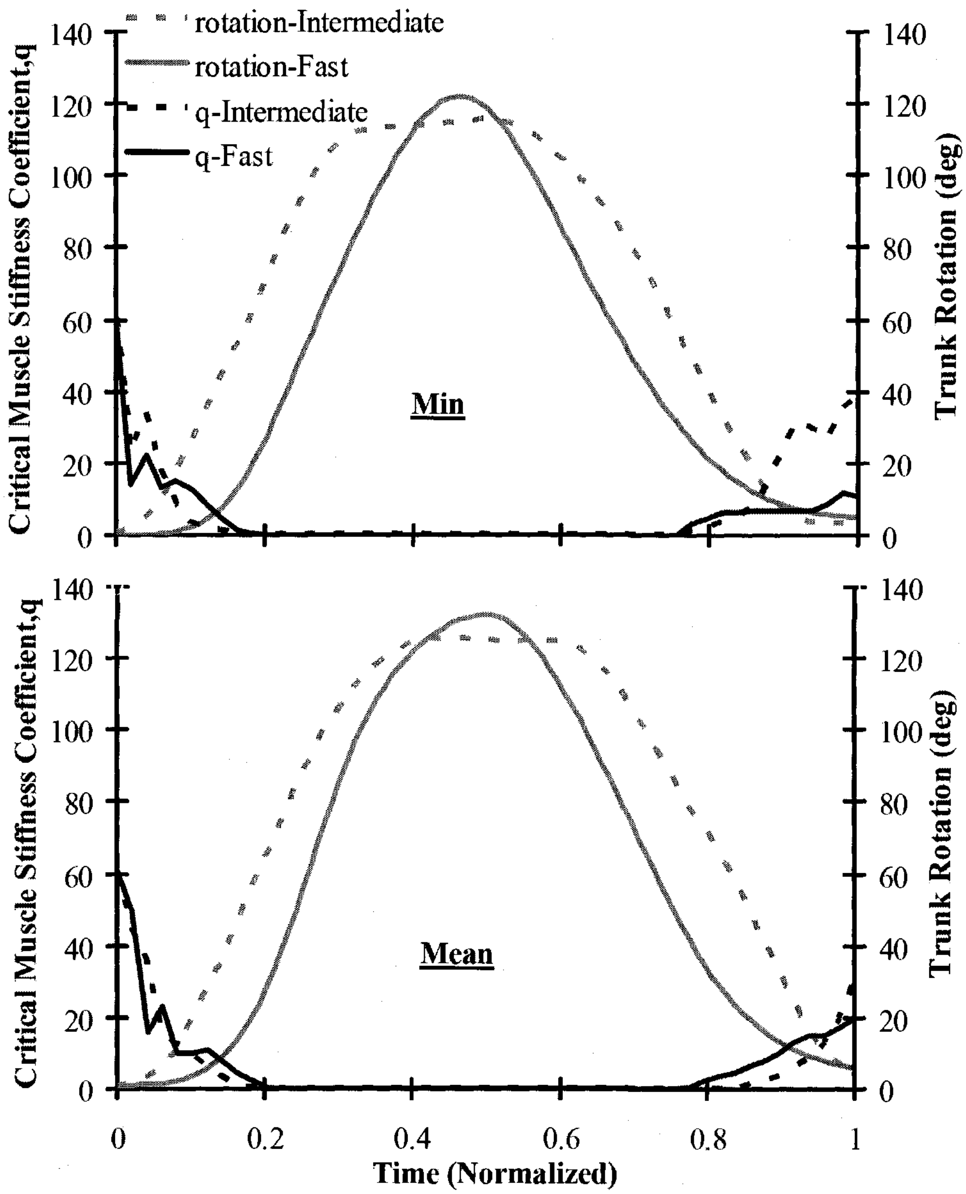

Figure 4.7 Predicted temporal variation of minimum (critical) muscle stiffness coefficient, q, required for the trunk stability for subjects 'Min' (top) and 'Mean' (bottom) at two different flexion-extension movement velocities. Lower q values indicate higher trunk stability and q=0 suggests that no muscle stiffness is needed to stabilize the trunk (although existing muscle forces stiffen the spine and contribute to its stability). The trunk rotations are also shown in these figures.

CHAPTER FIVE

(Article-IV)

COMPUTATION OF TRUNK MUSCLE FORCES, SPINAL LOADS AND STABILITY IN WHOLE BODY VIBRATION

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5.1 Abstract

Whole body vibration has been indicated as a risk factor in back disorders. Proper prevention and treatment managements, however, requires a sound knowledge of associated muscle forces and loads on spine. Previous trunk model studies have either neglected or over-simplified the trunk redundancy with time-varying unknown muscle forces. Trunk stability has neither been addressed. A novel iterative dynamic Kinematics-driven approach was employed to evaluate muscle forces, spinal loads and system stability in a seated subject under a random vertical base excitation with ~ ±1 g acceleration shock contents. This iterative approach satisfied equations of motion in all directions/levels while accounting for the nonlinear passive resistance of the ligamentous spine. The effect of posture, co-activity in abdominal muscles and changes in buttocks stiffness were also investigated. The computed vertical accelerations were in good agreement with measurements. The input base excitation, via inertial and muscle forces, substantially influenced spinal loads and system stability. The flexed posture in sitting increased the net moment, muscle forces and passive spinal loads while improving the trunk stability. Similarly, the introduction of low to moderate antagonistic coactivity in abdominal muscles increased the passive spinal loads and improved the spinal stability. A trade-off, hence, exists between lower muscle forces and spinal loads on one hand and more stable spine on the other. Base excitations with larger acceleration contents substantially increase muscle forces/spinal loads and, hence, the risk of injury.

5.2 Introduction

Low back pain (LBP) is the leading musculoskeletal disorder in terms of cost and work-absenteeism (Kittusamy and Buchholz, 2004). Long-term occupational exposure to whole body vibration (WBV) is reported to increase risk of lumbar spine disorders (Bovenzi, 2006; Mansfield, 2005b; Wilder and Pope, 1996). Sedentary working environment with whole body vibration exposure and/or awkward postures increase the risk of Low back pain up to four-fold as compared to sitting alone (Lis et al., 2007). Vehicle seat vibrations with high acceleration content likely cause more back injury than those with low vibration levels that contribute more to the time averaged measures of exposure defined in ISO 2631 (Stayner, 2001). The direct casual association between whole body vibration and Low back pain has, however, been questioned by some investigators (Gallais and Griffin, 2006; Lings and Leboeuf-Yde, 2000; Stayner, 2001) suggesting that, on the basis of existing literature, it is not possible to confirm whether whole body vibration exposure alone or in combination with other factors should be considered as a risk factor. Based on the premise that excessive spinal loads increase risk of back injuries, a sound risk assessment along with effective prevention, treatment, and rehabilitation programs of spinal disorders depend directly on accurate evaluation of muscle forces and spinal loads. A clearer picture of the causal role of whole body vibration environments in back disorders can thus emerge following improved understanding on associated trunk biodynamics. Since spinal loads cannot be measured directly in-vivo, biomechanical models are recognized to play indispensable role in spinal pathomechanics.

Measured driving point frequency response (impedance, apparent mass) and transmissibility functions between seat input and response at different spinal levels have led to the development of rather simple biomechanical whole body vibration models (Griffin, 2001; Mansfield, 2005a; Mansfield, 2005b; Robinson, 1999). Kitazaki and Griffin (1997a) developed a passive sagittaly symmetric finite element model of the upper body simulating the spine, viscera, head, pelvis and buttocks using beam, spring and mass elements. Such models can predict the overall passive response under various vibration and postural conditions and may be used to improve vehicle suspension design. The muscle forces and spinal loads cannot, however, be estimated using such deterministic models. A two dimensional dynamic finite element model of the lower lumbar vertebrae was developed by Pankoke et al. (1998). Under applied static (including a constant extensor muscle force that depended on the posture) and dynamic loads, spinal loads at lower lumbar levels were computed. A head to sacrum finite element model including the entire spinal column and the rib cage was used by Kong and Goel (2003) to study the trunk resonant frequency and transmissibility under base vertical vibration. The muscles were modeled deterministically as tension-only truss elements with a constant elastic modulus of 1.0 MPa. They reported the first vertical natural frequency in the range of 6.8-8.9 Hz depending on the muscle tension and gravity preload. Other 3D dynamic models have evaluated the trunk whole body vibration response under constant muscle forces that were estimated a-priori using a static analysis with an optimization approach (Buck and Wolfel, 1998; Pankoke et al., 2001).

The foregoing whole body vibration studies of the trunk have either neglected or over-simplified the redundancy in the trunk musculoskeletal system in which the spinal loads depend directly on unknown muscle forces that alter during whole body vibration period. The importance of a proper estimation of muscle forces in quantification of spinal loads in whole body vibration has been emphasized (Bluthner et al., 2002; Seidel, 2005). The dynamic stability of the spine in whole body vibration studies has also been overlooked. The trunk stability is maintained by activation in muscles as well as passive muscle/spinal stiffness properties and is influenced by changes in the posture, passive properties and load magnitude/height (Arjmand and Shirazi-Adl, 2005, 2006; Bazrgari and Shirazi Adl, 2007; El-Rich and Shirazi-Adl, 2005; Granata and Orishimo, 2001).

In continuation of our earlier isometric (Arjmand and Shirazi-Adl, 2005, 2006; El-Rich et al., 2004) and transient (Bazrgari et al., 2007; Bazrgari and Shirazi Adl, 2007) investigations of the trunk biomechanics using the iterative Kinematics-driven finite element approach, the spinal loads, trunk muscle forces and trunk stability under a random vertical base excitation is studied. The model accounts for nonlinear load- and direction-dependent properties of lumbar motion segments, complex geometry of spine, detailed muscle architecture, dynamic characteristics of the trunk, and wrapping of global extensor muscles. The input base excitation is taken from the literature (Seidel et al., 1997). It is hypothesized that high magnitude (shock) acceleration content in vehicular vibrations increases the risk of tissue injury by generating loads in the neighborhood of safe threshold values.

5.3 Methods

5.3.1 Kinematics-driven approach

This approach exploits kinematics data to generate additional equations at each spinal level in order to alleviate the kinetics redundancy in the system. Initially, measured trunk kinematics (sagittal plane rotations at different vertebral levels and base vertical acceleration in this study) along with external/gravity loads are prescribed into a nonlinear finite element model of the thorocolumbar spine (Fig. 5.1). Implicit algorithm with unconditionally stable Hilber-Hughes-Taylor integration operator (Hilber et al., 1978) is used to solve the nonlinear transient problem, resulting in the time variation of reaction moments at each vertebral level to be balanced by muscles attached to that level. To resolve the remaining redundancy at each level, an optimization approach with the cost function of minimum sum of cubed muscle stresses is used. The inequality equations relate to unknown muscle forces remaining positive and greater than their passive force components (calculated based on instantaneous muscle length and a tension-length relationship (Davis et al., 2003)) but smaller than the sum of their respective maximum active forces (i.e., 0.6 MPa times muscle's physiological cross-sectional area, PCSA (Winter, 2005)) and the passive force components is also considered. At the end of each iteration, the penalty of muscle forces in shear and axial directions is applied along with the external loads to the spine and the procedure is repeated until the convergence is achieved (i.e. calculated muscle forces in two successive iterations remain almost the same).

Once the muscle forces are calculated throughout the vibration period, the system stability is investigated by replacing muscles with uniaxial elements. The stiffness of each uniaxial element, k, is assigned using linear stiffness-force relation (i.e. $k=q\ F/L$) in which the muscle stiffness is proportional to the instantaneous muscle force, F, and inversely proportional to its current length, L, with q as a dimensionless muscle stiffness coefficient that is taken to be the same for all muscles (Bergmark, 1989). At each instance of time, the stability margin under different q values is investigated at the loaded deformed configurations by natural frequency and linear perturbation analyses. In general, to assess the stability of a nonlinear system using the static stability criterion (i.e., divergence type), one can use linear buckling, perturbation or free vibration analyses at a deformed stressed configuration evaluated based on a prior nonlinear analysis (Leipholz, 1970). In perturbation analysis, the translation of T1 vertebra under application of a unit load is used for different q values to identify the minimum (critical) q. On the other hand, smallest natural frequency of structure, determined using free vibration analysis, is also used as an indication of structural stability (i.e., system becomes unstable as the smallest natural frequency approaches zero). Finite element program ABAQUS (ABAQUS Inc. Version 6.5) is used to carryout nonlinear and linear stability analyses while the optimization procedure is analytically solved using an in-house program based on Lagrange Multiplier Method.

5.3.2 Finite element model

A sagittally symmetric head-pelvis model made of six nonlinear deformable beams to represent T12-S1 segments and seven rigid elements to represent head-T12 (as a single body) and lumbosacral (L1-S1) vertebrae is used (Arjmand and Shirazi-Adl, 2005, 2006; Bazrgari et al., 2007; Bazrgari and Shirazi Adl, 2007). The beams represent the overall nonlinear stiffness of T12-S1 motion segments (i.e., vertebrae, disc, facets and ligaments) at different levels with nonlinear axial compression-strain and sagittal/lateral/axial moment-curvature relations. The nonlinear load-displacement response in different directions along with flexion versus extension differences are represented based on numerical and measured results of previous single- and multi-motion segment studies (Oxland et al., 1992; Pop, 2001; Shirazi-Adl et al., 2002; Yamamoto et al., 1989). The flexural rigidity of the model depends also on the axial compression as reported recently (Shirazi-Adl, 2006). Trunk/head/arms/pelvis mass and mass moments of inertia are assigned at different levels along the spine at their respective gravity centers based on published data (de Leva, 1996; Pearsall et al., 1996; Zatsiorsky and Seluyanov, 1983) (Table 5.1). Connector elements parallel to deformable beams are added to account for the inter-segmental damping using measured values (Kasra et al., 1992; Markolf, 1970) where translational damping =1200 Ns/m and angular damping = 1.2 Nms/rad. Buttocks at the base is modeled by a connector element (compression only) with nonlinear stiffness defined based on reported data in the literature (Table 5.2) (Aimedieu et al., 2003; Kitazaki and Griffin, 1997a) and damping similar to that of lumbar segments.

5.3.3 Muscle architecture

A sagittally symmetric muscle architecture with 46 local (attached to the lumbar vertebrae) and 10 global (attached to the thoracic cage) muscles is used (Fig. 5.2). To simulate wrapping of global muscles (i.e., longissimus thoracis pars thoracic and iliocostalis lumborum pars thoracic), they are constrained to follow curved paths whenever their distances from T12-L5 vertebral centers decrease below 90% of their respective values at undeformed configuration (i.e., to reach the limit values of 53, 53, 55, 56, 54 and 48 mm for the global longissimus and 58, 56, 56, 55, 52 and 45 mm for the global iliocostalis at T12 to L5 vertebrae, respectively). This wrapping mechanism, similar to that formulated in our earlier simulations (Shirazi-Adl, 1989, 2006; Shirazi-

Adl and Parnianpour, 2000), is considered in order not to allow the line of action of these muscles approach unrealistically close to the vertebrae while at the same time simulating a maximum of 10% reduction in their lever arms as observed during forward flexion tasks (Jorgensen et al., 2003; Macintosh et al., 1993).

5.3.4 Input excitations and parametric studies

The input seat vertical acceleration at the base in the model is chosen based on measured un-weighted acceleration at the seat-driver interface of a hydraulic excavator (Seidel et al., 1997) (Fig. 5.3). Seidel et al. (1997) have measured and reported trunk response and seat-subject interaction of a subject under the same base excitation at three different postures (i.e. driving, bent forward, and erect). In accordance with these measurements, two different lumbar postures are considered in current model study that remain unchanged during the vibration (Table 5.3); an erect posture as the reference case (El-Rich et al., 2004) and a flexed posture in which the lumbar lordosis is flattened at all levels by a total of 10 degrees (Black et al., 1996; Lord et al., 1997) to simulate a relax sitting posture (i.e. slouch posture). To evaluate the effect of antagonistic co-activity in abdominal muscles on load distribution between active and passive systems and trunk stability, a-priori low to moderate abdominal coactivity levels of 2%, 1%, and 0.5% (Arjmand et al., 2007) are considered in the IO (Internal Oblique), EO (External Oblique), and RA (Rectus Abdominus) and analyses are repeated for the case with erect posture. Finally, effect of changes in the axial stiffness of the spring at the base simulating buttocks (i.e., softening as much to yield deflections of ~15 mm and stiffening as much to simulate a rigid base) on trunk response to base excitation as well as system natural frequency is also investigated.

5.4 Results

The input vertical acceleration at the base varied primarily in the range of ±5 ms^{-2} with rms value of 1.4 ms^{-2} and two relatively sharp peaks in input signals (i.e. -12 and +10 ms^{-2}) (Fig. 5.3). The computed vertical accelerations at the L3 level and head as well as base reaction force were in general agreement with measured values (Fig. 5.4) (Seidel et al., 1997). The effect of posture on computed accelerations was found to be negligible. The net moment at the S1 base remained almost positive during the entire period and was greater in the flexed posture than in the erect one (Fig. 5.5). This net moment is generated by trunk gravity and inertia forces above the S1 level and is balanced by passive ligamentous structure and muscle forces. The contribution of passive ligamentous spine in balancing the net external moment was initially at the onset of vibration at 48% and 87% that dropped at the peak positive acceleration to minimum values of 13% and 22% in erect and flexed postures, respectively (Fig. 5.5). Local compression and shear forces reached their maximum values at the L5-S1 level and were larger in flexed posture with maximum values of 1608 N and 771 N, respectively, as compared to 1131 N and 508 N for the erect posture. The associated computed forces in global extensor (LG and IC) and flexor (RA, IO and EO) muscles showed no coactivity and followed the temporal trends predicted for net moments and spinal loads (Fig. 5.6). Due to larger net moments, muscle forces were greater in the flexed posture than in the erect posture.

The critical (minimum) muscle stiffness coefficient to maintain trunk stability was substantially influenced by muscle activity and posture (Fig. 5.7). The trunk was much more stable (i.e., smaller critical q values) during periods with activity in abdominal muscles and on the contrary less so at the periods with peak activity in extensor muscles (Fig. 5.7). The flexed posture resulted in an overall improvement in trunk stability due primarily to greater resistance provided by passive tissues.

Introduction of 2%, 1%, and 0.5% as a priori antagonistic coactivity in the abdominal IO, EO, and RA muscles, respectively, substantially improved the trunk stability in the erect posture (i.e., requiring much smaller critical q values) during periods when no abdominal activity was otherwise present (Fig. 5.8). The foregoing prescribed antagonistic abdominal coactivities diminished the critical muscle stiffness coefficient from 89 to 21 at the onset of vibration and from 131 to 39 at the peak upward acceleration. The spinal loads, on the contrary, increased in the same periods by as much as 196 N in compression force and 138 N in shear force.

Softening of the buttocks did not affect the response as much. On the contrary, stiffening of buttocks increased the magnitude of accelerations; for example the peak accelerations at the head reached ±17 ms^{-2} for the case with rigid buttocks. Furthermore first natural frequency of the system under gravity load and prescribed postures was ~5.5 Hz that altered to ~12.4 Hz or 3.9 Hz as buttocks became completely rigid or softer, respectively.

5.5 Discussion

The iterative Kinematics-driven finite element approach was used to solve the redundant passive-active trunk system at the seated position subject to an input random whole body vibration. The time variations of trunk muscle forces, spinal loads and trunk stability were evaluated. This is a novel investigation performed in response to the recognized need for development of more anatomically detailed biomechanical models of trunk in whole body vibration biodynamic (Seidel, 2005; Seidel and Griffin, 2001).

There was a good agreement between predicted and measured (Seidel et al., 1997) accelerations at the L3 and head levels (Fig. 5.4). Noticeable differences in the force magnitudes at the seat-subject interface at the times of peak acceleration (Fig. 5.4) could partly be due to the dynamic contributions of thighs and legs that were absent in the model but present in measurements. The predicted dynamic component of base reaction force reached the maximum of 400 N at peak positive acceleration (i.e. 10 ms^{-2}) whereas the minimum of -460 N (i.e. equal to the trunk weight) at the peak negative acceleration (i.e. -12.2 ms^{-2}).

A single L2-L3 lumbar motion segment has been reported to yield an axial natural frequency of ~32 Hz (Kasra et al., 1992). This natural frequency drops to ~18 Hz when considering a finite element model of two lumbar motion segments, L4-S1 (Goel et al., 1994) and furthermore to ~11 Hz as the entire lumbar spine is considered (Kong and Goel, 2003). The incorporation of buttocks in the current model with proper stiffness and damping values (Aimedieu et al., 2003; Kitazaki and Griffin, 1997a) diminished the first vertical natural frequency of the seated trunk under gravity from ~12 Hz to ~5.5 Hz, in agreement with earlier measurement studies (Kong and Goel, 2003; Pope et al., 1990). With no constraint on sagittal displacements, vibration analysis at the loaded configuration yielded the lowest frequency of ~1 Hz that is also in agreement with the reported values in the literature (Kitazaki and Griffin, 1997b).

The excitation at the base substantially influenced, via the inertial and muscle forces, the spinal compression and shear forces. The maximum spinal compression and shear forces predicted at the L5-S1 increased significantly from the initial (static) values of, respectively, 535 and 222 N in the erect posture and 717 and 297 N in the flexed posture to peak (dynamic) values of 1131 and 508 N in the erect posture and 1608 and 771 N in the flexed posture. Back pain prevention and rehabilitation programs are designed based on the premise that excessive loads in the ligamentous spine could cause injury. The compression strength of lumbar motion segments has been reported to be in the range of 2-10 kN (Brinckmann et al., 1989; Jager and Luttmann, 1997; Ortoft et al., 1993). Jager and Luttmann (1991) reported values of 5.81±2.58 kN for males and 3.97±1.5 kN for females based on relatively large sample populations. The strength in shear force has been reported to be >1 kN (Cyron et al., 1976; Miller et al., 1986). Notwithstanding the effect of strain rate, earlier injuries/degeneration, fatigue and combined loading on these strength values (Seidel et al., 1998), lower risk of injury could be associated with conditions that yield smaller loads on spine. Hence, considering the mild shock contents of ~ ±1 g in vibration input data of current study as compared with reported larger shock values of ~ ±2-6 g in off-road and industrial vehicles

(Robinson, 1999), it is likely that spinal loads in latter vibration environments approach and even exceed strength limits causing injury.

The biomechanical advantages of preservation or flattening (i.e. flexing) of the lumbar lordosis in various activities such as sitting or lifting remain controversial. In a recent study (Arjmand and Shirazi-Adl, 2005), the free style posture or a posture with moderate lumbar flexion was advocated as the posture of choice in static lifting tasks. The 10° flattening in the lumbar lordosis from the erect to flexed posture in the current sitting position under whole body vibration increased the net moment, muscle forces and passive spinal loads and hence the risk of tissue injury while on the other hand substantially improved the trunk stability. The latter effect was due to the increased stiffness of muscles and passive ligamentous spine in the flexed posture. The moment at the S1 as well as compression and shear forces at the L5-S1 increased in the flexed posture by, respectively, 6.8 Nm, 182 N, and 75 N at the onset of vibration and by 19 Nm, 482 N, and 240 N at the time of peak positive acceleration response. Similarly, the introduction of antagonistic coactivity in abdominal muscles at low to moderate levels increased the passive spinal loads while improving the spinal stability. It appears hence that an increase in trunk stability can be achieved only at the cost of higher passive loads and hence greater risk of tissue injury suggesting a tradeoff in muscle activities.

Activation in muscles has opposite effects on the spinal stability; on one hand, it increases compression force on spine (i.e. destabilizing role); but on the other, it offers greater stiffness associated with larger activation (i.e. stabilizing role). Furthermore, due to larger lever arms (Arjmand et al., 2007), abdominal muscles even with much smaller forces are more efficient than extensor muscles in stabilizing the trunk. Results demonstrate much lower critical q values in presence of abdominal activities, being agonistic or antagonistic. The positive role of antagonistic activities in enhancement of the spine stability has been recognized in the literature (El-Rich et al., 2004; Gardner-Morse et al., 1995; Granata and Orishimo, 2001). It should be emphasized that the choice of the widely used linear force-stiffness relationship (i.e. intrinsic muscle stiffness) (Arjmand and Shirazi-Adl, 2005; Bergmark, 1989) assumed in this study rather than a nonlinear one (i.e. reflexive stiffness) (Shadmehr and Arbib, 1992; Zeinali Davarani et al., 2007) have absolutely no influence on computed muscle forces and, hence, internal spinal loads. Nevertheless, the choice of force-stiffness relationship would influence the system stability. Moreover, consideration of delays in muscle spindle reflex response (Zeinali Davarani et al., 2007) in vibration would have no bearing on muscle forces computed in this study as these forces are inclusive of passive, reflexive and areflexive contributions.

Due to changes in contact area between buttocks/thighs and the seat as the lumbar posture alters, the stiffness of the element simulating buttocks may need to be modified (Kitazaki and Griffin, 1997b). Although no such changes were considered in the current study, the first natural frequency of the model was found to be highly dependent on the buttocks stiffness demonstrating the likely indirect effect of posture on the natural frequency. Rigid buttocks increased the system resonant frequency from 5.5 Hz to 12.4 Hz, increased the acceleration response especially at the time of peak acceleration, and resulted in larger net moments. Reverse trends were found as the buttock stiffness decreased.

In conclusion, a detailed trunk muscle architecture along with nonlinear properties of the ligamentous spine, wrapping of global extensor muscles and trunk dynamic characteristics (inertia and damping) was used in our Kinematics-driven model to evaluate muscle forces, spinal loads and trunk stability under a random base whole body vibration reported in the literature (Seidel et al., 1997). The predictions satisfied kinematics and dynamic equilibrium conditions at all levels and directions. Large net moments, muscle forces and spinal loads along with a deteriorated stability margin were predicted at the vibration periods with peak accelerations. The flexed posture, compared to the erect one, increased muscle forces and spinal loads but improved trunk stability.

Agonistic or antagonistic activities in abdominal muscles substantially improved spinal stability. Results points to a likely tradeoff between the opposing effects of higher muscle forces in improving trunk stability on one hand but increasing spinal loads and risk of injury on the other. Additional muscle coactivity, as a compensatory response to an injury in the passive spine or to insufficient stability for example, would further increase the spinal loads and the risk of fatigue and failure.

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5.7 References

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Table 5.1 Trunk segments mass, mass moment of inertia and mass center locations

Level	% TM*	% BM	Ixx	Iyy	Izz	CG - z	CG - x
Head-Neck	--	6.94	27.18	29.34	20.13	597.60	-10.00
Upper Arms	--	2*2.8	12.63	11.30	3.80	447.38	30.00
Lower Arms	--	2*1.6	6.45	5.99	1.20	426.85	30.00
Hands	--	2*0.6	1.31	0.88	0.50	405.81	30.00
T1	3.59	1.28	6.70	2.00	8.70	467.60	-8.00
T2	3.88	1.38	3.40	2.40	9.10	447.38	-12.00
T3	4.15	1.47	8.40	3.20	11.50	426.85	-20.00
T4	4.46	1.58	8.30	3.40	11.70	405.81	-28.00
T5	4.72	1.68	8.00	3.50	11.50	384.14	-33.00
T6	5.03	1.78	7.80	3.90	11.60	361.70	-39.00
T7	5.29	1.88	7.40	4.10	11.50	338.40	-43.00
T8	5.60	1.99	7.20	4.40	11.60	314.12	-45.00
T9	5.91	2.10	7.20	4.70	11.80	288.94	-48.00
T10	6.17	2.19	8.90	6.20	15.00	262.94	-48.00
T11	6.47	2.30	9.00	6.20	15.20	235.30	-46.00
T12	6.74	2.39	11.00	7.20	18.10	204.56	-44.00
L1	7.04	2.50	11.10	6.50	17.50	171.07	-37.01
L2	7.30	2.59	10.90	6.00	16.80	135.03	-29.00
L3	7.61	2.70	10.70	5.50	16.10	97.55	-17.00
L4	7.87	2.79	11.20	5.30	16.40	58.90	-10.00
L5	8.19	2.91	12.20	5.60	17.70	20.57	-6.00
S1	0.00	0.00	0.00	0.00	0.00	0.00	0.00
Pelvis	--	11.00	75.00	30.00	80.00	-89.00	0.00

* TM: Trunk mass, BM: Body mass, *Ixx*, *Iyy*, *Izz*: Mass moments of inertia respectively in anterior-posterior, transverse and longitudinal directions ($Kg.m^2 * 10^{-3}$), CG-z: height of the centers of mass with respect to the S1 (*mm*), CG-x: anterior-posterior distance from corresponding vertebral centers with negative indicating anterior position (*mm*). Upper arms, lower arms, and hands centers of mass are considered posteriorly at T2, T3, and T4 vertebral levels, respectively.

Table 5.2 Buttocks' nonlinear stiffness properties

Force (N)	Deflection (m)	Stiffness (kN/m)
0.0	1.0	0.0
0.0	0.0	0.0
-12.5	-0.001	12.5
-62.5	-0.002	50
-1232.5	-0.02	65
-7232.5	-0.05	200

Table 5.3 Prescribed total sagittal rotations (degree) at different levels in two modeled seated postures (negative: flexion)

Level	Erect	Flexed
Head-T12	-9.9	-15.9
L1	-6.2	-11.5
L2	-2.5	-6.5
L3	0.3	-2.1
L4	1.8	1.8
L5	3.5	6.0
S1	5.0	9.0

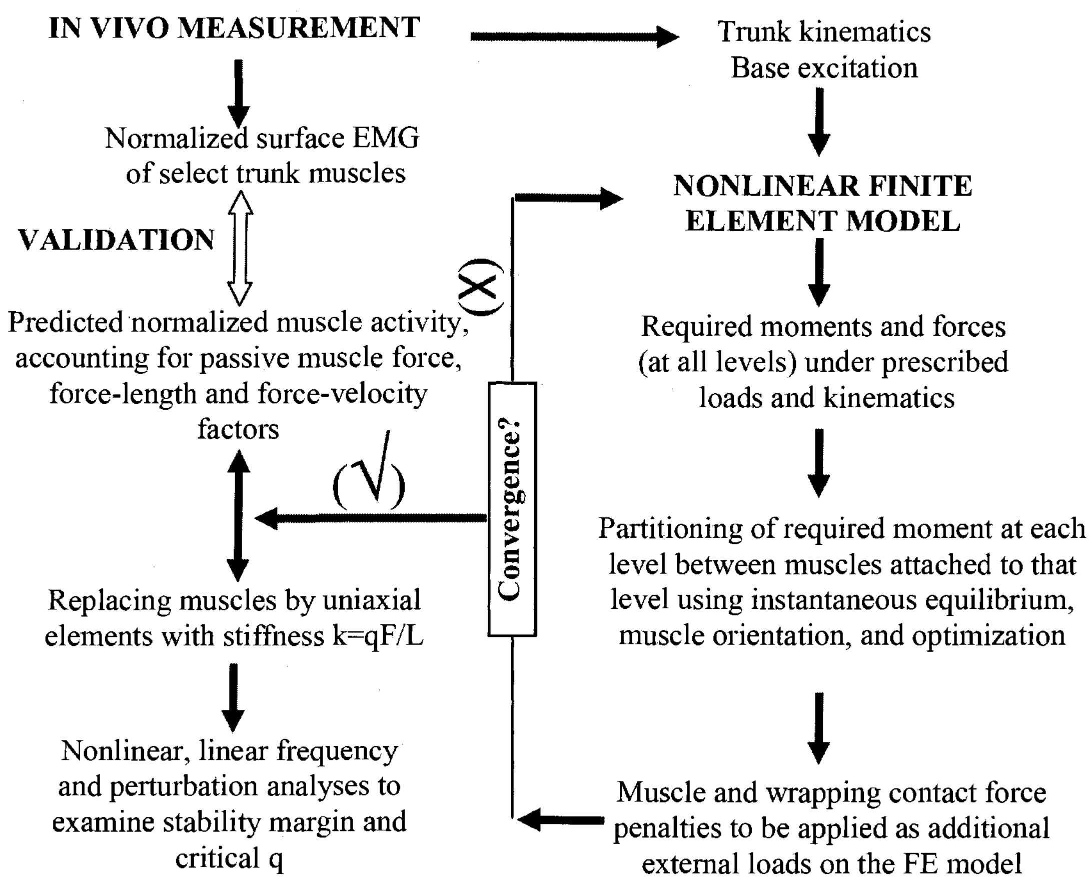

Figure 5.1 Flowchart for the application of the Kinematics-based approach used to determine trunk muscle forces, internal loads and spine stability under whole body vibration base excitation. Convergence is attained if calculated muscle forces in two successive iterations remain the same.

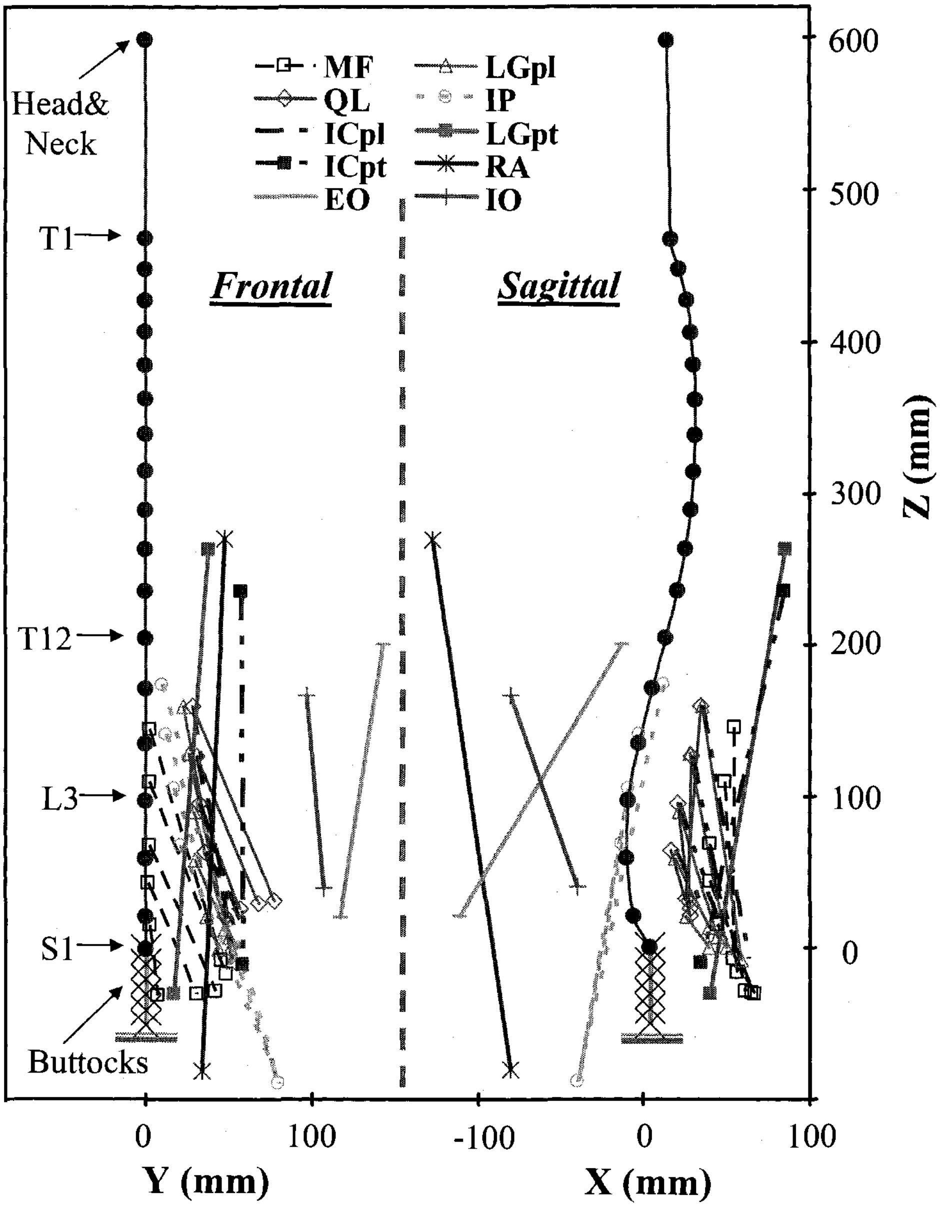

Figure 5.2 Representation of the spine including global and local musculatures in the sagittal (on the right) and frontal (on the left, fascicles on one side are shown) planes. ICpl: Iliocostalis lumborum pars lumborum, ICpt: Iliocostalis lumborum pars thoracic, IP: Iliopsoas, LGpl: Longissimus thoracis pars lumborum, LGpt: Longissimus thoracis pars thoracic, MF: Multifidus, QL: Quadratus lumborum, IO: Internal oblique, EO: External oblique, and RA: Rectus abdominus

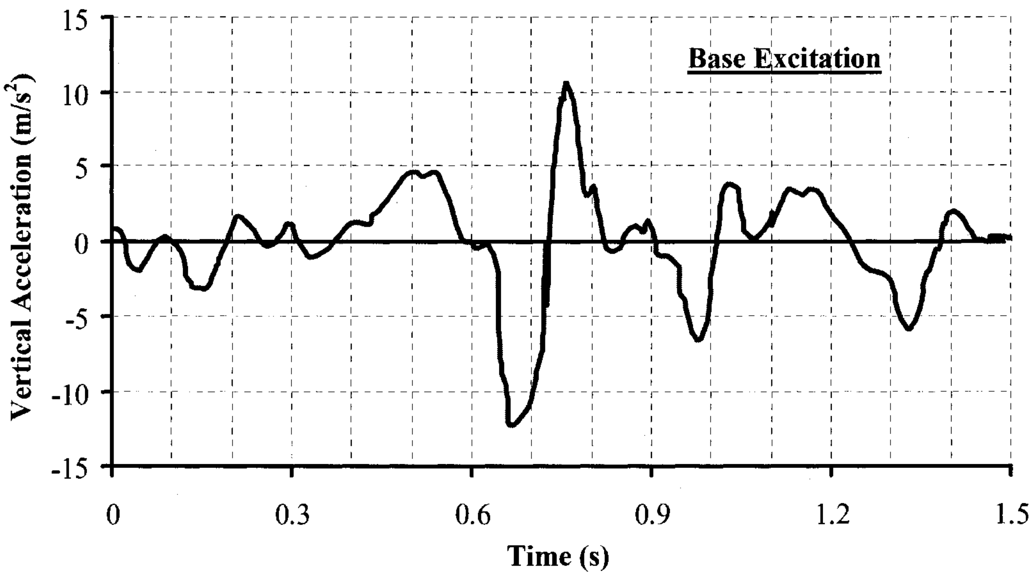

Figure 5.3 The input random seat vertical acceleration at the base considered in the model based on the measured un-weighted acceleration at the seat-driver interface of a hydraulic excavator (Seidel et al., 1997).

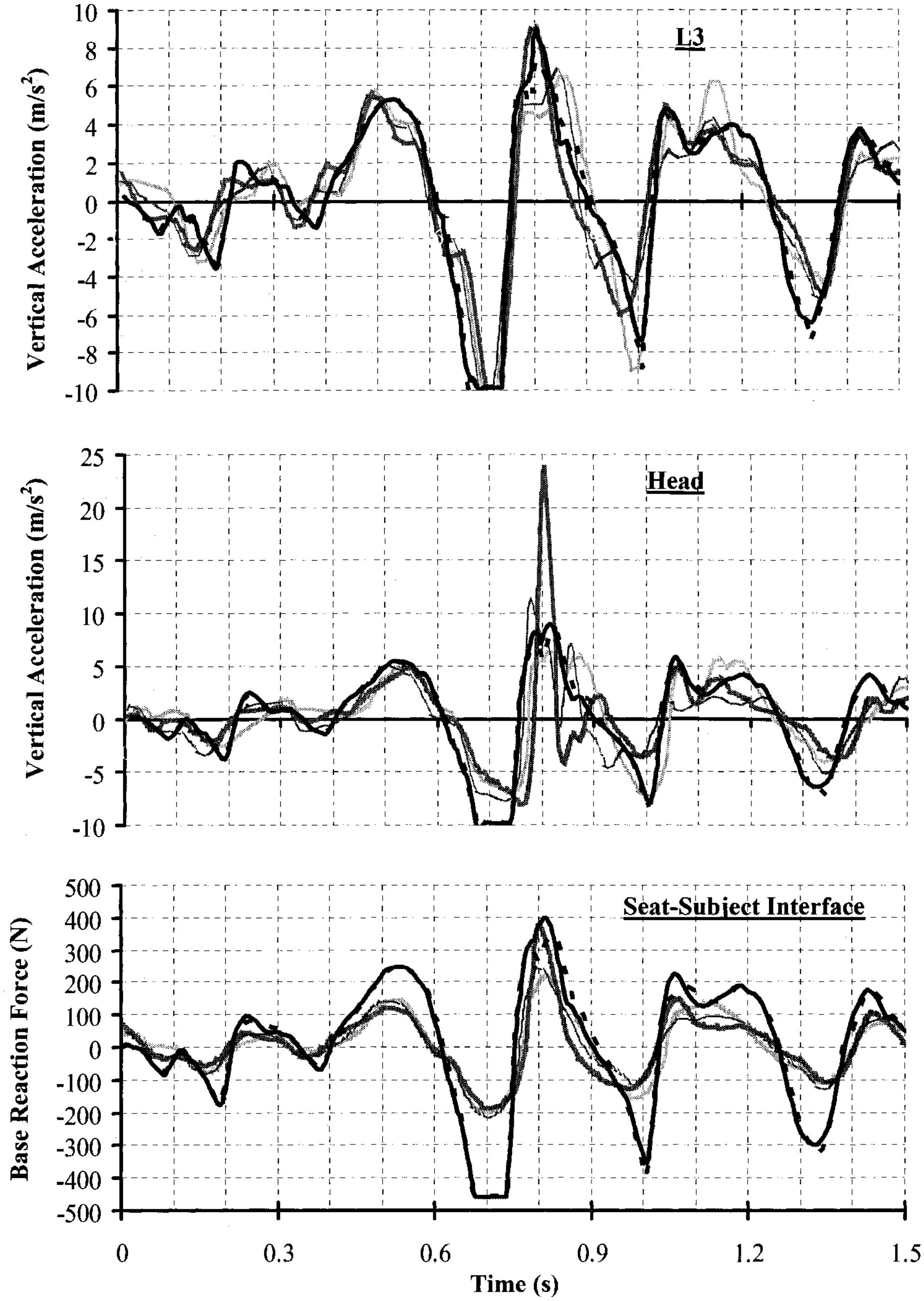

Figure 5.4 Measured and predicted response at L3 (top) and Head (middle) along with reaction force at seat-subject interface (bottom). Three postures considered in experimental studies (i.e. Driving, Bent, and Upright) and two in the current model studies (i.e. Erect, Flexed) are presented.

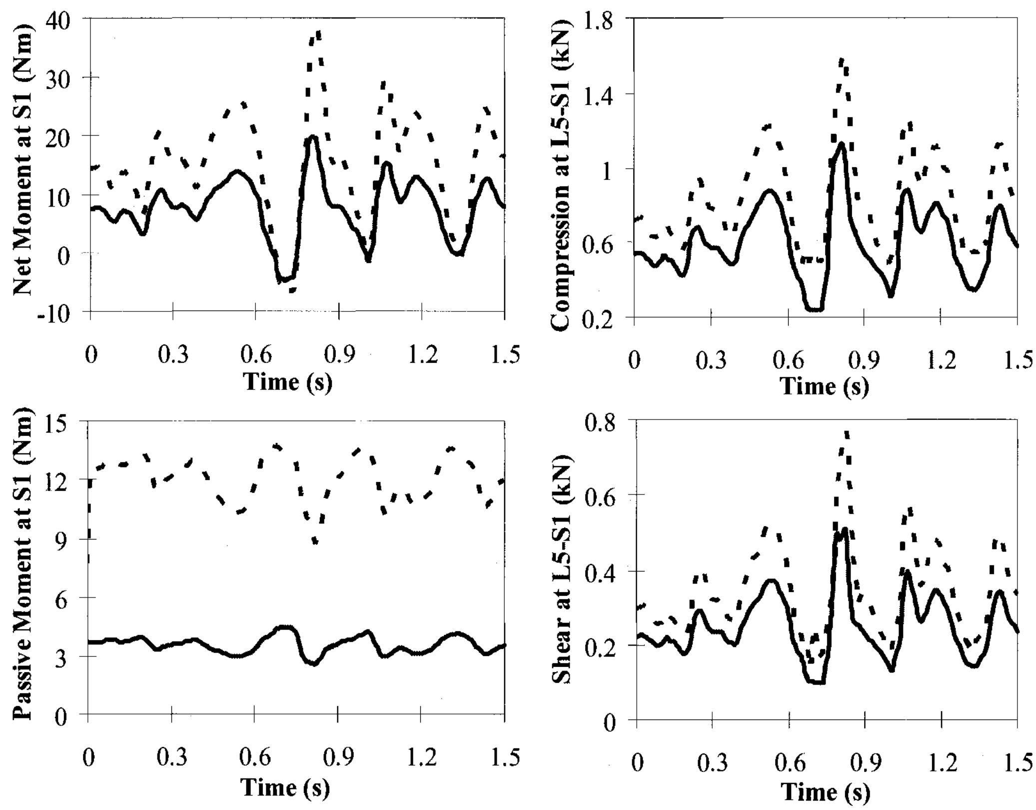

Figure 5.5 Predicted temporal variations of internal spinal forces at the L5-S1 disc mid-height in local directions (right-top: compression force, right-bottom: shear force), the net moment at the S1 (to be resisted by all muscles and passive ligamentous spine, left-top) and the passive ligamentous moment at the S1 (left-bottom) for erect and flexed postures. ▬▬ Erect, ▪▪▪▪▪ Flexed

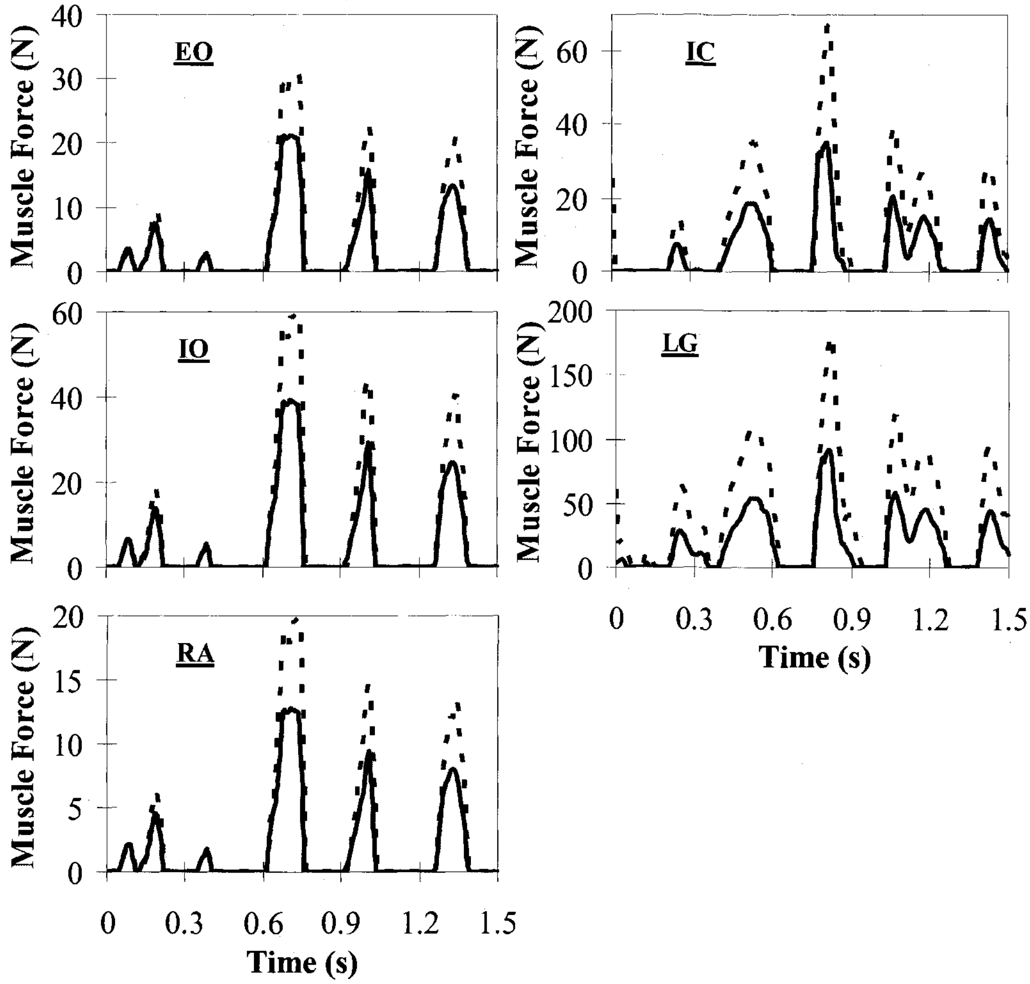

Figure 5.6 Predicted temporal variations of active forces in global muscles (one side only), IO: External oblique, EO: Internal oblique, RA: Rectus abdominus, IC: Iliocostalis, and LG: Longissimus. ▬▬ Erect, ■■■■■ Flexed

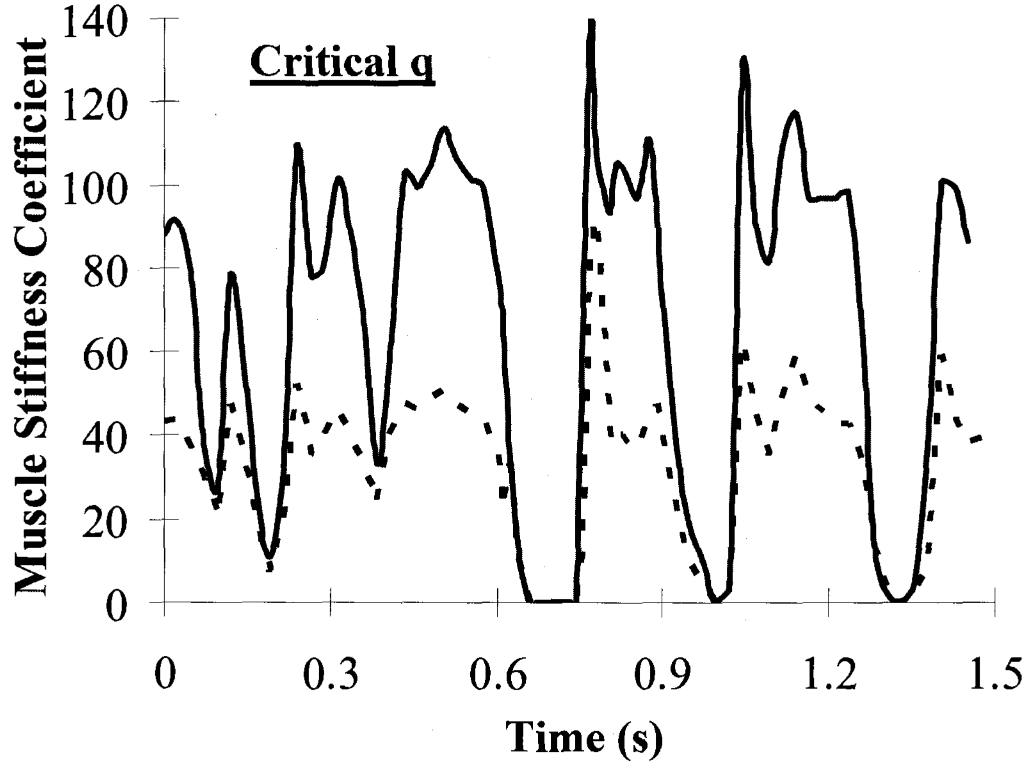

Figure 5.7 Predicted temporal variations of minimum (critical) muscle stiffness coefficient, q, for two different lumbar postures. A lower q value indicates higher trunk stability. ▬▬ Erect, ▪▪▪▪▪ Flexed

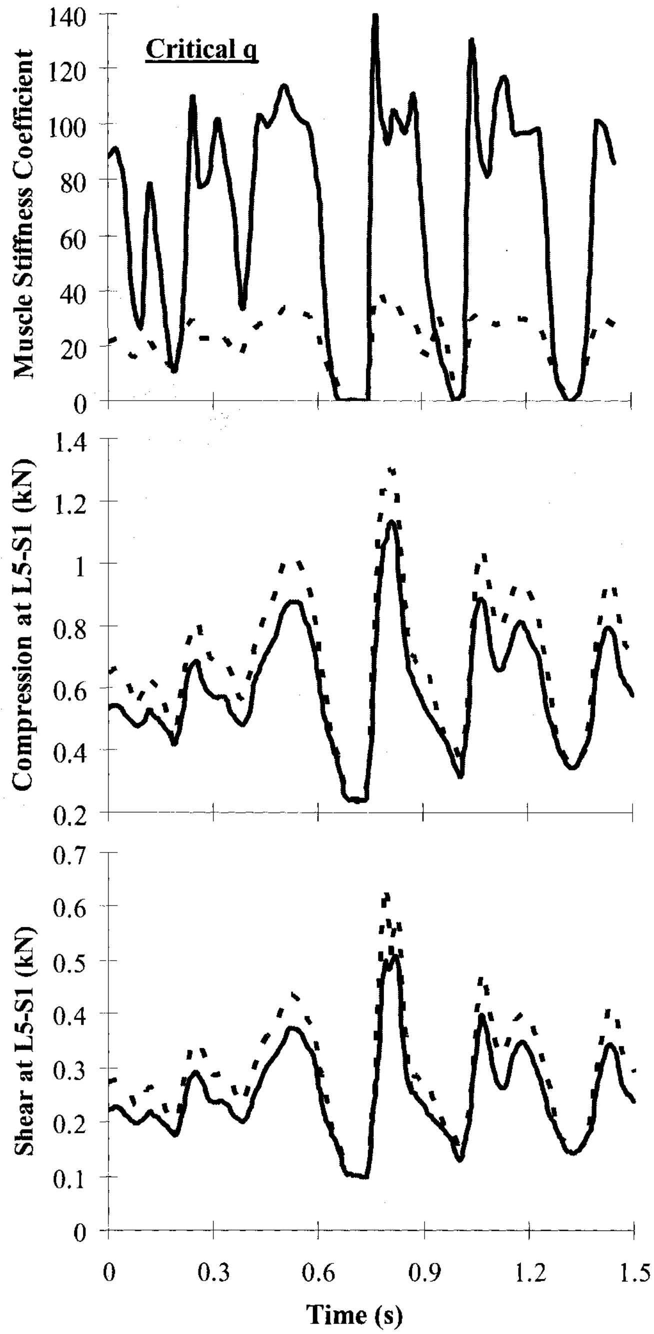

Figure 5.8 Predicted temporal variations of minimum (critical) muscle stiffness coefficient, q (top) and compression/shear forces at the lowermost L5-S1 levels in the erect posture for cases with and without prescribed coactivity in abdominal muscles (see the text for coactivity levels). ▬▬ No Coactivity, ▪▪▪▪▪ With Coactivity

CHAPTER SIX

DISCUSSION AND CONCLUSIONS

6-1- Overview

Back pain is a significant socioeconomic burden on our aging society and has considerable impact on people's quality of life. Work environments expose workers to conditions that have been demonstrated to increase risk of back disorders. Manual material handling, fast trunk motion, repetitive task, awkward postures, whole body vibration and shock are reported to be the major mechanical risk factors in the work place (NIOSH, 1997). Personal as well as psychosocial parameters can also affect biomechanics of the human spine (Marras, 2005), the risk involved and the subsequent treatment outcome. Effective management of back disorders, however, depends partly on accurate estimation of trunk mechanical behaviour (i.e. spinal loads, muscle forces, stress/strain in passive and active trunk components, and spinal stability) under different working activities. In the absence of any measurement device, such crucial data can only be obtained using biomechanical models.

Rigid link segment models, essentially used to evaluate manual material handling tasks, provide the net external moment at a spinal joint accounting for gravity, external and inertial loads. While this output can by itself be assessed as an indirect index of spinal loading, a complementary detailed model is needed to estimate muscle forces and passive spinal loads (i.e. compression and shear forces). The major shortcoming of such commonly used models lies behind the calculation of muscle forces and spinal loads based on consideration of equilibrium at only one spinal level (Arjmand et al., 2007b). Other models (i.e. lumped parameter and finite element models) employed to evaluate trunk response to whole body vibration have generally overlooked the muscle forces and its crucial role in spinal loads and trunk stability. In brief, existing biodynamic models of the spine have either neglected or oversimplified the nonlinear passive resistance of the spine motion segments, complex geometry/loading/dynamics of the spine, maintenance of equilibrium in all spinal levels and directions, and the wrapping of extensor muscles. More accurate estimation of spinal loads, hence, calls for improved biodynamic models of the human spine.

In the current study, in continuation of our earlier isometric model studies (e.g., (Arjmand and Shirazi-Adl, 2006; El-Rich et al., 2004; Kiefer et al., 1997; Shirazi-Adl et al., 2002)), an iterative dynamic kinematics-based approach was developed to alleviate limitations of earlier biomechanical models. Measured kinematics data were input into a nonlinear finite element model and differential equations of motion were numerically solved to calculate required joint moments and forces, subject to gravitational and inertial loading. In this manner, while satisfying equilibrium at all spinal levels and direction, calculated muscle forces, spinal loads, and system stability were in full accordance with prescribed kinematics and nonlinear passive properties. The method was employed to evaluate the effect on trunk biodynamic of some loading conditions and mechanical parameters such as different lifting techniques (i.e. stoop and squat), velocity of trunk movement (slow, medium and fast velocities), whole body vibration and shock, wrapping of global extensor muscles, antagonistic trunk muscles activities, and alterations in passive properties of the spine and buttocks.

6-2- Critical Evaluation of the Model

Dynamic kinematics-based approach combines experimental and model studies to evaluate biomechanics of the spine during different activities. Trunk kinematics as input for model studies along with ground reaction force and EMG of select trunk muscles for the sake of comparison with model predictions are measured. A nonlinear finite element model of the spine along with detailed muscle architecture is used to calculate trunk muscle forces, spinal loads and to assess stability of the trunk. Predictions expectedly depend on underlying assumptions made in the model as well as accuracy of measured input kinematics data. Therefore, subsequent validation of model by comparison of its predictions with available in-vitro and in-vivo measurements is important.

6-2-1- Finite element model

The current model has evolved from those developed by Kiefer et al (1998), El-Rich et al (2004), and Arjmand and Shirazi-Adl (2005a). The geometry of ligamentous spine model has been constructed using CT scan images of a cadaver lumbar specimen and data in the literature (Shirazi-Adl and Parnianpour, 1993, 1996). The geometry of our model stands in agreement with reported trunk geometries in earlier studies (Fig. 6.1). Pearsall et al (1996), for instance, reconstructed the spine geometry of four subjects (close to 85^{th} percentile for stature and body mass index for their age and gender groups) from CT scan images obtained in supine position (Fig. 6.1). In another study by Stokes and Gardner-Morse (1999), spine geometry in standing posture was obtained using stereo-radiographs of four healthy young subjects. Kitazaki and Griffin (1997a) have also presented the spinal curve in their model of whole body vibration by adapting data of Liu and Wickstrom (1973), Belytschko et al (1976), and Singley III and Haley (1978). The geometry of the model was not changed from a subject to another as the T5-S1 height and total mass of subjects chosen for lifting and flexion-extension simulations as well as the corresponding average values for all subjects matched relatively well the values taken in the model (Chapters 2 to 5).

The finite element model, in the present study, was a sagittally symmetric head-pelvis model made of six nonlinear deformable beams to represent T12-S1 segments and seven rigid elements to represent head-T12 (as a single body) and lumbosacral (L1-S1) vertebrae. The beams represented the overall nonlinear stiffness of T12-S1 motion segments (i.e., vertebrae, disc, facets and ligaments) at different levels with nonlinear axial compression-strain and sagittal/lateral/axial moment-curvature relations.

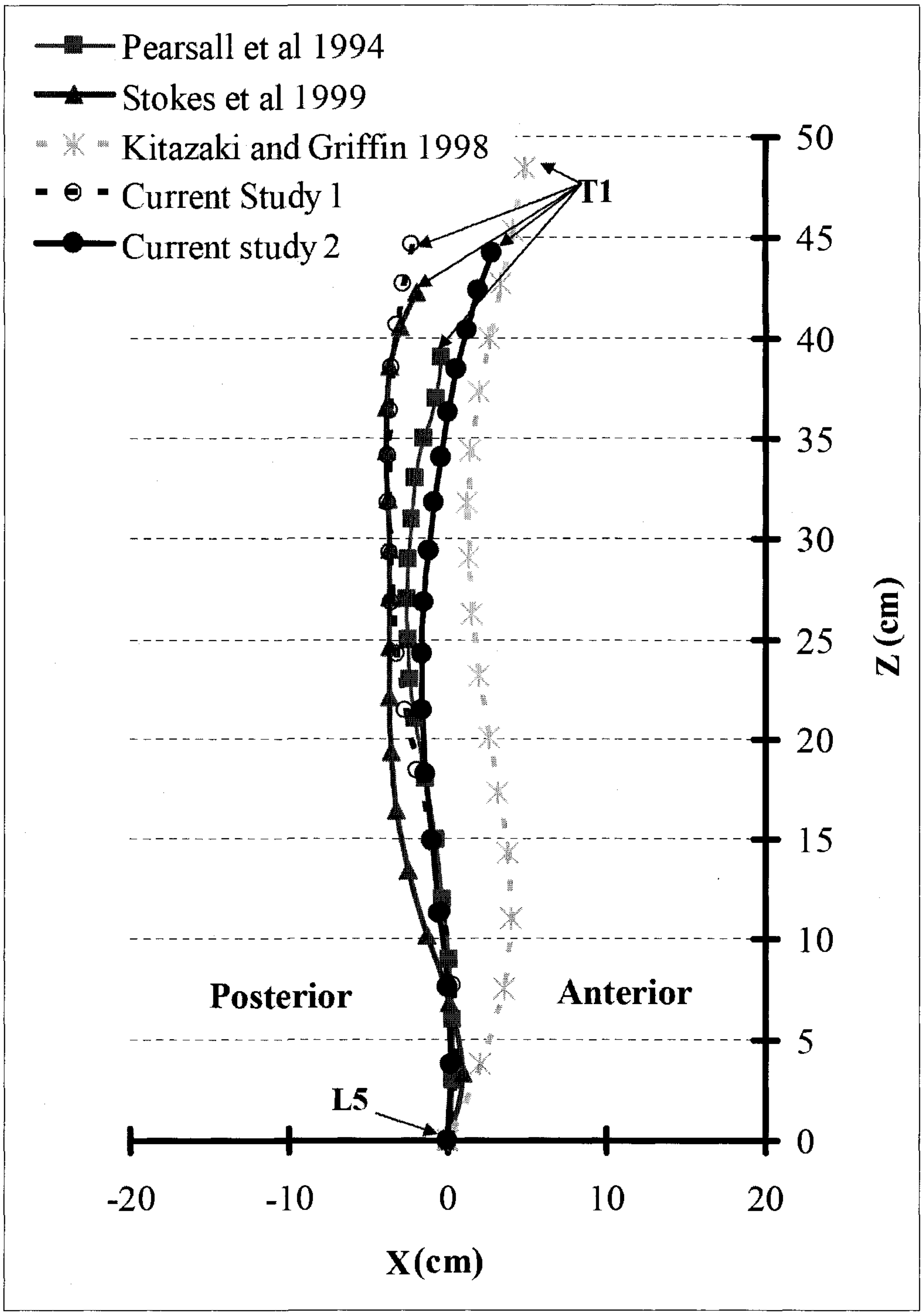

Figure 6.1 Comparison of geometries of the current model (i.e. standing posture without (Current study 1) and with (Current study 2) gravity load) with those reported in the literature (Kitazaki and Griffin, 1997a; Pearsall et al., 1996; Stokes and Gardner-Morse, 1999).

The nonlinear load-displacement response in different directions along with flexion versus extension differences were represented based on numerical and measured results of previous single- and multi-motion segment studies (Oxland et al., 1992; Pop, 2001; Shirazi-Adl et al., 2002; Yamamoto et al., 1989). Axial compression loads significantly increase stiffness of lumbar motion segments in different planes (Broberg, 1983; Janevic et al., 1991; Kasra et al., 1992; Shirazi-Adl et al., 1986; Stokes and Gardner-Morse, 2003); however, experimental and numerical investigations on the spinal multi-motion segments in presence of physiological compression loads cannot adequately be carried out due to structural instability and artefact loads. To circumvent this problem, different methods have been developed, both experimentally and numerically, which allowed application of large compression on such an unstable structure. For instance, Patwardhen et al (2003) used a follower load path which enabled them to measure the lumbar stiffness under compression loads of up to 1200 N with no sign of instability. In another effort, Shirazi-Adl (2006) introduced a wrapping cable element in his detailed nonlinear finite element model of the lumbosacral spine (L1-S1) which allowed for the application of compression loads of up to 2700 N with no instability or artefact loads. Accordingly, these predicted increases in load-displacement responses of lumbar motion segments were considered in the current study (Fig. 6.2).

Trunk/head/arms/pelvis masses and mass moments of inertias were assigned at different levels along the spine at their respective gravity centers based on published data (de Leva, 1996; Pearsall et al., 1996; Zatsiorsky and Seluyanov, 1983). At each spinal level, a mass was assigned as a percentage of total trunk mass based on measurements of Pearsall et al (1996) (Fig. 6.3 and 6.4). They used computed tomography (CT) imaging technique to determine in-vivo mass, center of mass, and moment of inertia of discrete segments of the trunk. The whole trunk mass was estimated to comprise 41.6% of the total body mass with body mass percentages at vertebral levels ranging from 1.1% at the T1 to 2.6% at the L5 (Fig 6.3). Mass centers at different vertebral levels were found to lie anterior to their respective vertebral center by up to 5 cm. Similar results were also estimated by Liu and Wickstrom (1972) using direct measurements on cadaveric segments. Mass and mass moment of inertia of the head, arms, hands, and buttocks in the current study were estimated using regression equations given by Zatsiorsky and Seluyanov (1983).

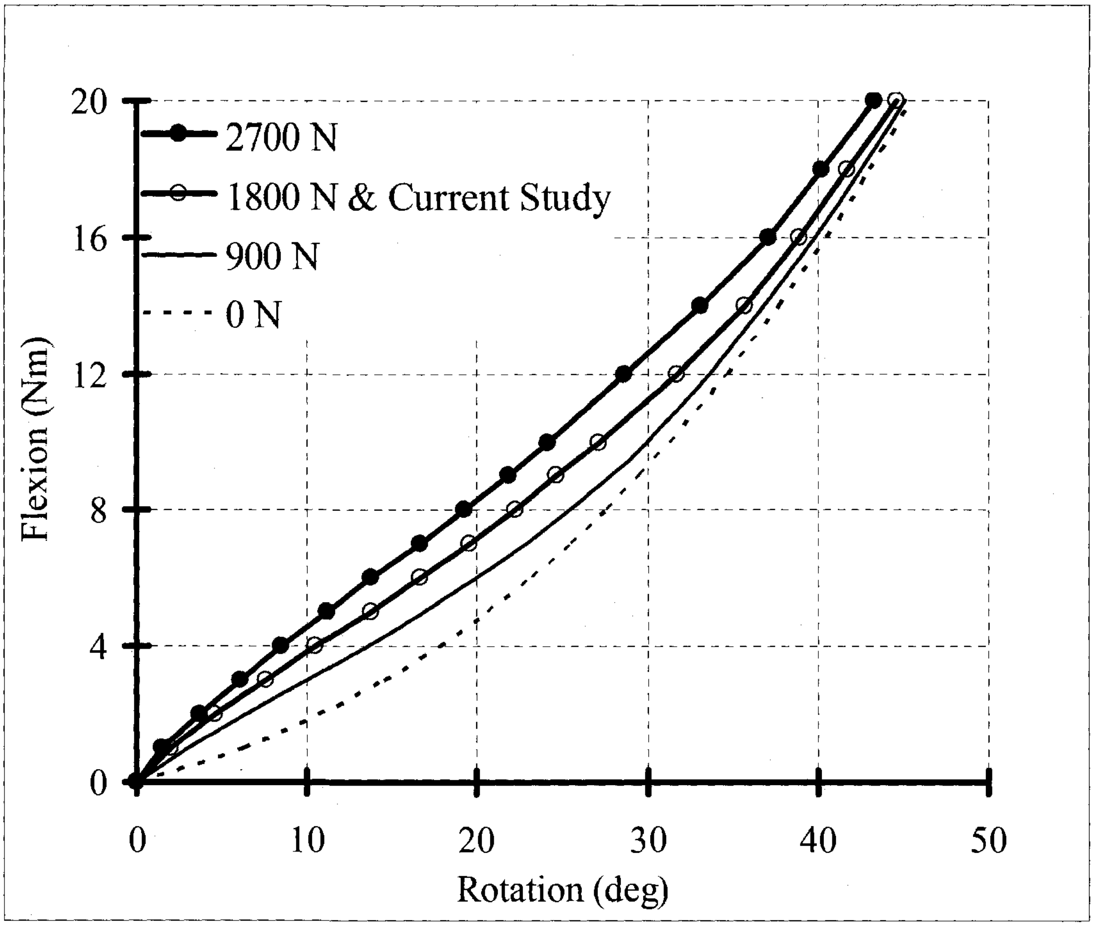

Figure 6.2 Nonlinear lumbar responses to applied flexion moment under 0, 900, 1800, and 2700 N wrapping compression preloads

Mass of the head was located 1 cm anterior and 13 cm superior to the center of the T1 vertebra according to works of Takashima et al. (1979) and de leva (1996), while those of arms were distributed and applied at 3 cm posterior to the center of T2, T3, and T4 vertebrae. To simulate inertial and gravity loads of external load (for the case of lifting

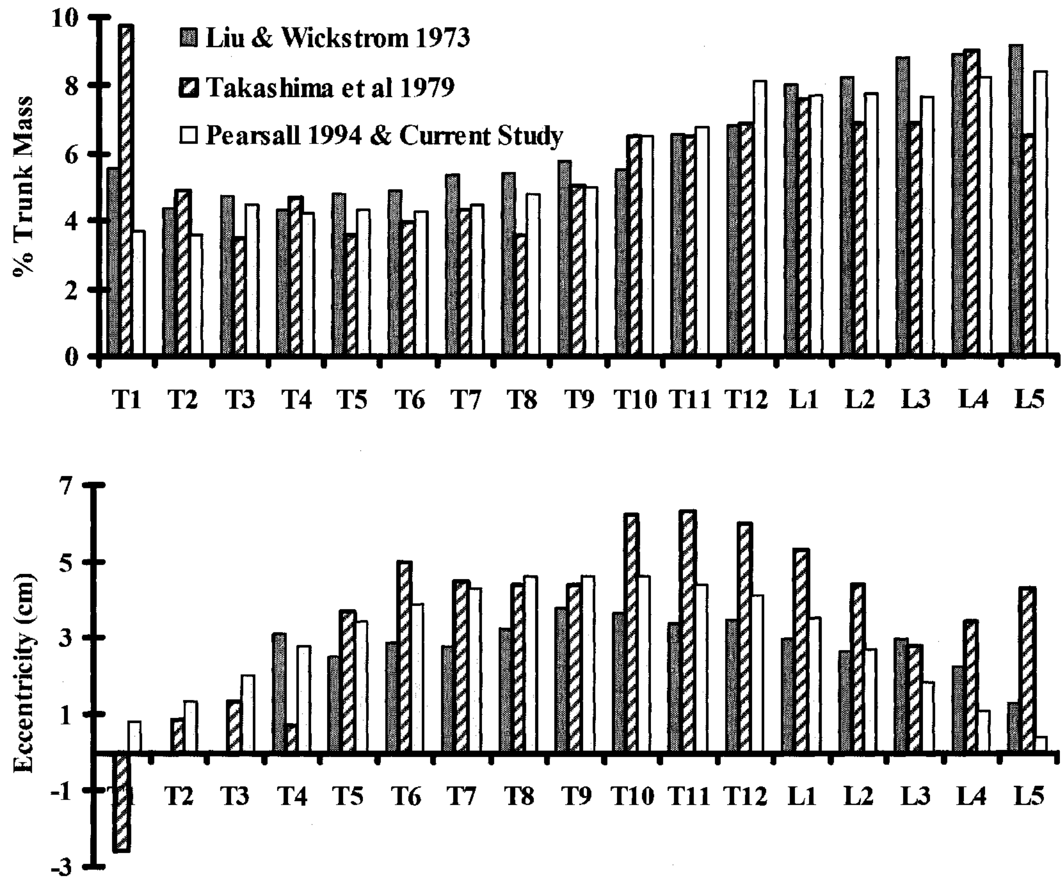

Figure 6.3 Distribution of trunk mass and its eccentricity to vertebral center along the spine at each vertebral level. Data of Pearsall (1994) was considered in current study.

in Chapters 2 and 3), a concentrated mass (~18.5 Kg) was connected at the T3 using two rigid elements and two actuators (i.e. one connecting two rigid elements together and the other connecting the assembled set to the spine). Time variations of actuators' rotations were calculated based on measured kinematics data of the load and the thorax.

To simulate the inter-segmental damping, connector elements parallel to deformable beams were introduced to the model. Kasra et al (1992) measured an axial damping ratio of about 0.08 (~1239 Ns/m) for individual lumbar motion segments under varying compression preloads using resonant amplification, half-power, and energy loss per cycle methods. Values of 1800 Ns/m by Markolf (1970) and 2567 Ns/m by Izambert et al (2003) for the axial damping of lumbar motion segment have also been reported. Markolf (1970) reported a rotational damping value of 1.2 Nms/rad based on in-vitro measurements of cadaveric specimens. In a more recent study, Crisco et al (2007) measured rotational damping of lumbar motion segments using a pendulum system and reported damping coefficients of ~1.4 to 4 Nms/rad for preloads of 78 to 488 N. In the current study measurement data of Kasra et al (1992) and Markolf (1970) were increased by fivefold to account for damping of both lumbar motion segments and surrounding soft tissue (see Chapter 5 and Appendix A). Buttocks at the base was modeled by a connector element with a mass of about 11% of body mass located 8.9 cm below the lower endplate of the L5-S1 disc. Kitazaki and Griffin (1997a) adapted a linear stiffness value of ~65.5 kN/m based on Payne and Band (1971) for buttocks of their whole body vibration model. In a recent study, using a measurement set up along with a two-parameter viscoelastic model, Aimedieu et al (2003) measured a monotonous increase in stiffness of a soft tissue (i.e. porcine muscle) from 8.5 kN/m at 5 Hz to 347 kN/m at 30 Hz. Accordingly, we considered a nonlinear stiffness (see Appendix A) for the buttocks in our model based on reported data in those studies (Aimedieu et al., 2003; Kitazaki and Griffin, 1997a). The buttocks damping was, however, considered similar to that of lumbar segments.

In presence of nonlinearity in equations, numerical integration using an unconditionally stable implicit method was employed in this study. For this purpose, the software (i.e. ABAQUS) was set to use Hilber-Hughes-Tylor (Hilber et al., 1978) integration operator to solve the nonlinear transient problem.

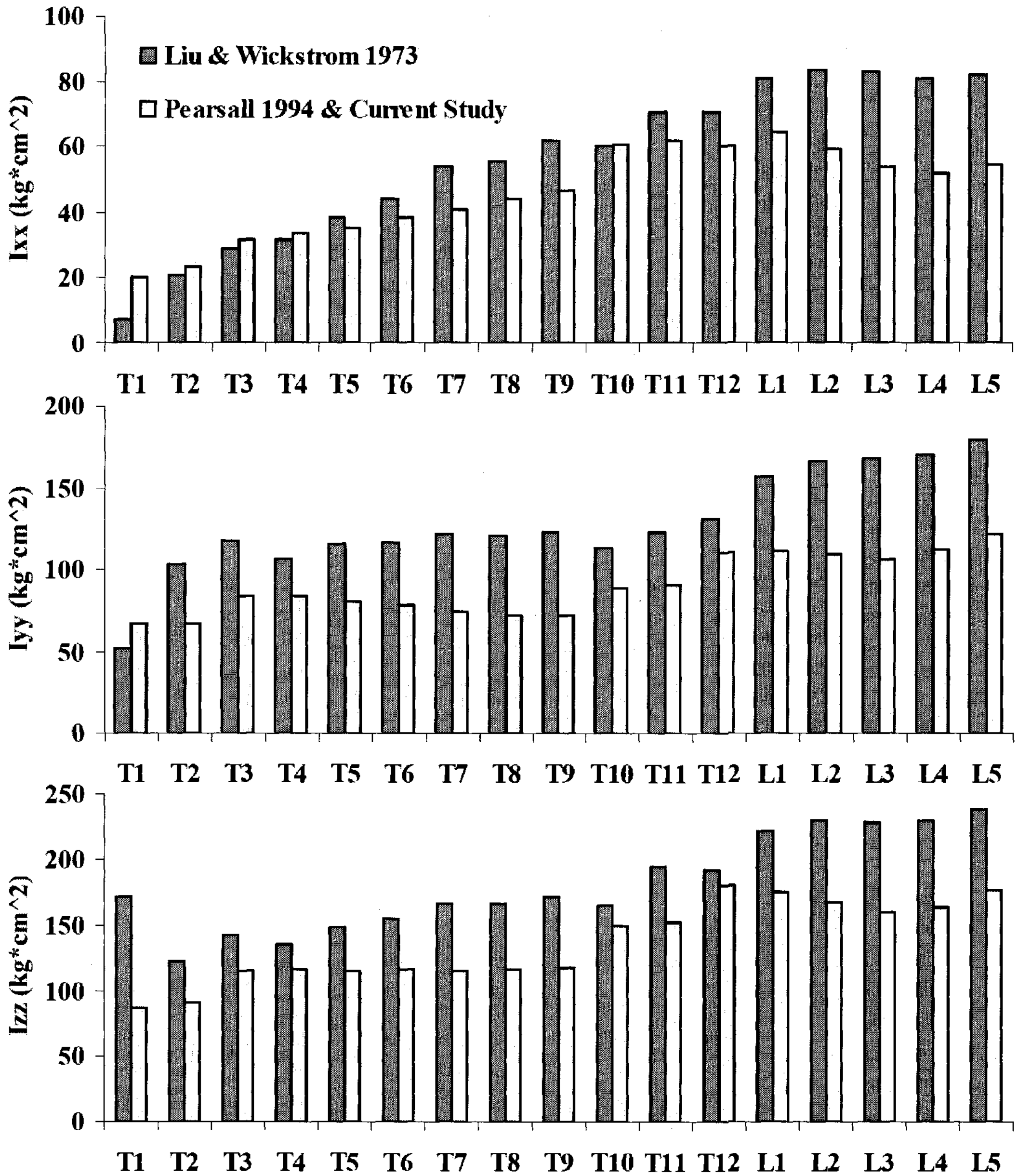

Figure 6.4 Comparison of trunk segmental mass moments of inertia at different vertebral levels considered in the current study with earlier investigations. x, y, and z represent transverse (lateral), anteroposterior, and longitudinal directions respectively (mass of the T1-L5 section were 22 and 26 kg for subjects considered in Pearsall (1994) and Liu and Wickstrom (1973) respectively)

Other numerical methods (e.g. modal analysis) were not, however, suitable for solution of differential equation of motion due to material as well as geometrical nonlinearities. The choice of time increment for implicit integration method as compared to explicit integration method is limited by considerations of accuracy rather than numerical stability (Cook et al., 2002). The accuracy of results was verified to remain unchanged with refinement of mesh or much smaller time increments (i.e. 0.0001 sec.). The damping values at inter-segmental levels assigned based on measurements of disc-body-disc units was not sufficient to attenuate the high frequency fluctuations (i.e. noise content) in required moments during simulation of fast trunk movements and vibration. They were, hence, increased by fivefold while accounting also for additional damping of other abdominal tissues surrounding the spine. Another subsequent fourfold increase was verified not to influence predictions.

Changes in ligamentous properties under various movement rates simulated in this work were not expected to be substantial in view of earlier studies on the effect of changes in loading rate on viscoelastic properties of the spine (Neumann et al., 1994; Wang et al., 2000). Experiments of Neumann et al. (1994) on the response of anterior longitudinal ligament (ALL) at different strain rates clearly indicated that the ultimate tensile load and axial stiffness of ALL did not significantly change for the displacement rates in the range of 0.1 mm/s to 1-4 mm/s; the speed of physiological lifts and that used in our studies fell within these values. More recent tensile tests of vertebra-disc-vertebra (Kasra et al., 2004) also suggested insignificant changes in moduli (at toe and linear regions) and ultimate force when strain rates were changed from 0.07-0.17 %/s to 1.4-3.3%/s. The ultimate stress and strain values did not significantly change when the strain rate further increased to 140-333 %/s that exceeds physiological range of rates in lifting. Furthermore, viscoelastic model studies (Wang et al., 2000) on the entire lumbar motion segments demonstrated that the total force in all ligaments as a function of segmental sagittal rotation remained almost unchanged when the rotation of about 9° (combined with 2000N compression and 200N shear) were applied in 0.3s, 3s or 30s. These results suggest that within the physiological speeds of trunk movement including those used in current study, the viscoelasticity of the motion segments and ligaments do not significantly influence results.

For the sake of qualitative comparison with EMG measurements, the computed muscle forces were partitioned, at the post-processing phase of the analysis, into passive and active components using a passive tension-length relationship for all muscles (Davis et al., 2003). The maximum allowable muscle stress of 0.6 MPa was assumed for all muscles neglecting the effect of activation level on this value. It is important to emphasize that the passive load-length relationship considered for muscles in the current study have absolutely no bearing at all on the predicted spinal loads and total muscle forces. The predicted activity in muscles, calculated from computed total muscle forces and presented primarily for comparison with normalized EMG data, could substantially be influenced by assumptions both on the constant maximum muscle stress (in terms of its absolute magnitude and likely variations with changes in muscle length and contraction velocity) and on the passive force-strain relationship considered in this work. In some local muscles (primarily in the subject 'Max' in slower flexion-extension; Chapter 4), passive forces estimated based on muscle strains exceeded the corresponding calculated total muscle forces in which cases they were constrained by the latter computed forces.

The issues of intra-abdominal pressure, IAP, and thoraco-lumbar fascia, TLF, have remained controversial despite numerous investigations. Intra abdominal pressure was initially proposed to reduce compressive forces on the spine during lifting (Bartelink, 1957). The postulated mechanism of IAP was that abdominal muscle contraction in the presence of a closed glottis increases IAP, hence unloading the spine both directly by pressing upwards on the rib cage via diaphragm and indirectly by generating an extensor moment on the lumbar spine that decreases the back-muscle activities (Cholewicki and Reeves, 2004; Daggfeldt and Thorstensson, 1997, 2003). In contrast, some investigators have found that co-activity of abdominal muscles increased intra-discal pressure as well as spinal loads (McGill and Sharratt, 1990; Nachemson et al., 1986). Furthermore, a recent study by Arjmand and Shirazi-Adl (2005b) pointed out to the posture dependency of IAP during isometric lifting. As for the TLF, the literature points to its minor role in lifting and that in contrast to earlier suggestions by Gracovetsky and colleagues (Gracovetsky, 1988; Gracovetsky et al., 1981) on the importance of posterior ligamentous system, PLS. It had been suggested that the PLS transmits the power of hip extensor muscles through the lumbar spine to the trunk via the posterior ligamentous system and the thoraco-lumbar fascia using one of following mechanisms:

1- Passively; when the spine is flexed and the posterior spinal ligaments are taut, the posterior rotation of pelvis due to contraction of the hip extensor muscles is then transmitted to the lumbar spine through the lumbo-sacral joints, the L5-S1 interspinous ligament, the ilio-lumbar ligaments, and thoracolumbar fascia.
2- Actively; through the crosshatching fibers of the two laminae of the posterior layer of the thorocolumbar fascia and their attachment to the lateral raphe and transversus abdominis muscle.
3- Hydraulic amplifier mechanism; by contraction of the erector muscles that increases tension in the fascia's posterior layer, which augments its anti-flexion moment.

Anatomical, experimental and model studies of Tesh et al (1987), Macintosh et al. (1987) and McGill and Norman (1988) have challenged the viability of PLS mechanism and shown that the TLF forces have been overestimated and that the contribution of TLF in resisting the trunk moment is only very small. Macintosh, et al. (1987) reported in their anatomical/biomechanical investigations that "Too little of the abdominal musculature attaches to the TLF to generate a significant tension in it. Previous calculations of the forces in the TLF have overestimated the tension developed in it because of erroneous assumptions and interpretations of the relevant anatomy."

Furthermore, Tesh et al. (1987) concluded that "These definite experiments showed that the resistance to bending in the sagittal plane offered by the abdominal muscles acting through fascial tension was of a similar magnitude to that offered by a raised intra-abdominal pressure, both being relatively small in the fully flexed position". Low abdominal activation has been observed by in-vivo studies to coincide with the time of high loading on the spine (i.e. picking the load in flexed posture) (McGill and Norman, 1986). Besides, contribution of latissimus dorsi has been shown to be small (Bogduk et al., 1998; McGill and Norman, 1986; McGill et al., 1988). These studies demonstrate that the TLF is not an important extensor of the trunk during forward lifts. Other passive ligamentous tissues of lumbar spine have, however, been accounted in the model using nonlinear beam elements (See chapter 2 to 5).

6-2-2- Kinematics measurements based on skin markers

Since simulations were done in the sagittal plane, lateral translations and rotations in the frontal plane and axial rotations in the transverse plane were all fixed. Evaluation of the segmental rotations from skin markers is recognized to involve errors in identification of vertebral positions, skin movement relative to the underlying vertebrae, and deformability of vertebrae themselves (Cappozzo et al., 1996; Lee et al., 1995; Lundberg, 1996; Shirazi-Adl, 1994; Zhang and Xiong, 2003). Zhang and Xiong (2003) reported a difference of 1.1 to 5.8 degree between the external marker-defined inter-segmental motions and corresponding internal vertebral rotations calculated using a kinematics model. Cappozzo et al (1996) showed that improper positioning of the skin marker (e.g. markers located directly on the skin above anatomical landmarks such as greater trochanter, lateral epicondyle of the femur, head of fibula, and lateral malleolus undergo) resulted in displacements relative to the underlying bone in range of 10-30 mm. Due to these inherent errors, the measurements were used to evaluate only temporal variations of the pelvic tilt and thorax rotation while the intervening lumbar segmental rotations were evaluated based on the partitioning of the difference between foregoing measured rotations using the relative values reported in the literature (i.e. ; ~11%, 15%, 14%, 18%, 22%, and 20% for T12-L1 to L5-S1 levels).

Different percentages have been reported in earlier studies to partition lumbar rotation among its vertebrae (Clayson et al., 1962; Dvorak et al., 1991; Frobin et al., 1996; Gracovetsky et al., 1995; Hayes et al., 1989; Lee et al., 1995; Pearcy et al., 1984; Plamondon et al., 1988; Potvin and McGill et al., 1991; Shirazi-Adl, 1989, 2006; Yamamoto et al., 1989) (Fig. 6.5). Arjmand (2007) performed a sensitivity analysis to evaluate the effect of two completely different relative partitions (i.e. between works of Potvin et al (1991); 10%, 11.88%, 11.88%, 18.9%, 26.1%, 21.24%; and Frobin et al (1996); 11.46%, 15.03%, 17.71%, 18.09%, 20.89%, 16.81%) on predicted results of kinematics-based approach in a static loading condition. He found that the passive spine moment and muscle forces at different lumbar levels changed significantly, while the compression and shear forces at L5-S1 disc (i.e. the most critical level) were different negligibly by only one Newton. He further reported that stability of the spine was not affected by these relative variations in lumbar rotations. Alternatively, some investigators have developed regression equations to predict inter-segmental rotations based on some measured variables (e.g. lumbar posture, L1-S1 skin distraction) (Chen and Lee, 1997; Lee and Chen, 2000; Lee et al., 1995). It is interesting to mention that wearing abdominal belts or lumbosacral orthoses has been shown to change the partitioning of lumbar rotation among various vertebrae (Tuong et al., 1998; Woldstad and Sherman, 1998).

Although these proportions were assumed constant during the entire lifting tasks, such may not necessarily be true in vivo as the relative demand at different levels could vary during lifting. However, the recent experimental study of Wong et al (2004), showed a linear variation of segmental rotations in flexion-extension movements that supports the constant proportion of rotations taken throughout the motion in our work. Speed of movement has neither been found to significantly influence vertebral rotations

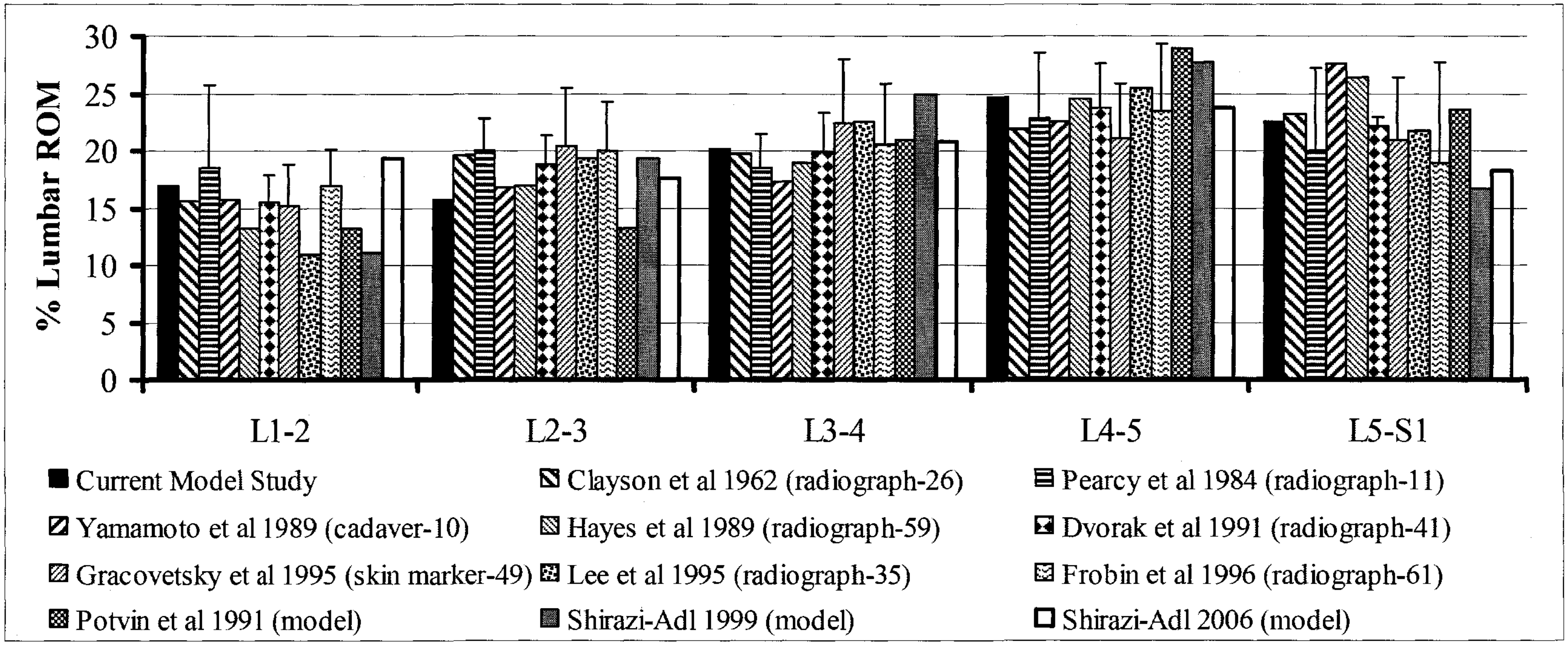

Figure 6.5 Reported mean lumbar rotation at each lumbar motion segment expressed as a percentage of total lumbar rotation. Standard deviations are depicted when available.

(Zhang et al., 2003), hence justifying application of such assumption in simulation of dynamic trunk movement under different velocities.

Temporal variations of pelvic translations (i.e. vertical and horizontal) were evaluated using the coordinates of the S1 marker on the skin and its relative rigid rotation with respect to the S1 itself and was subsequently prescribed along with the pelvic tilt into the model at the S1 base. To prescribe measured rotations in the model, kinematics data of some specific subjects (i.e. one in lifting and three in flexion-extension studies) rather than the mean of all subjects were considered. This was done due mainly to noticeable variations in duration of lowering/lifting phases in between subjects for lifting tasks (Chapter 2 and 3) and different lumbar/pelvic rhythm for flexion-extension tasks (Chapter 4). Another reason was to avoid loss of important inertial loads and introduction of artificial high acceleration content to input data due to averaging of kinematics data. Moreover, the TableCurve software (SYSTAT Software Inc., San Jose, CA) was used to fit the best smooth curves on kinematics data of each trial (R^2 >0.98 for all cases).

6-2-3- Muscle architecture

Two groups of muscles have been considered in our model; local muscles that originate from lumbar vertebrae and insert into the pelvis and femur (comprise of 46 muscles) and global muscles that originate from thorax, span over lumbar and local muscles and insert into the pelvis (comprise of 10 muscles). Anatomy of these muscles including insertion points and physiological cross sectional area (PCSA) in the present study, given in Appedix A, were taken mainly from the study of Stokes and Gardner-Morse (1999) which was, in turn, based on previous studies (Bogduk and Macintosh et al., 1992; Bogduk and Pearcy et al., 1992; Bogduk and Twomey, 1991; Panjabi et al., 1991) as well as the database of the Visible Human Project™ (National Library of Medicine, Bethesda, MD).

In current study, the transverse abdominal, latissimus dorsi, lumbodorsal fascia, intersegmental and multisegmental muscles were neglected while the oblique abdominal muscles were represented by straight single lines rather than curved sheets of muscle. Consideration of several fascicles instead of just one for oblique muscles (EO and IO) has influenced the estimated spinal loads significantly in asymmetric lifting tasks but only slightly in symmetric ones (Davis and Mirka, 2000). Further information regarding to trunk muscle anatomy can be find elsewhere (Bogduk and Twomey, 1991). While local muscles were modeled as straight lines between their respective insertion points (Macintosh et al., 1993), curved muscle paths were considered for global extensor muscles by wrapping them over all T12-S1 vertebrae whenever in the course of motion (flexion) their distances to associated vertebral bodies reduced more than 10% of their initial distances. This allowed for a maximum of ~10% reduction in muscle lever arms at different levels during flexion in accordance with published data in the literature (Jorgensen et al., 2003; Macintosh et al., 1993). The wrapping of global muscles occurred at all levels under larger flexion angles and resulted in curved paths with more realistic lever arms at different levels. Wrapping contact forces between vertebrae and muscles were found to be function of trunk flexion, external load, and lumbar posture. Simulation of wrapping muscles without the proper consideration of wrapping contact forces in equilibrium and stability analyses is questionable adversely affecting the accuracy of simulations. Few dynamic studies (McGill and Norman, 1986) also considered curved paths for global muscles but with no indication of the associated wrapping contact forces and their effect as additional loads on results. It is to be noted that while neglecting wrapping of global extensor muscles would have been resulted in significantly larger spinal loads with a less stable system, it would not at all change the conclusions of our studies regarding the safer lifting technique, effect of velocity, and trunk response to vibration and shock.

6-2-4- Optimization algorithm

In continuation and accordance with our earlier studies, we considered the cost function of sum of cubed muscle stresses in all simulations done in current work. For this purpose, Lagrange multiplier method was used to analytically solve the optimization problem which guarantees the convergence of results to a global minimum. Arjmand et al (2005c) have evaluated effects of different linear (i.e. summation of muscle stresses, summation of muscle forces, axial compression and double-linear) and nonlinear (i.e. summation of cubed muscle stresses, summation of square muscle stresses, summation of square muscle forces and muscle fatigue) cost functions on results of the kinematics-based approach by comparing measured EMG activity in global muscles with those predicted by the model under isometric conditions. The study advocated cost functions of sum of squared and cubed muscle stresses in yielding plausible results comparable with measured EMG activities and disc pressures. They further showed that predictions of cubed muscle stress were less dependent on the inequality constraint equations.

6-2-5- Stability of the spine

The majority of the works on the stability of dynamical systems is based on a formal definition of stability given by Liapunov (Leipholz, 1970). The stability of a system can be characterized by the eigen values of the system. In fact, a given linear system is stable if and only if it has no eigen value with positive real part (Inman, 2006). Moreover, the system will be asymptotically stable if and only if all of its eigen values have negative real parts. For a conservative system of the form

$$M\ddot{x} + Kx = 0$$

If M and K are positive definite, the eigen values of K are all positive, and hence the eigen values of the system are all purely imaginary (i.e. the system is stable). It has been shown that adding a symmetric and positive definite damping matrix to such stable system makes it asymptotically stable (Inman, 2006). If the damping matrix is only positive semi-definite, the system still remains stable.

Since the damping matrix in our model was symmetric and positive definite, static stability routines like perturbation and buckling analyses as well as frequency analysis were all applicable and employed to evaluate spinal stability at loaded deformed configurations of the system. It is to be emphasized that the stability margin of a mechanical system alters constantly with the applied loads and, hence, should be determined at loaded (stressed) configurations and not at initial, un-deformed, state. In this manner, the spinal system ceases to be stable when its lowest natural frequency reaches zero or equivalently as the second variation of the potential energy vanishes. Static perturbation and frequency analyses, in this case and as expected, yielded identical critical muscle stiffness coefficient values when instantaneous stressed configurations were used

6-3- Comparison with Experimental Measurements

It would be convenient to validate a model if the model predicts outputs that can be measured independently. However, principle justification of some models (including our spine model) is that measuring those variables is either difficult or infeasible at all. Hence, full validation of our human spine model is currently impossible due to lack of such adequate knowledge on spinal structure and loading. Nevertheless the model can be partially validated by comparing certain model outputs that are measurable currently. This includes comparison of model predictions with measured vertical ground reaction force, measured acceleration at different spinal levels, and measured EMG activity of select trunk muscle (not done in current study). Measurement of disc pressure or fixation forces, although have been performed under static conditions (Wilke et al., 2001; Wilke et al., 2003), may cause injury or pain for dynamic tasks, hence impossible currently.

The predicted temporal variation of reaction forces in all three subjects at three velocities, performing flexion-extension (Chapter 4), agreed well with our measurements bearing in mind that the latter measured curves account also for the inertia of lower body. Due to the extended knee position during flexion-extension movements, the measured inertial effects should nevertheless be attributed primarily to the upper body (above pelvis) rather than the lower body (Lindbeck and Arborelius, 1991; Plamondon et al., 1995). The computed and measured results were also in good agreement with reported vertical ground reaction forces measured during lifting weights at different speeds (Dolan et al., 1999). There was also a good agreement between predicted and measured (Seidel et al., 1997) accelerations at the L3 and head levels in the study of trunk response to vibration (Chapter 5) (Fig. 5.4). Noticeable differences in the force magnitudes at the seat-subject interface at the times of peak acceleration in this latter study (Fig. 5.4) could partly be due to the dynamic contributions of thighs and legs that were absent in the model but present in measurements. The predicted dynamic component of base reaction force reached the maximum of 400 N at peak positive acceleration (i.e. 10 ms^{-2}) whereas the minimum of -460 N (i.e. equal to the trunk weight) at the peak negative acceleration (i.e. -12.2 ms^{-2}).

It has been suggested that the component validation might be an indication of whole model validation (Lewandowski, 1982). Extensive data on such validation of our spine sub-models can be found in earlier works of our research team (Arjmand and Shirazi-Adl, 2006; El-Rich et al., 2004; Pop, 2001; Shirazi-Adl et al., 2005; Shirazi-Adl et al., 2002). Ligamentous thoraco-lumbar and lumbar spine have been shown experimentally to buckle at loads of about 20 N and 100 N respectively (Crisco and Panjabi, 1992; Crisco et al., 1992; Lucas and Bresler, 1961; Patwardhan et al., 2003; Patwardhan et al., 2001). In comparison our model buckled at load of 103 N when it was applied at the center of T12 vertebra. Linear perturbation analyses at standing posture while applying vertical loads at center of T12 or T1 vertebra showed a critical load of about 120 or 20 N in sagittal plane respectively. First natural frequency of the system in vertical direction was ~12 Hz that agrees well with the reported value of ~11 Hz (Kong and Goel, 2003). The incorporation of buttocks in the current model with proper stiffness and damping values (Aimedieu et al., 2003; Kitazaki and Griffin, 1997a) diminished the first vertical natural frequency of the seated trunk under gravity from ~12

Hz to ~5.5 Hz, in agreement with earlier measurement studies (Kong and Goel, 2003; Pope et al., 1990).

6-4- Clinical and Biomechanical Implications

6-4-1- Velocity of trunk movement

Generally, faster trunk movements have been associated with a decrease in trunk strength but increases in trunk moments, muscle coactivity, muscle forces, and spinal loads (Davis and Marras, 2000; Dolan et al., 1994; Dolan et al., 1999; Granata and Marras, 1995a, b; Lariviere and Gagnon, 1998; McGill and Norman, 1985). Inertia effects of the trunk and external load have been indicated to play a noticeable role at the onset of a lift with jerky movements (Lariviere and Gagnon, 1998). Peak axial thrust, ground reaction force and joint moment have been reported to significantly increase with lifting speed regardless of lifting technique (Dolan et al., 1999). Lumbar moment have been shown to be on average 9% (not significant), 21% (p=0.005) and 42% (p=0.0001) larger than static moment in slow (mean velocity in a complete lifting/lowering cycle, 0.2 m/s), normal (0.4 m/s), and fast (0.8 m/s) speed conditions respectively with no effect of lifting technique on results (de Looze et al., 1994). While faster paces were generally associated with larger net spinal moments (Davis and Marras, 2000), certain pattern of motion were shown to result in inertial forces that can even reduce moments (McGill, 2007; McGill, 1997). Granata and Marras (1995a) depicted that trunk moment decreased by increasing lifting velocity mainly because the subject pull the load close to their bodies earlier in excretion than during slower lifts. Marras and Mirka (1993) found that muscle activities decreased by increasing trunk acceleration with most decrease being in erector spinae. However, no changes in intra abdominal pressure as well as increase in muscle co-activities were found to be associated with trunk acceleration.

For the lifting tasks considered in current study (Chapter 2 and 3) and due mainly to relatively slow velocity of lift (i.e., lowering and lifting periods each lasting ~2s), our results showed a negligible effect of inertia forces on trunk moment and spinal loads except at three time intervals; the beginning and end of the tasks as well as a short period after picking the load up (Fig. 2.7) which agrees with earlier observations (Holmes et al., 1992). Apart from these periods, a quasi-static analysis would, hence, yield sufficiently accurate results with no real need to account for inertia forces. Our results also demonstrated that the inertia of the trunk, and not that of the load, was the major factor for the observed differences in these three time periods. In a different lifting condition, however, the latter has been estimated to be more important than the former (McGill and Norman, 1985). Other dynamic studies have also reported minor inertial contributions in the computed net moment at the L5-S1 for tasks with similar durations (Davis et al., 1998; Fathallah et al., 1998; Kingma et al., 2001; Toussaint et al., 1995).

In contrast to our lifting simulations, inertial loads were found quite dominant during fastest flexion-extension movements (Chapter 4). The significance of inertial forces in fastest movement as compared with slower ones was apparent both in computed temporal variation of base reaction force in the model and in measured ground reaction forces via the force-plate in all three subjects studied (Fig. 4.3). Sharp initial downward acceleration followed by a deceleration when reaching the peak trunk flexion in the forward flexion phase plus the initial upward acceleration and subsequent deceleration to reach back to the upright position in the extension phase resulted in the predicted pattern of reaction force at the base. The temporal variation of axial reaction force at the base for the fastest flexion-extension movement (Fig. 4.3) followed closely those of net moments and compression forces (Fig. 4.6) suggesting the importance of inertia in fastest motion for all subjects. To further help identify the inertial component of spinal loads from those due to gravity loads, some results of Chapter 4 are represented in a revised format showing the variations with respect to the normalized trunk flexion rather than the normalized time (Figs. 6.6 to 6.8).

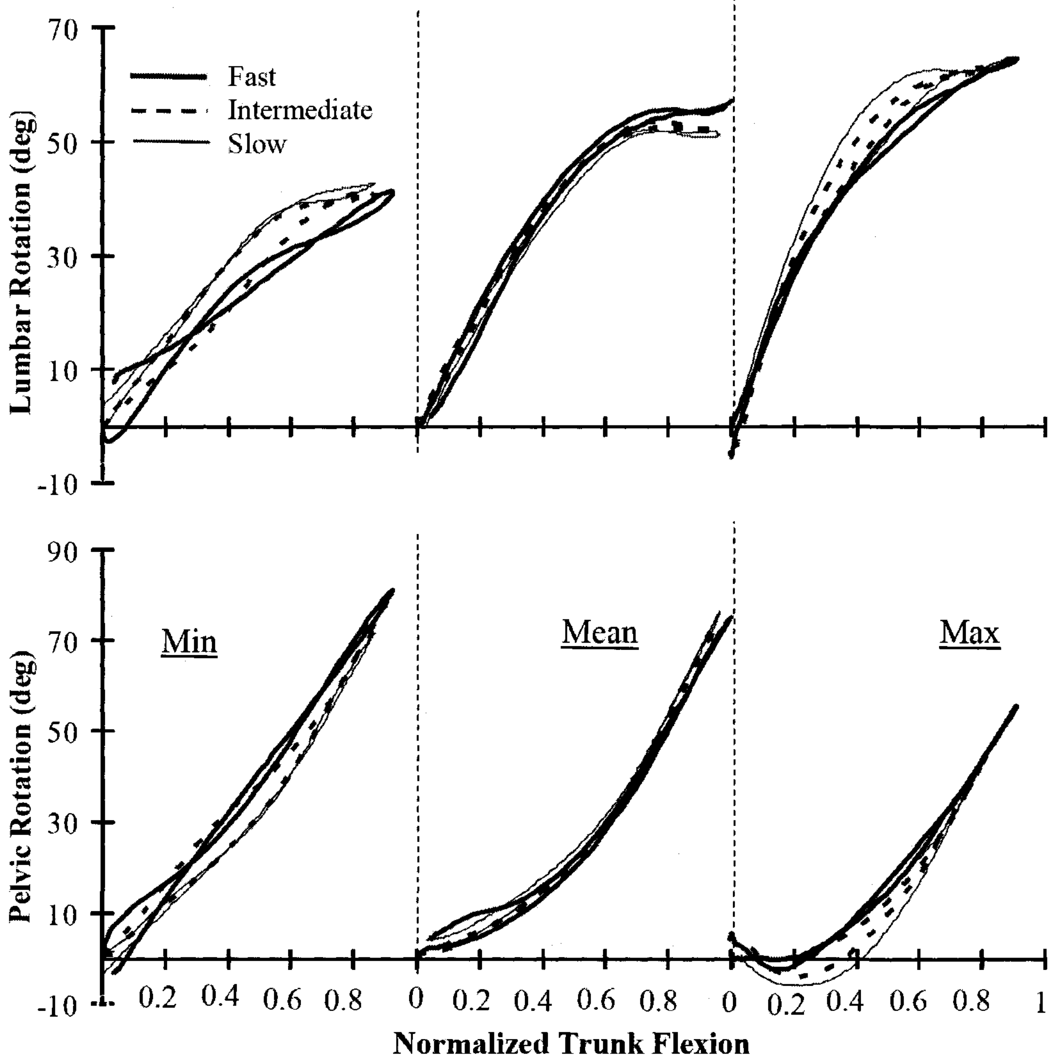

Figure 6.6 Prescribed lumbar and pelvis rotations in the model for three subjects 'Min' (left), 'Mean' (middle), and 'Max' (right) at three movement velocities based on in vivo measurements. (Results are all normalized with respect to peak trunk flexion of 132°)

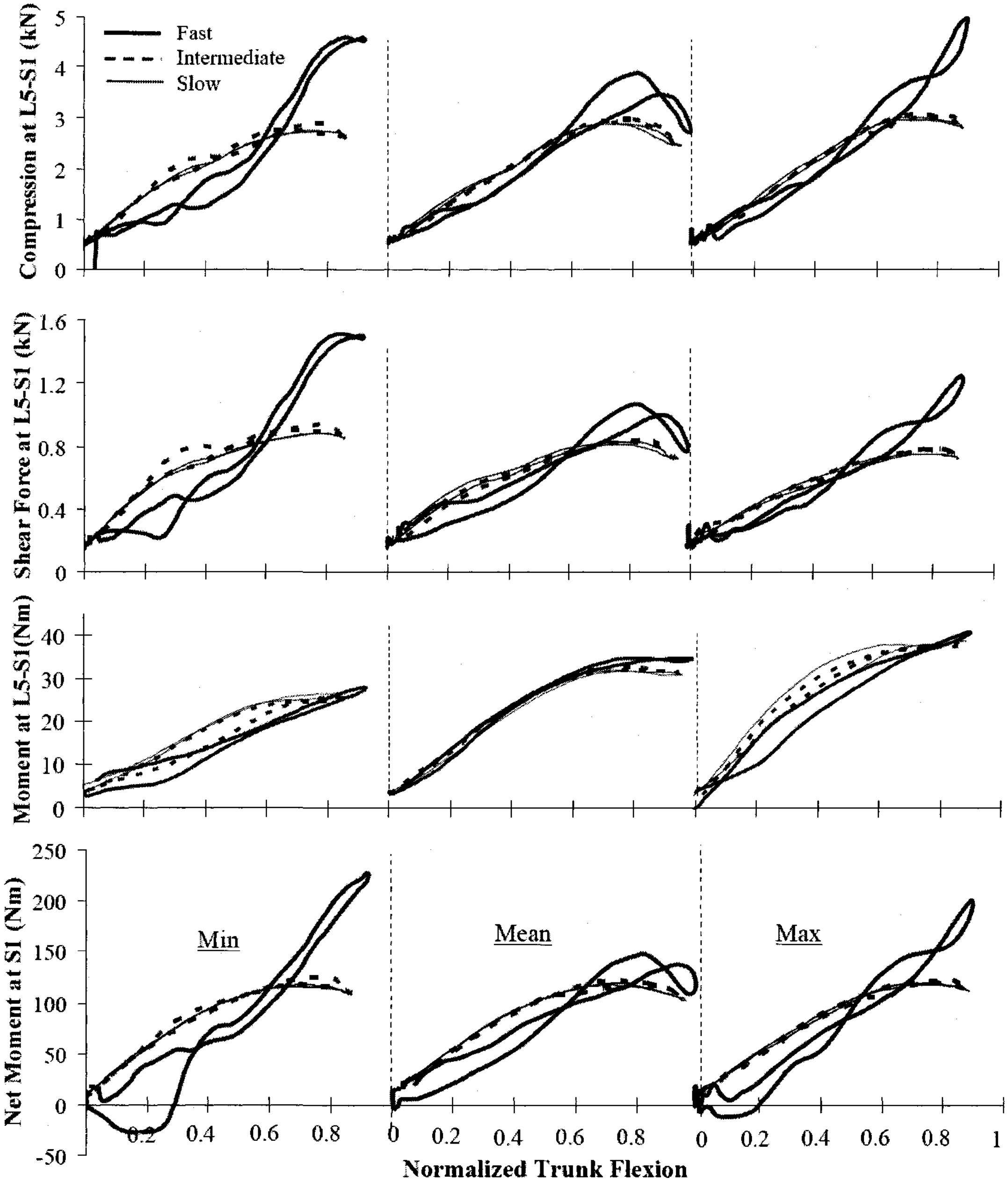

Figure 6.7 Predicted temporal variation of internal spinal loads at the L5-S1 disc mid-height in local directions (compression force, shear force and sagittal moment) and the net moment at the S1 (to be resisted by all muscles and passive ligamentous spine) for three subjects 'Min' (left), 'Mean' (middle), and 'Max' (right) at different movement velocities. (Results are all normalized with respect to peak trunk flexion of 132°)

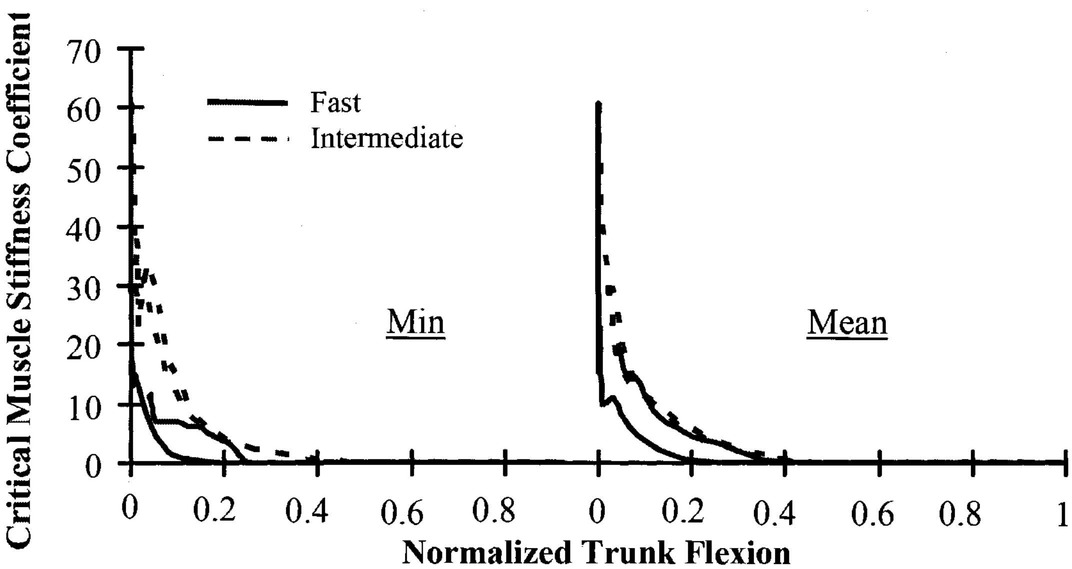

Figure 6.8 Predicted temporal variation of minimum (critical) muscle stiffness coefficient, q, required for the trunk stability for subjects 'Min' (top) and 'Mean' (bottom) at two different flexion-extension movement velocities. (Results are all normalized with respect to peak trunk flexion of 132°)

Contribution of gravity in spinal loads was rather minimal in vertical vibration of trunk in sitting posture (i.e. Chapter 5) as compared to that in flexion-extension tasks (i.e. Chapter 4) that altered as a function of trunk flexion. The excitation at the base substantially influenced, via the inertial and muscle forces, the spinal compression and shear forces. The maximum spinal compression and shear forces predicted at the L5-S1 increased significantly from the initial values (static; due to gravity alone) of, respectively, 535 and 222 N in the erect posture and 717 and 297 N in the flexed posture to peak values (dynamic; due to both gravity and inertia) of 1131 and 508 N in the erect posture and 1608 and 771 N in the flexed posture. Role of inertial loads would likely further increase in presence of larger shock values of ~ ±2-6 g reported in off-road and industrial vehicles (Robinson, 1999).

Our results suggest that velocity of movement influences the spinal stability depending on the trunk flexion angle. The spinal stability improved under the fastest movement at the beginning of the flexion-extension that could be due to the activity of abdominal muscles (Fig. 4.7 and 6.8). Under the same trunk flexion angle (Fig. 6.8), the fastest movement actually yielded the most stable configuration of all in forward flexion for both subjects and in backward extension in the subject 'Mean'. At larger trunk angles and in agreement with our earlier isometric simulations (Arjmand and Shirazi-Adl, 2006), the spine was found much more stable with no need for a muscle stiffness. The stability margin at large flexion angles, evaluated based on the minimum eigen value or perturbation analysis, suggests only a slightly more stable configuration under the fastest movement despite the much greater net moment in this latter case (especially in the subject 'Min'). The fastest movement also reached in both subjects the critical q=0 at much smaller trunk angles (Fig. 4.7 and 6.8). During the whole body vibration (i.e. Chapter 5), the spinal stability was affected more by the lumbar posture and abdominal co-activities than the base acceleration (Fig. 5.7, 5.8).

6-4-2- Lifting technique

The controversy on a safer lifting technique persists due to the complexity of the problem, confounding parameters (e.g., dependence on changes in the posture and load positioning) and over-simplifications (assumptions involving kinematics, constraints, geometry, material properties, loading, dynamic characteristics, etc.) in model studies. In this work parallel in vivo studies were performed to collect kinematics required as input data to drive the kinematics-based model. The entire forward-backward movements were carried out over 4-5s with either squat or stoop techniques but no instructions on the lumbar posture.

Thorax/pelvis/lumbar rotations were all larger in stoop lifts compared to those in squat lifts (Fig. 2.2). This resulted in additional flattening of the lumbar spine in stoop lifts as well as increased wrapping contact forces (Table 2.2) and moment-carrying contribution of passive ligamentous spine and trunk muscles. Moreover, despite identical lever arms considered (based on measurements) for the external load of 180 N at the T12 level, the net moments and hence muscle forces and internal loads were all greater in stoop lifts than in squat ones; e.g., maximum net moments of 200 N-m and 160 N-m were predicted at the L5-S1 level for stoop and squat lifting, respectively. Same trends were also found in the absence of external loads or even when the external load was shifted by ~88mm mm closer to the body in the stoop lift in order to reach the same lever arm with respect to the S1 as that considered in the squat lift.

Therefore, results of this study appear to firmly suggest the squat lift as the safer lifting technique in reducing the net moment, muscle forces and internal ligamentous loads at all levels. In this regard, however, the literature remains inconclusive as some similarly report smaller net moment and trunk load in squat lifting (Buseck et al., 1988; de Looze et al., 1994; Hagen and Harms-Ringdahl, 1994; Potvin and McGill et al., 1991) while others indicate otherwise (de Looze et al., 1998; Dolan et al., 1999; Lindbeck and Arborelius, 1991; Troup et al., 1983; van Dieen et al., 1994). A more recent study by Kingma et al (Kingma et al., 2006), has pointed out that no single lifting technique can be advised for all lifting conditions.

The reduction in net moment in squat lifts, found in this work under all cases with and without external load, is due primarily to smaller pelvic and lumbar (and hence thorax) rotations in this technique resulting in much reduced net moments from the mass of the upper body and the external load about the L5-S1 (Fig. 2.8). Variations in location of external loads and rotations of pelvis and lumbar spine from a lift to another, as expected in different studies, are important and could substantially influence the results and subsequent comparison of lifting techniques towards identification of the optimal one. The biomechanical advantages for the squat lifts in our study would become even more apparent had a smaller lever arm for the external load been considered in these lifts (Bendix and Eid, 1983; Troup et al., 1983).

The relative stability of stoop lift versus squat lift altered over time in forward flexion phase of the task (Fig. 3.4). Initially, the stability margin was larger in squat lift due mainly to greater activity of abdominal muscles at the beginning of the task (Figs. 3.2 and 3.3). This was, however, reversed in stoop lift as the abdominal activity disappeared and flexion angle increased in order to pick the load (Fig. 3.4) thereby generating greater stiffness in both active and passive sub-systems.

Results of previous works on extensor muscle activities in stoop lifts usually demonstrate two peaks; the first and smaller one occurs in the lowering phase while the second and larger one during the returning lifting phase (Haig et al., 1993; Lariviere et al., 2000; McGill et al., 1999; Paquet et al., 1994; Peach et al., 1998). Our predictions on active extensor muscle forces also show similar variations during the tasks (Fig. 2.4). Due to the relatively small flattening of the lumbar spine (T12-S1) considered in the model (remaining <26°), no flexion relaxation was observed which would otherwise have influenced the results in the final periods of the lowering phase of the study.

6-4-3- Lumbar posture

The advantages in preservation or flattening (i.e., flexing) of the lumbar lordosis during lifting tasks are not well understood. Lifting has been categorized as either squat or stoop with often no recording of changes in the lumbar lordosis which may influence the risk of injury (McGill et al., 2000; Potvin and Norman et al., 1991). The kyphotic lift (i.e., flexed lumbar spine) is recommended by some as it utilizes the passive posterior ligamentous system (i.e., posterior ligaments and lumbodorsal fascia) to their maximum thus relieving the active extensor muscles (Gracovetsky, 1988; Gracovetsky et al., 1981). In contrast, however, others advocate lordotic and straight-back postures indicating that posterior ligaments cannot effectively protect the spine and an increase in erector spinae activities during the early phase of the lift is beneficial in increasing stability and reducing segmental shear forces (Delitto et al., 1987; Hart et al., 1987;

Holmes et al., 1992; McGill, 1997; Vakos et al., 1994). Moderate flexion has been recommended by model (Shirazi-Adl and Parnianpour, 1999, 2000) as well as experimental studies (Adams et al., 1994) to reduce risk of failure under high compressive forces. As the lumbar posture alters from a lordotic one to a kyphotic one, the effectiveness of erector spinae muscles in supporting the net moment (due to smaller lever arms (Jorgensen et al., 2003; Macintosh et al., 1993; Tveit et al., 1994)) and the anterior shear force (due to changes in line of action (McGill et al., 2000)) decreases while the passive contribution of both extensor muscles and the ligamentous spine increases (Arjmand and Shirazi-Adl, 2005a, 2006; Macintosh et al., 1993).

In an earlier combined in vivo-model study on the effect of changes in the lumbar curvature on trunk response in isometric lifts with identical thorax rotations (Arjmand and Shirazi-Adl, 2005a), the maximum segmental shear/compression forces and activity in extensor muscles occurred in the lordotic posture while the maximum segmental flexion moment occurred in the kyphotic posture. The kyphotic postures exploited primarily the passive ligamentous/muscle force components while the active muscle forces played more important role in lordotic postures. The study advocated the free style posture or a posture with moderate flexion as the posture of choice in static lifting tasks when considering both internal spinal loads and active/passive muscle forces. One must note that in that study the thorax rotation remained nearly the same irrespective of changes in the lumbar curvature. In the current study (Chapter 2), however, the thorax rotations of 66.9° and 70° in stoop lifts respectively with and without 180 N in hands were much greater than corresponding rotations of 38.4° and 49.7° in squat lifts (Fig. 2.8). Although we did not investigate the effect of changes in lumbar curvature in dynamic stoop and squat lifts, the conclusions of the previous isometric study advocating a flattened lumbar spine and current dynamic one advocating a squat lift (involving more lordotic lumbar curvature) do not contradict each other due to the crucial effect of posture (i.e., thorax and pelvic rotations) on results.

Lumbar posture was also found to significantly alter trunk response to whole body vibration and shock. The 10° flattening in the lumbar lordosis from the erect to flexed posture in the sitting position under the whole body vibration, considered in the current study, increased the net moment, muscle forces and passive spinal loads and hence the risk of tissue injury while on the other hand substantially improved the trunk stability. The latter effect was due to the increased stiffness of muscles and passive ligamentous spine in the flexed posture. The moment at the S1 as well as compression and shear forces at the L5-S1 increased in the flexed posture by, respectively, 6.8 Nm, 182 N, and 75 N at the onset of vibration and by 19 Nm, 482 N, and 240 N at the time of peak positive acceleration. Minimum muscle stiffness coefficient of the erect posture dropped (i.e. improving the spinal stability) from ~ 87 to 43 at the beginning of the analysis (static position) and from ~ 140 to 92 at the time of peak positive acceleration when the subject flattened their back in a moderately sitting flexed posture. It appears hence that an increase in trunk stability can be achieved only at the cost of higher passive loads and hence greater risk of tissue injury suggesting a trade-off in muscle activities.

6-4-4- Flexion-relaxation

Upon progressive forward flexion of the trunk from the upright standing posture towards the peak flexion, a partial or complete silence in EMG activity of superficial extensor muscles has been recorded. This phenomenon has been well documented in healthy asymptomatic subjects and is called as the flexion–relaxation phenomenon (FRP) (Floyd and Silver, 1951) that may persist even in presence of weights carried in hands. The FRP has been recorded to occur at about 84-86% of peak voluntary flexion in slow movements irrespective of the magnitude of load in hands (Sarti et al., 2001). The presence and absence of the FRP could also be used as a signature to discriminate LBP patients from healthy controls as in former group the FRP is frequently absent (Kaigle et al., 1998; Kippers and Parker, 1984; Watson et al., 1997). The FRP assessment has, thus, been suggested as a valuable clinical tool to aid in the diagnosis and treatment of LBP patients (Colloca and Hinrichs, 2005). The effect of changes in the velocity of trunk flexion-extension movements on the flexion-relaxation FRP has been investigated. Greater flexion-extension movement velocity has been reported either to reduce the frequency of FRP observation in repetitive flexion extension (Mathieu and Fortin, 2000) or to delay its occurrence (Sarti et al., 2001). No effect on the trunk angle at which the FRP occurs was, however, found in other studies (Mathieu and Fortin, 2000; Steventon and Ng, 1995).

In order to explain the partial or full relaxation in back muscles in large trunk flexion postures, several hypotheses have been put forward; the load is transferred from extensor muscles to passive tissues (Floyd and Silver, 1951; McGill and Kippers, 1994), from superficial muscles to deeper ones (Andersson et al., 1996), or from lumbar extensors to thoracic ones (Toussaint et al., 1995). Since the FRP is likely related to the relatively large axial strain (or elongation) in extensor muscles during forward flexion, it is expected to also depend on the lumbar rotation and pelvic-lumbar rhythm. The relative activity of various back muscles in deep flexion movements remain controversial as some suggest relaxation in global extensor muscles (Mathieu and Fortin, 2000; Sarti et al., 2001) while others report relaxation only in lumbar extensor muscles (McGill and Kippers, 1994; Toussaint et al., 1995). Using deep wire electrodes, Andersson et al. (1996) reported silence only in superficial lumbar erector spinae muscles with activity remaining in deeper ones.

Consider for example subject “Mean” in the current study (Chapter 4) while performing flexion-extension at slowest pace As the trunk flexes forward from upright posture, initially both active and passive components of forces in global extensor muscles increase with the former reaching its peak values at about 45° (Fig. 6.9). Thereafter up to the trunk flexion of about 95°, active forces in thoracic extensor muscles diminish despite the continuous increase in net external moment reaching its maximum of 118 Nm. On the contrary, passive muscle forces as well as passive ligamentous moment increase, though at a very slow rate, throughout the movement to peak lumbar flexion (Figs. 6.9 and 6.10). The progressive relieve in activity of global back muscles is due, therefore, to higher passive contribution of muscles and ligamentous spine as the lumbar rotation increases.

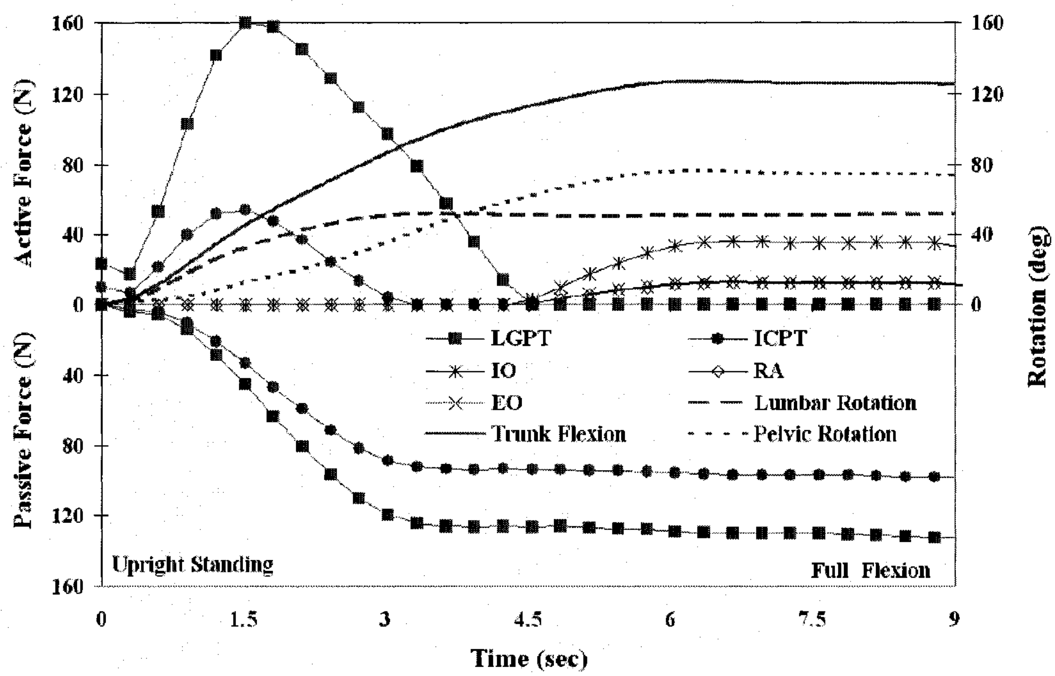

Figure 6.9 Predicted temporal variation of active and passive forces in global trunk muscles (on each side of the body) as well as measured trunk and pelvic rotations with time advancing in a slow forward flexion task from upright standing posture (left) toward full flexion (right). Total lumbar rotation is the difference between trunk and pelvic rotations. The initial drop in global extensor muscle forces is due to the inertia of the upper body at the onset of the task (Arjmand et al., 2008).

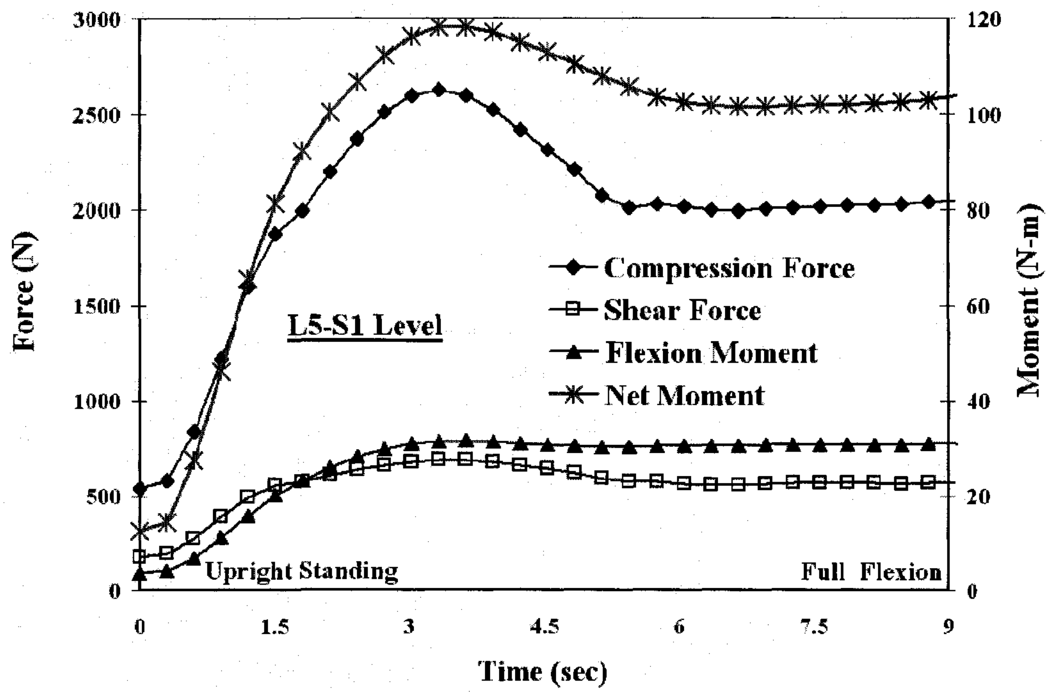

Figure 6.10 Predicted temporal variation of passive ligamentous loads as well as net external moment at the L5-S1 disc level with time during a slow forward flexion movement (Arjmand et al., 2008).

As the trunk flexion exceeds about 95° (at about 3.3 sec), lumbar rotation (Fig 6.9) and consequently both passive muscle force and moment resistance of the ligamentous spine remain nearly unchanged while activity of back muscles continue to drop. In this case, the reduction in net external moment due to the decrease in the effective lever arm of the trunk centre-of-mass is the primary cause in progressive decrease in back muscle activities. Global longissimus, LGPT, and iliocostalis, ICPT, become completely silent at trunk flexion angles of about 114° and 95°, respectively. With the exception of the multifidus that only partially relaxed, local lumbar muscles also demonstrated full relaxation in activity, but at larger flexion angles as compared with global extensor muscles (Fig. 6.11). Abdominal muscles remain silent up to trunk flexion angles of about 115° at which angles global extensor muscles become inactive. Subsequently, abdominal muscles (especially internal oblique, IO, Fig. 6.9) initiated activation up to the peak rotation generating flexor moments that offset the moments produced by the passive component of back muscle forces. In other words, abdominals were activated to increase and maintain the large flexion angles. Activities in abdominal muscles have also been reported in earlier studies during full flexion as extensor muscles become silent (Mathieu and Fortin, 2000; McGill and Kippers, 1994; Olson et al., 2006).

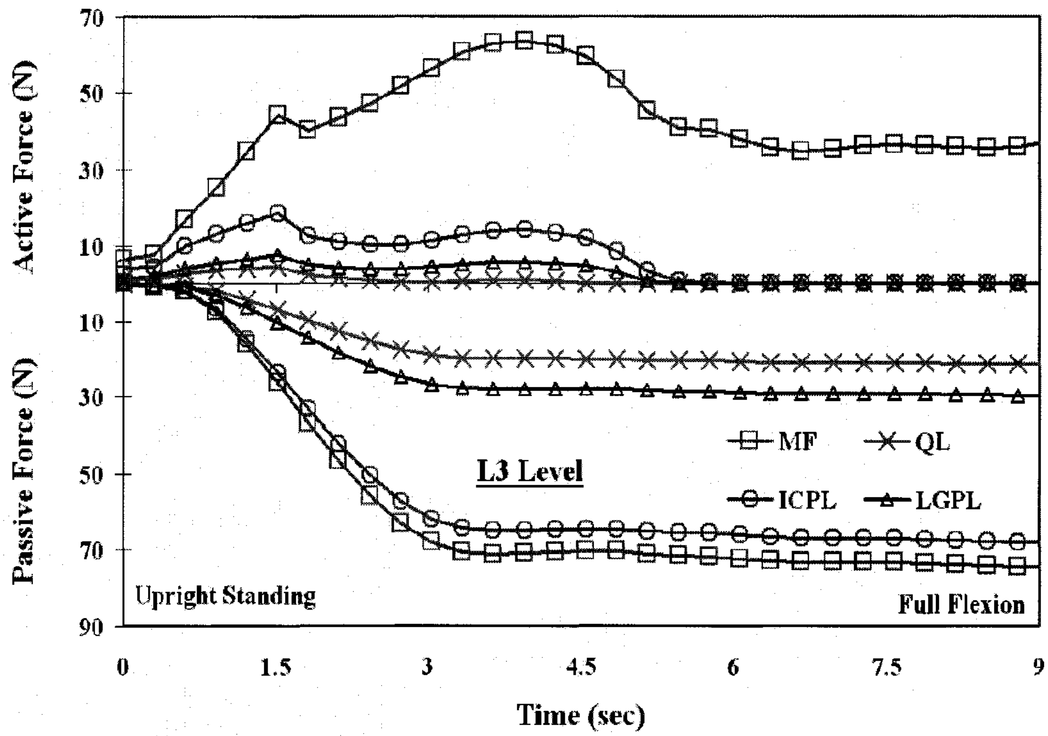

Figure 6.11 Predicted temporal variation of active force in local trunk extensor muscles (on each side) attached to the L3 level with time during slow forward flexion movement. Muscle forces at other lumbar levels (not shown) follow similar trends (Arjmand et al., 2008).

It should be emphasized that the choice of force-strain used to represent the muscle passive response in the model would influence the extent of decrease in extensor muscle activity and concurrent increase in abdominal activity. In contrast to slower movement in which partial or full FR occurred at full flexion, local and global extensor muscles had their maximum activities at full flexion during the fastest tasks during which time abdominal muscles were silent.

6-4-5- Role of passive properties

Loads on the human spine are influenced not only by the gravity, inertia and external loads but also by the stiffness of the ligamentous spine and forces in trunk muscles. Changes in passive properties of the ligamentous spine have been demonstrated to affect equilibrium and stability of the spine in isometric (Arjmand and Shirazi-Adl, 2006; Gardner-Morse et al., 1995; Stokes and Gardner-Morse, 2003) and dynamic (Cholewicki and McGill, 1996) tasks. Greater passive stiffness would reduce muscle forces resulting in smaller compression forces on the spine while larger compression forces would occur under greater muscle forces compensating for a decrease in the passive resistance. Passive properties of the human spine may alter with ageing, degeneration, injury, surgical intervention or even during daily activities as a function of single/combined loads and time (Shirazi-Adl, 2006; Wang et al., 1996).

Alterations in passive stiffness of the ligamentous spine substantially influenced results in both stoop and squat lifts, though more so in former than in latter (Table 3.1, Figs. 3.2 and 3.3). Contribution of passive components of the spine and musculature in counterbalancing net moments of gravity, inertia and external loads reached the maximum values of 56.4%, 24.5% in stoop lift right before and after picking the load, respectively. These values increased to 67.6% and 31.2% as passive stiffness increased by 20% whereas they decreased to 46.9% and 18.6% and further to 37.9% and 14.1% as the passive stiffness reduced by 20% and 40%. In squat lift, due to smaller lumbar rotations (Fig. 3.1), similar trends but at smaller magnitudes were predicted. The effect of changes in the ligamentous passive properties on stability was more evident in the earlier forward flexion phase of the lifting task in which higher passive properties demanded lower muscle stiffness coefficients in order to maintain stability. As the load was lifted, the critical muscle stiffness coefficient dropped to nil irrespective of the lifting technique and alterations in passive properties. These predictions highlight the crucial role of passive and active components in stabilization of the spine.

Gardner-Morse et al (1995) reported that critical active muscle stiffness coefficient changed from 4.5 to 4.7/3.7 as the spine stiffness decreased/increased by 10% at maximum extension efforts in standing posture. In a study by Cholewiki and McGill

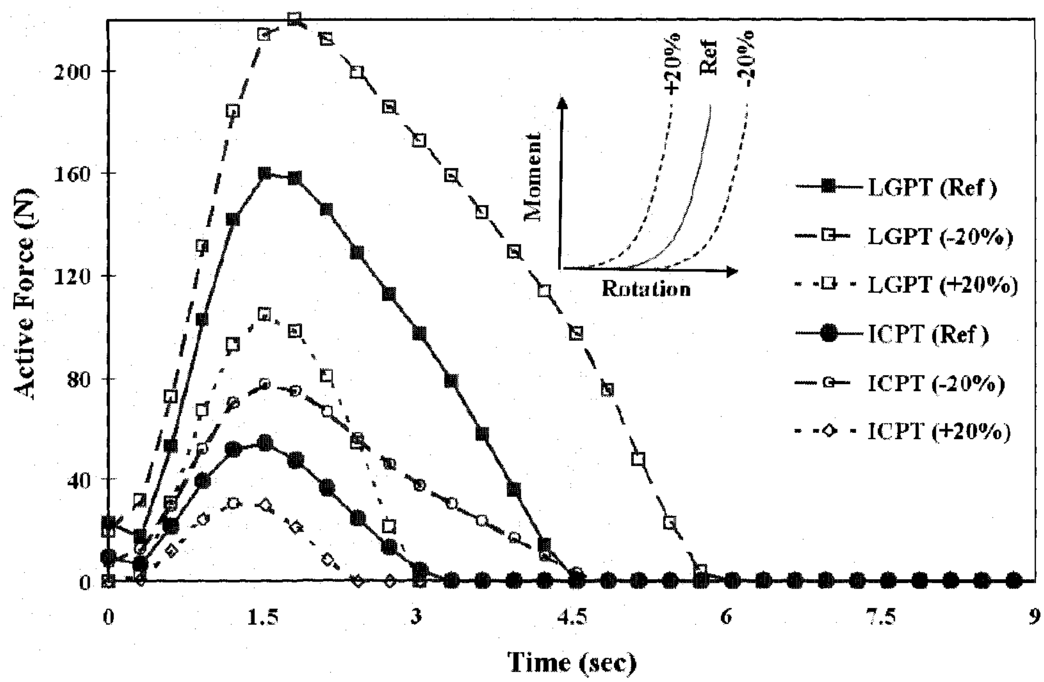

Figure 6.12 Predicted temporal variation of active force (on each side) in extensor global muscles with time from upright standing posture toward full flexion as the bending stiffness of motion segments is altered by ±20% at all levels (Arjmand et al., 2008).

(1996), stability was also found to be ameliorated by increasing spine stiffness under squat lifting. Results of a more recent investigation (Arjmand and Shirazi-Adl, 2006) on changes in spinal stability due to alterations in passive rotational stiffness of the spine showed that the stability of the spine deteriorated as its stiffness decreased by 10 or 30% despite greater predicted muscle activities.

In another study (Arjmand et al., 2008), we examined the likely role of passive properties of the ligamentous spine and muscles on flexion-relaxation phenomenon (FRP). A decrease in the passive ligamentous spine stiffness (case with – 20%) (Figure 6.12) markedly increased activity in global extensor muscles and diminished that in abdominal muscles at larger flexion angles. A reverse trend was computed when the passive contribution was increased resulting in an earlier and greater activity in abdominal muscles concurrent with an earlier flexion relaxation in extensor muscles. Similar effects were also predicted as the passive contributions of extensor muscles were altered. A decrease in passive stiffness due to an injury or joint relaxation could delay flexion relaxation in extensor muscles. The abdominal muscles are also affected by such changes.

6-4-6- Wrapping of global extensor muscles

Unlike in upright postures in which pathways of global extensor muscles can accurately be assumed as straight lines between insertion points, such may not be the case in tasks involving large lumbar flexions. In latter tasks, a straight line assumption for global muscles could violate kinematics constraints by penetrating into intervening hard/soft tissues. The wrapping contact mechanism acts to enforce kinematics constraints on deformations whereby the penetration of global muscles into underlying muscles and vertebrae are prevented as the spine flexes forward. Such constraints change the orientation of global muscles and result in contact forces in between global muscles and the spine at different levels.

Consideration of curved global muscles and their interaction with spine was found to have significant effects on predicted spinal loads and muscle forces at larger flexion angles under the load when the wrapping contact forces reached their maximum values. In this case, the maximum muscle forces and consequently spinal compression forces diminished (Chapter 3). The improvement in spinal stability was, however, evident throughout the range of flexion and that despite larger muscle forces in the model with straight global muscles. This deterioration in system stability in presence of straight muscles is due to the generated larger compression forces on the spine and smaller lever arms of global muscles. On the other hand, the contact forces between wrapping global muscles and the spine along the lumbar spine could increase the system stability. Simulation of wrapping without the proper consideration of these contact forces in equilibrium and stability at deformed configurations of the spine is not, hence, reliable adversely affecting the accuracy of simulations.

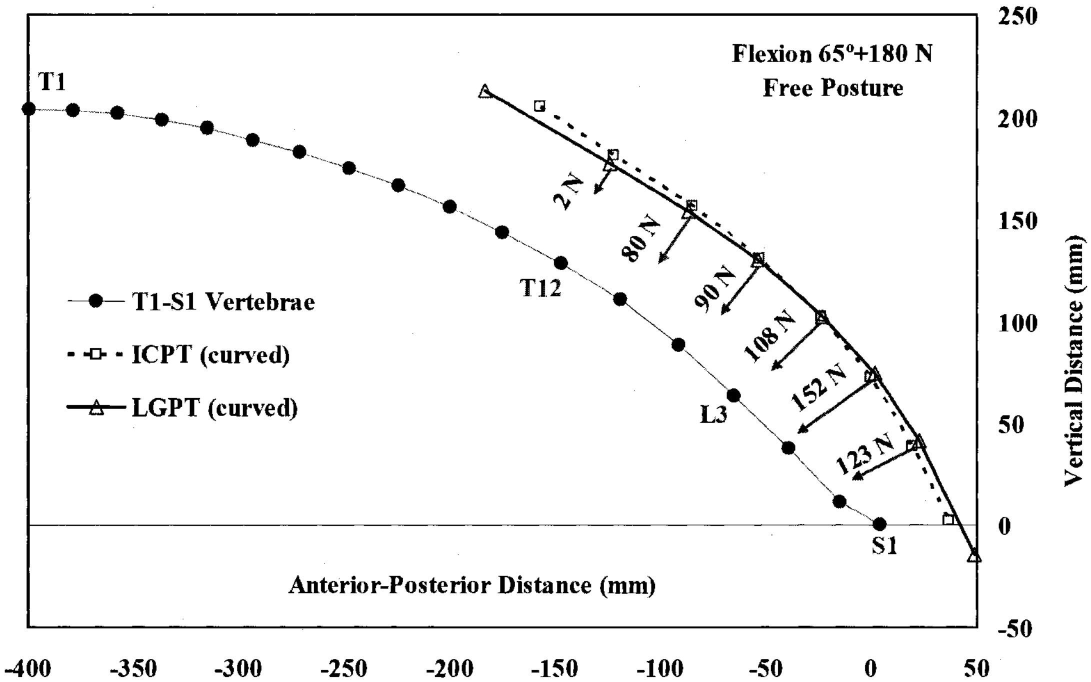

Figure 6.13 Magnitude and direction of wrapping contact forces on the spine due to wrapping of global muscles (Longissimus Thoracis pars thoracic, LGPT, and Iliocostalis Lumborum pars thoracic, ICPT) for the case with no reduction in lever arms from upright posture under trunk flexion of 65° with 180 N in hands (Arjmand et al., 2006)

Incorporation of muscle wrapping in isometric lifting has also generated significant changes in results depending on the trunk flexion, lumbar posture and external load (Arjmand et al., 2006). It was shown that wrapping contact forces, with the exception of that at the L5/S1 level, acted approximately perpendicular to the compressive axis of the spine (Fig. 6.13) thus primarily increasing anterior shear forces with smaller effect on the axial compression. The effect of contact forces in increasing anterior shear force was especially obvious at the lower levels (L3–S1) (see (Arjmand et al., 2006)) while at the upper levels this effect disappeared due to more horizontal line of action of global muscles in cases with curved paths. It was further found that the wrapping contact forces generally increased as the trunk flexion angle increased and kyphotic posture was adopted (see (Arjmand et al., 2006)). These were due to the fact that the magnitude of the contact force depends not only on the muscle force but on the change in the muscle line of action at a contact point; the larger this change is the greater the contact will become.

6-4-7- Lumbar/Pelvic rhythm (motion pattern)

During flexion–extension movements and compared to the lumbar rotation, the pelvic rotation has been reported to become predominant at the end of flexion and beginning of extension phase (Paquet et al., 1994). Others, however, suggest that lumbar and pelvic rotations act simultaneously during flexion and/or extension phases (Nelson et al., 1995; Sarti et al., 2001). The three subjects (Fig. 6.6) considered in this work (Chapter 4) demonstrated generally sequential lumbar/pelvic rotations (with greater lumbar rotations at the beginning and end of tasks especially in 'Max' and 'Mean' subjects) at all movement velocities.

Faster trunk movement significantly increased spinal loads and moments, however, the extent of these increases markedly alters from a subject to another due mainly to different movement patterns. The temporal variation of internal spinal loads and net moments (Fig. 4.6) clearly indicate that their corresponding peak values, irrespective of the velocity of movement, may not occur at the time of peak trunk flexion angle. In other words, the foregoing relative effects of velocity of movement on internal loads and net moments could substantially diminish or even reverse when considering the subject 'Mean' (Chapter 4) at the instance of peak trunk flexion angle. This could be due to the relatively longer duration as well as the rather peculiar kinematics of this subject in fastest movement (Figs. 4.2 and 4.3) that rapidly attained his peak lumbar rotation and then preserved it for about a second time while the pelvic and trunk rotations reached their maximum values and began to reverse for the extension phase of the movement both together. This observation may in part help understand the existing controversy in the published literature on the effect of movement velocity on trunk biodynamic.

As for lifting simulations (Chapter 2 and 3), the relative lumbar/pelvic rotations during lowering/lifting phases showed greater contributions in all cases from the pelvis than the lumbar spine (by as much as two-fold) and remained within the range of data reported in the literature (Esola et al., 1996; Granata and Sanford, 2000; McClure et al., 1997; Porter and Wilkinson, 1997). Thorax and pelvic rotations were both larger in stoop lifts compared to those in squat lifts (Fig. 2.2) resulting in greater lumbar (T12-S1) rotations in stoop lifts by 10.4° and 5.2° in cases with and without 180 N load in hands, respectively. These additional flattening of the lumbar spine in stoop lifts increased the wrapping contact forces (Table 2.2) and moment-carrying contribution of passive ligamentous spine and trunk muscles.

6-4-8- Trunk response to vibration and shock

Long-term occupational exposure to whole body vibration (WBV) has been reported to increase the risk of lumbar spine disorders (Bovenzi, 2006; Mansfield, 2005; Wilder and Pope, 1996). A clearer picture on the causal role of whole body vibration environments in back disorders can emerge following an improved understanding on associated trunk biodynamics. The iterative kinematics-driven finite element approach was used to solve the redundant passive-active trunk system at the seated position subject to an input random whole body vibration. The time variations of trunk muscle forces, spinal loads and trunk stability were evaluated. This novel investigation would also respond to the recognized need for development of more anatomically detailed biomechanical models of trunk in whole body vibration biodynamics (Seidel, 2005; Seidel and Griffin, 2001).

A single L2-L3 lumbar motion segment has been measured to yield an axial natural frequency of ~32 Hz (Kasra et al., 1992). This natural frequency drops to ~18 Hz when considering a finite element model of two lumbar motion segments, L4-S1 (Goel et al., 1994) and furthermore to ~11 Hz as the entire lumbar spine is considered (Kong and Goel, 2003). The incorporation of buttocks in the current model with proper stiffness and damping values (Aimedieu et al., 2003; Kitazaki and Griffin, 1997a) diminished the first vertical natural frequency of the seated trunk under gravity from ~12 Hz to ~5.5 Hz, in agreement with earlier measurement studies (Kong and Goel, 2003; Pope et al., 1990). With no constraint on sagittal rotations, vibration analysis at the loaded configuration resulted in the lowest frequency of ~1 Hz that is also in agreement with the reported values in the literature (Kitazaki and Griffin, 1997a).

Although trunk muscle forces and spinal loads remained relatively low during the vibration input considered in our studies, shock contents in base excitation with accelerations exceeding 2 g on one hand and deteriorations in passive resistance of the spine and muscle reflexive response due to fatigue on the other hand can generate a condition involving high risk of back injury. Moreover, trunk posture as well as stability demands can further impose larger spinal loads and consequently higher risk of injury. For example, the shock value of ~ ± 1 g in the current study increased the net external moment at the S1 by ~ ±160% and the axial compression at the L5-S1 by ~110% when compared with the corresponding static results.

The stiffness of the element simulating buttocks may need to be modified accounting for alterations in the contact area between buttocks/thighs and the seat (Kitazaki and Griffin, 1997b). Although no such changes were considered in the current study, the first natural frequency of the model was found to be highly dependent on the buttocks stiffness demonstrating the likely indirect effect of posture on the natural frequency. Rigid buttocks increased the system resonant frequency from 5.5 Hz to 12.4 Hz, the acceleration response especially at the time of peak acceleration, and net moments. Reverse trends were found as the buttock stiffness decreased.

6-4-9- Muscle co-activities and spinal stability

In the current study, the stability of the spine at each instance of time was investigated using nonlinear analyses assuming different muscle stiffness coefficient values, q. As q decreased, the loss of stability at a critical q value was also confirmed by parallel perturbation and free vibration analyses at deformed stressed configurations. At each time instance, the critical q value in a specific case was identified as the lowest eigen value in free vibration analyses approached zero and the displacement under unit force perturbation analyses increased substantially, These latter analyses should necessarily be performed at the instantaneous deformed states of the system in order to avoid over-estimation of stability margin in such nonlinear and imperfect structure.

Activation in muscles has opposite effects on the spinal stability; on one hand, it increases compression force on spine (i.e. destabilizing role); but on the other, it offers greater stiffness associated with larger activation (i.e. stabilizing role). Furthermore, due to larger lever arms (Arjmand et al., 2007a), abdominal muscles even with much smaller forces are more efficient than extensor muscles in stabilizing the trunk. Results of current study in all simulations demonstrated much lower critical q values in presence of abdominal activities, being agonistic or antagonistic. Simultaneous involvement of agonist active abdominal muscles and antagonist passive extensor muscles at the beginning of the task resulted in a more stable configuration when considering the fastest flexion-extension (Chapter 4) as well as both stoop and squat lifts (Chapter 3). The positive role of antagonistic activities in enhancement of the spine stability has been recognized in the literature (El-Rich et al., 2004; Granata and Orishimo, 2001; Potvin and O'Brien, 1998). The magnitude of zero computed as he critical muscle stiffness coefficient at larger trunk flexion angles, especially in presence of an external load, indicates that the system is sufficiently stable when accounting for its passive and stress-dependent matrices (Cook et al 2002).

In order to further increase spinal stability and decrease critical q values at configurations with rather small trunk flexion in the neighbourhoods of neutral standing position (e.g. whole body vibration), one should artificially introduce (or increase) coactivity in the model. Such consideration in the current study (Chapter 5) resulted in considerable amelioration of spine stability (i.e. a drop of 67 unit in minimum muscle stiffness coefficient at the time of peak positive acceleration) at the cost of slight increase in spinal loads (e.g. 168 N in compression force and 24 N in shear force at the time of peak positive acceleration) during whole body vibration. These results further highlight the opposing effects of greater abdominal co-activity in improving system stability while increasing the spinal loads and risk of tissue injury.

6-4-10- Tissue tolerance

The compression strength of lumbar motion segments has been reported to be in the range of 2-10 kN (Brinckmann et al., 1989; Jager and Luttmann, 1997; Ortoft et al., 1993). Jager and Luttmann (1991) reported values of 5.81±2.58 kN for males and 3.97±1.5 kN for females based on relatively large sample populations. The strength in shear force has been reported to be >1 kN (Cyron et al., 1976; Miller et al., 1986) while that in flexion moment exceeds 70 Nm (Miller et al., 1986; Neumann et al., 1992; Osvalder et al., 1993). Notwithstanding the effect of strain rate, existing injuries/degeneration, fatigue and combined loading on these strength values (Seidel et al., 1998), lower risk of injury could be associated with conditions that yield smaller loads on spine. Hence, considering the mild shock contents of ~ ±1 g in vibration input data of current study (Chapter 5) as compared with reported larger shock values of ~ ±2-6 g in off-road and industrial vehicles (Robinson, 1999), it is likely that spinal loads in latter vibration environments approach and even exceed strength limits causing injury.

Similarly, safer lifting techniques could be established based on the premise that excessive compression forces, shear forces or flexion moments in the ligamentous spine could cause injury. While maximum passive moments at the L5-S1 (i.e. 33 and 16 Nm in stoop and squat lifts respectively) were below threshold values of the motion segments, compression and shear forces at the L5-S1 disc (i.e. compression forces of 4800/4000 N, and shear forces of 1600/1400 N) reached or even exceeded the reported tolerance limits in stoop/squat lifts. Accordingly, this study advocate a squat lift over stoop lift in reducing the risk of fatigue and injury to passive and active components without necessarily deteriorating the spinal stability.

Results of this study have also highlighted the risk of flexion-extension movements when performed at the fastest pace to full voluntary trunk flexion (e.g. maximum spine moment of 40 Nm, compression force of 4600 N, and shear force of 1500 N in fast movement of subject "Min"). Although the presence of external loads in hands, not considered in this work, could further increase the internal loads and hence the risk of injury but it is highly unlikely that subjects voluntarily carry out the flexion-extension movement to the maximum flexion angle as fast when they carry loads in hands.

6-5- Concluding Remarks

A detailed trunk muscle architecture along with nonlinear properties of the ligamentous spine, wrapping of global extensor muscles and trunk dynamic characteristics (inertia and damping) were used in our kinematics-driven model to evaluate muscle forces, spinal loads and trunk stability under stoop/squat lifting, flexion-extension movements, and random base whole body vibration. The predictions satisfied kinematics and dynamic equilibrium conditions at all levels and directions. Our study confirmed the crucial role of movement velocity and lumbar rotation on response dynamics, muscle activation, FR, internal spinal loads and trunk stability. Whole body vibration with high acceleration content (i.e. shock) was found to substantially increase spinal loads and deteriorate stability of the system. Spinal loads and stability were further shown to be sensitive to lifting technique, lumbar posture, lumbar/pelvic rhythm, load and muscles lever arms, wrapping of global muscles, activity/co-activity levels, and passive properties of the spine and trunk muscles.

6-6- Future Studies

6-6-1- Asymmetric lifting

Manual material handling task with twisting and lateral bending of the trunk (i.e. a situation with high incidence in work place), has been associated with higher risk of back injury (Marras, 2005; NIOSH, 1997). Traditionally, these tasks were analyzed by ergonomic experts using link segment models (Plamondon et al., 1995). Application of kinematics-based approach would provide a clearer image of spine biomechanics under asymmetric lifting tasks by generating more accurate results. The outcome of such simulation can consequently provide a better ground for design of lifting guidelines. To do so, experiments should be conduct initially to acquire trunk kinematics for model studies as well as muscle activities and ground reaction force for validation of predictions.

6-6-2- Dynamic stability of spine

There are certain issues in assessment of dynamic stability of the spine that can be addressed in future. First, due to symmetry of damping matrix in our current investigation, static stability criteria (i.e. perturbation, and natural frequency analyses) at loaded deformed configuration were used to assess spinal stability. However, such assumption with an asymmetric damping matrix is not valid any more. Formulating of an eigen value problem by conversion of motion equations to state variable equations and a comprehensive assessment of its complex eigen values will enable us to evaluate spinal stability at each instance of any dynamic activities.

Second, the nonlinear force-stiffness relation in reflexive muscle activations has been demonstrated to noticeably enhance the trunk stability (see for example (Moorhouse and Granata, 2007; Shadmehr and Arbib, 1992)). In a recent study (Zeinali Davarani et al., 2007), such contribution of muscle spindle has been shown to decrease the error in positioning and velocity of trunk movement. In the current study, however, only a linear force-stiffness relation was considered that has been suggested to results in higher energy expenditure as compared to nonlinear reflexive response (Franklin and Granata, 2007). Moreover, the optimization routine considered in current work is merely subjected to equilibrium equation as equality equation and therefore unable to predict co-activity (of abdominal or back muscles). Such antagonistic muscle activity can be predicted by our model if some stability constraints are included in the optimization problem. Further enhancement in muscle response modeling to account for both intrinsic and reflexive muscle stiffness as well as consideration of stability in optimization procedure can hence set as objectives for future studies.

6-6-3- Inclusion of cervical spine and simulation of athletic and whiplash injuries

Because of the potentially catastrophic and life-threatening nature of cervical spine injuries (CSIs), there is need to better understand the biomechanics of CSIs. The more severe CSIs associated with athletics can be attributed to compressive forces from axial loading (Swartz et al., 2005). Whiplash injury associated with vehicle crashes form another CSI risk factor due to both extreme motions of head and neck and excessive compressive force. The kinematics-based approach can be applied to cervical spine by developing a proper nonlinear finite element model of cervical spine along with neck muscles to quantify muscles response along with resulting cervical loads to a variety of excitation input. Such model can be subsequently included in existing model of thoraco-lumbar spine for future studies on spine response to whole body vibration and shock.

6-6-4- Failure of spine due to long term low magnitude whole body vibration

Muscle reflexive response has been suggested to deteriorate under long term vibration, challenging the spinal stability calling for compensatory muscle activity that will result in larger spinal loads (Gade and Wilson, 2007). Long term exposure to whole body vibration, on the other hand, can result in weakening of the tissue tolerance due to fatigue (McGill, 2007). In-vivo and in-vitro experimental studies may be needed to quantify trunk response to long term vibration for further model studies.

6-6-5- Optimal thorax/lumbar/pelvis posture under whole body vibration (seat design)

Vehicle seat vibrations with high acceleration content have been suggested to cause more back injury than those with low vibration levels that contribute more to the time averaged measures of exposure defined in ISO 2631 (Stayner, 2001). It has been shown that some off-road and military vehicles contain peak accelerations in the range of 2 to 6 g (Robinson, 1999). Such high magnitude acceleration contents along with compensatory antagonistic activity of trunk muscles due to stability requirements likely increase spinal loads to 5000 N and beyond that can cause injury. Investigation of optimal trunk posture to alleviate such deteriorating conditions can further provide better database for seat design.

6-6-6- Lumbar spine response to impact due to fall or collision

Simulation of lumbar response to large horizontal and vertical acceleration impact and the likely role of muscle intrinsic and reflexive response along with passive musculoskeletal system can help to better understand the mechanism of lumbar injury due to fall or collision. Slipping and tripping are more likely to happen while performing pulling or pushing activities. On the other hand, it has been estimated that nearly half of

MMH tasks consists of pushing and pulling manoeuvres. Despite the significant role of active-passive musculoskeletal system in spine response to impact (Liu, 1982), such contribution has been overlooked in most of corresponding biomechanical model studies. Application of kinematics-based approach to such conditions would certainly shed some light on trunk musculoskeletal system response to impact.

6-6-7- Subject specific modeling

Personal as well as psychosocial factors have been associated with back pain and have been suggested to influence the pain behaviour (Marras, 2005; NIOSH, 1997). Some recent studies have shown that these factors alone cannot generate back pain, but in combination with other physical risk factor can substantially increase the risk of back injury (Marras, 2005). Subject specific modeling will enable us to evaluate such effects by including these factors to dynamic models. It is of significant importance to understand how trunk biodynamic change adaptively due to fatigue and as a consequence of aging, learning and rehabilitation.

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APPENDIX A

KINEMATICS-BASED APPROACH

A.1 Overview

Since certain salient details on the models used have not been given in the "Method" section of papers presented in the current work in compliance with existing journals limits on the number of words/pages/figures/tables, this section is devoted to provide additional information. The kinematics-based approach was introduced by Shirazi-Adl and colleagues in 1997. Since then the method has extensively been used to evaluate trunk muscle forces and spinal loads as well as spinal stability in isometric sagittaly-symmetric standing and flexed postures (Kiefer et al 1997, El-rich et al 2004, Arjmand and Shirazi-Adl 2006). Similarly, dynamic kinematics-based approach was formulated in the current study to investigate trunk biodynamic under daily activities. This approach exploits kinematics data to drive the model by generating additional equations at each spinal level that alleviate the kinetics redundancy in the system. Initially, measured trunk kinematics (sagittal plane rotations at different vertebral levels and base translations at either S1 or buttocks) along with external/gravity loads are prescribed into a nonlinear finite element model of the thorocolumbar spine (Fig. A.1). Implicit algorithm with unconditionally stable Hilber-Hughes-Taylor integration operator (Hilber et al 1987) is used to solve the nonlinear transient problem, resulting in the time variation of reaction moments at each vertebral level to be balanced by muscles attached to that level. To resolve the remaining redundancy at each level, an optimization approach with the cost function of minimum sum of cubed muscle stresses is used. The inequality equations relate to unknown muscle forces remaining positive and greater than their passive force components (calculated based on instantaneous muscle length and a tension-length relationship (Davis et al 2000)) but smaller than the sum of their respective maximum active forces (i.e., 0.6 MPa times muscle's physiological cross-sectional area, PCSA (Winter 2005)) and the passive force components. At the end of each iteration within each increment of time, the updated penalty of muscle forces in shear and axial directions along with the external/gravity loads are applied to the spine and the procedure is repeated until the convergence is achieved (i.e. calculated muscle forces in two successive iterations remain almost the same).

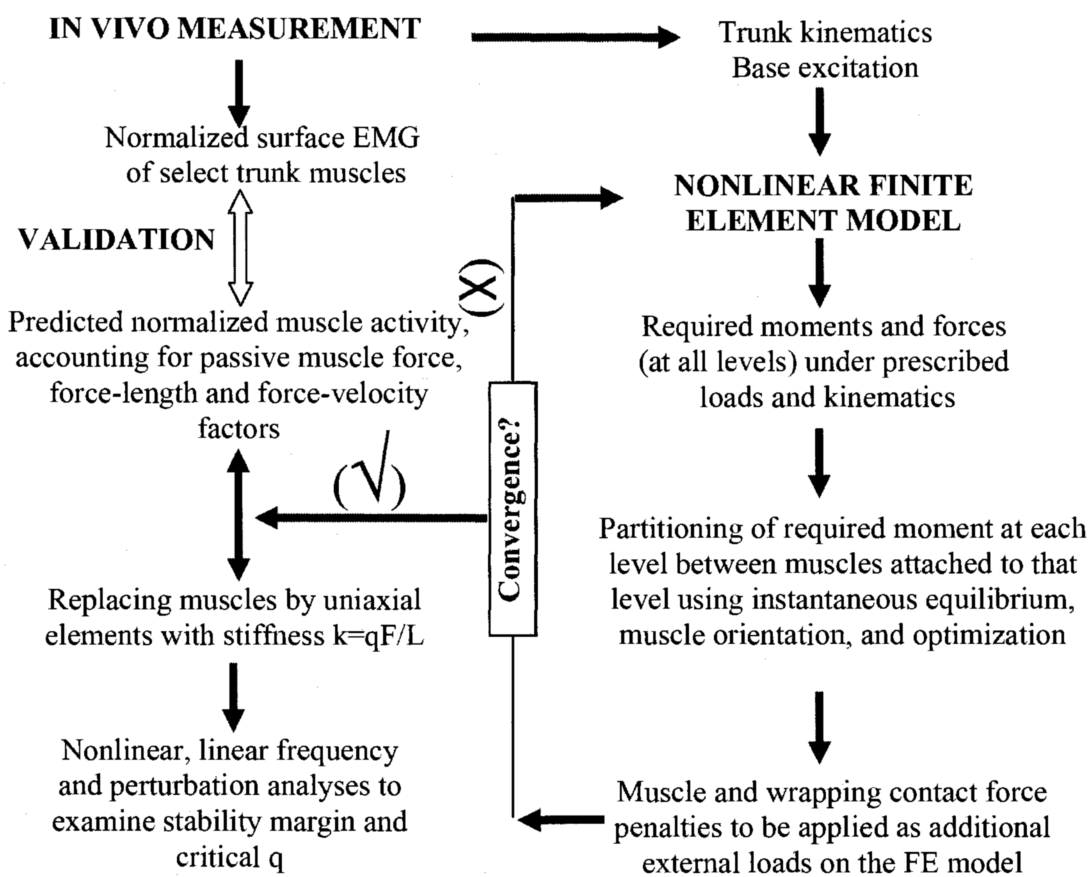

Figure A.1 Flowchart for the application of the Kinematics-based approach used to determine trunk muscle forces, internal loads and spine stability. Convergence is attained if calculated muscle forces in two successive iterations remain the same.

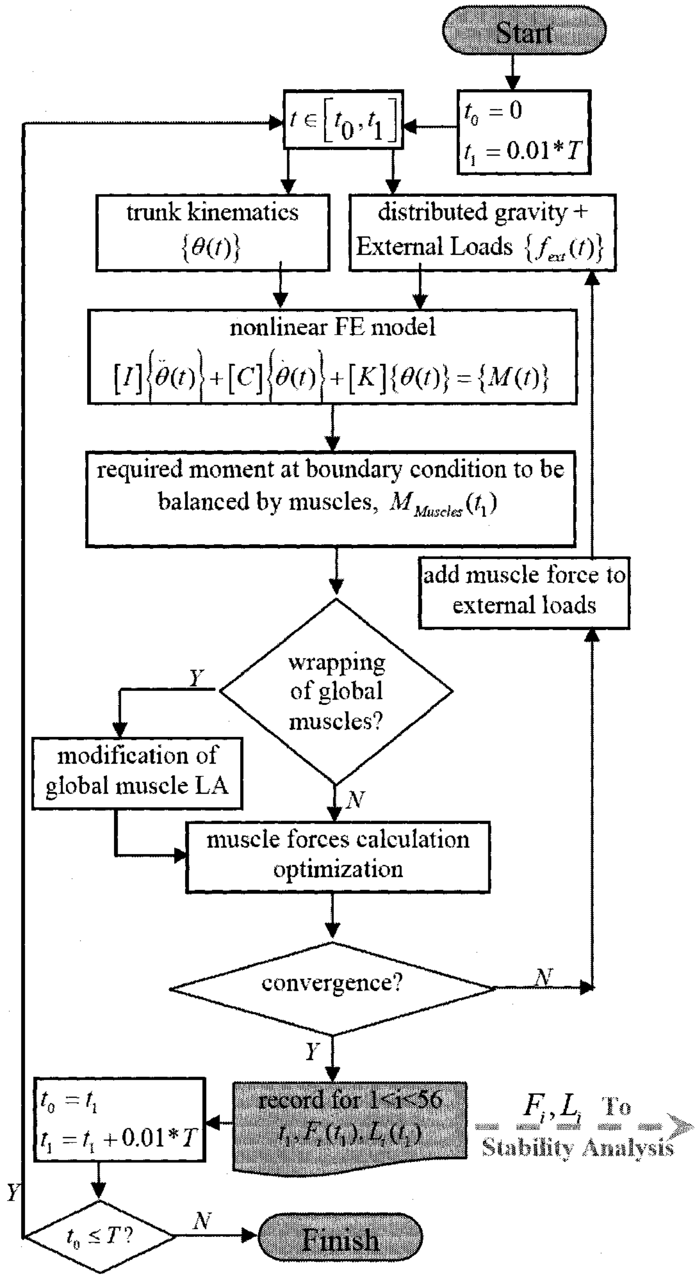

Figure A.2 Flowchart of the routine written in Python (Python 2.5) to perform iterative equilibrium analyses in Kinematics-based approach at each instance of time; L_i denotes the instantaneous muscle length, T denotes total time, and F_i is the total muscle force.

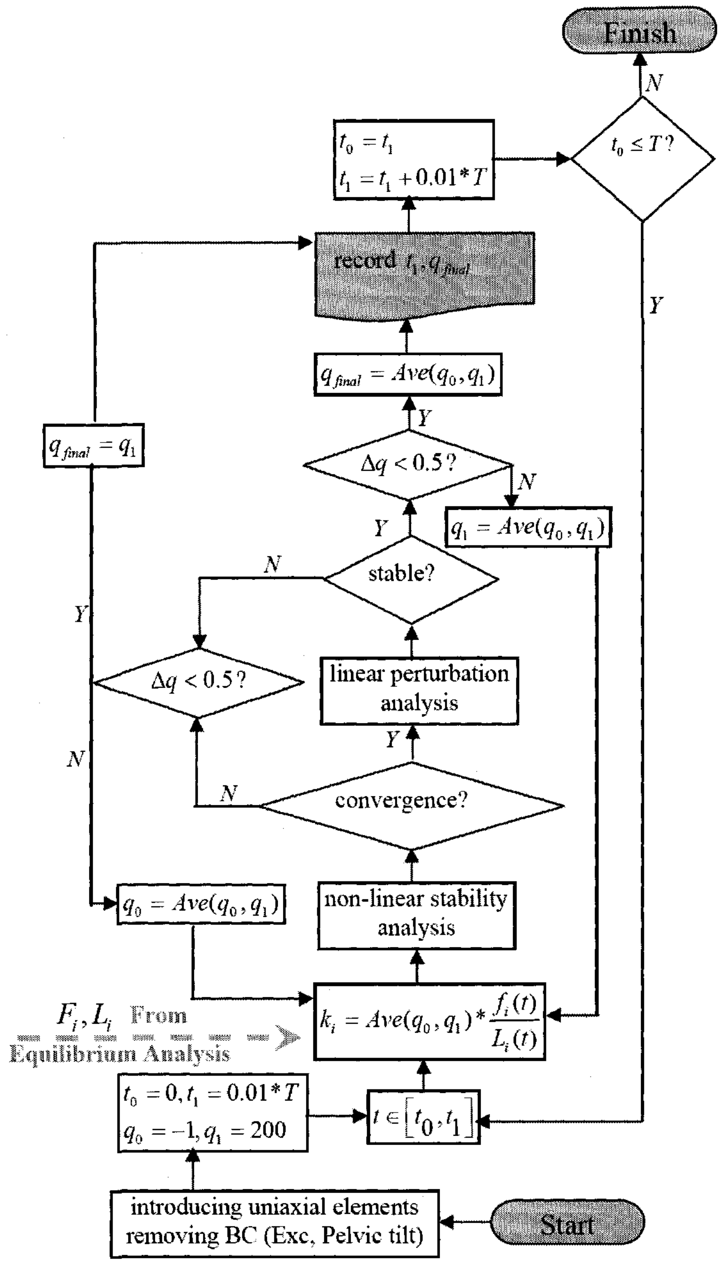

Figure A.3 Flowchart of the routine written in Python (Python 2.5) to perform iterative stability analyses in Kinematics-based approach at each instance of the time; L_i denotes the instantaneous muscle length, T denotes total time, F_i denotes the total muscle force, q_i is muscle stiffness coefficient, and k_i is muscle stiffness.

Once the muscle forces are calculated throughout the simulation period, the system stability is investigated by replacing muscles with uniaxial elements. The stiffness of each uniaxial element, k, is assigned using a commonly-used linear stiffness-force relation (i.e. k=q F/L) in which the muscle stiffness is proportional to the instantaneous muscle force, F, and inversely proportional to its current length, L, with q as a dimensionless muscle stiffness coefficient that is taken to be the same for all muscles (Bergmark 1989). At each instance of time, the stability margin for different q values is investigated at the loaded deformed configurations by natural frequency and linear perturbation analyses. The commercial Finite element package program ABAQUS (ABAQUS Inc. Version 6.5) is used to carryout nonlinear and linear stability analyses while the optimization procedure is solved analytically based on Lagrange Multiplier Method. Control and management of different parts of this iterative procedure at each instances of time (i.e. execution of FE analyses, extracting required results from output database, optimization procedure, calculation of spinal loads, application of muscle forces to model, perturbation and frequency analyses) are all performed by a computer script (Figs. A. 2, A. 3) in Python (Python 2.5).

A.2 Finite Element Model

As described in earlier chapters, the spine model used in the current study is a sagittally symmetric head-pelvis model made of six nonlinear deformable beams to represent T12-S1 segments and seven rigid elements to represent head-T12 (as a single body) and lumbosacral (L1-S1) vertebrae (Arjmand and Shirazi-Adl, 2005, 2006; Bazrgari et al., 2007; Bazrgari and Shirazi Adl, 2007). The geometry of the ligamentous spine model has been constructed (Table A.1) using CT scan images of a cadaver lumbar specimen and data in the literature (Shirazi-Adl and Parnianpour 1993, 1996). Lumbar posture in neutral standing position is based on segmental rotations given in El-Rich et al (2004) resulting in minimum sum of required equilibrating moments at lumbar L1-L5 levels and a deformed shape in agreement with measured lumbar lordosis in standing posture with no loads in hands (Table A.2).

Table A.1 Sagittal geometry of the spine in the sagittal plane (mm)

(X: anterior-posterior, **Z:** vertical)

Level	Initial Unloaded Geometry		Standing Geometry With Gravity Loads	
	X (mm)	*Z (mm)*	*X (mm)*	*Z (mm)*
T1	-12.6	467.6	37.0	463.4
T2	-17.7	447.4	28.4	444.4
T3	-22.1	426.9	20.6	424.9
T4	-25.1	405.8	14.0	404.7
T5	-26.8	384.1	8.6	383.6
T6	-27.6	361.7	4.0	361.7
T7	-27.6	338.4	0.0	338.7
T8	-26.7	314.1	-3.3	314.6
T9	-24.9	288.9	-5.8	289.5
T10	-21.7	262.9	-7.1	263.4
T11	-16.7	235.3	-6.9	235.3
T12	-9.6	204.6	-5.2	203.8
L1	-1.1	171.1	-1.4	169.9
L2	6.8	135.0	3.7	133.9
L3	13.0	97.6	9.2	96.9
L4	14.1	58.9	11.0	58.8
L5	10.2	20.6	8.8	20.8
S1	0.0	0.0	0.0	0.0

Table A.2 Prescribed total sagittal rotations (degree) at different lumbar levels for the neutral standing posture (+ve: backward extension)

Level	Standing
Head-T12	-9.9
L1	-6.2
L2	-2.5
L3	0.3
L4	1.8
L5	3.5
S1	5.0

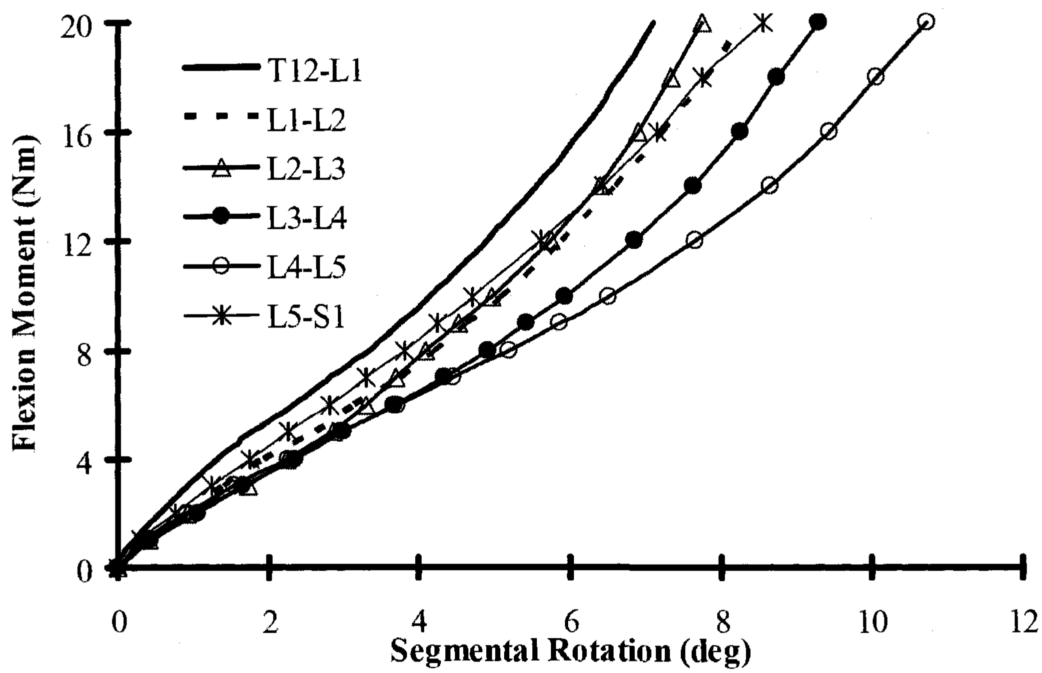

Figure A.4 Variation of flexion moment with segmental rotation for different lumbar motion segments under 1800 N axial compression.

Trunk motion segments are modeled using beam elements with nonlinear axial compression-strain and sagittal/lateral/axial moment-curvature relations (Figs. A.4, A.5). The nonlinear load-displacement response in different directions along with flexion versus extension differences are represented based on numerical and measured results of previous single- and multi-motion segment studies (Oxland et al., 1992; Pop, 2001; Shirazi-Adl et al., 2002; Yamamoto et al., 1989). The flexural rigidity of the model depends also on the axial compression as reported recently (Fig. 6.2) (Shirazi-Adl, 2006). Masses, mass moments of inertias, and corresponding mass centers at different trunk levels along the spine, given in Table A.3, are based on published data (de Leva, 1996; Pearsall et al., 1996; Zatsiorsky and Seluyanov, 1983). The inter-segmental dampings are assigned using measured values (Kasra et al., 1992; Markolf, 1970) where translational damping =1200 Ns/m and angular damping = 1.2 Nms/rad. Buttocks at the base is modeled by a connector element (compression only) with nonlinear stiffness defined based on reported data in the literature (Table A.4) (Aimedieu et al., 2003; Kitazaki and Griffin, 1997a) and by a damping similar to that of lumbar segments.

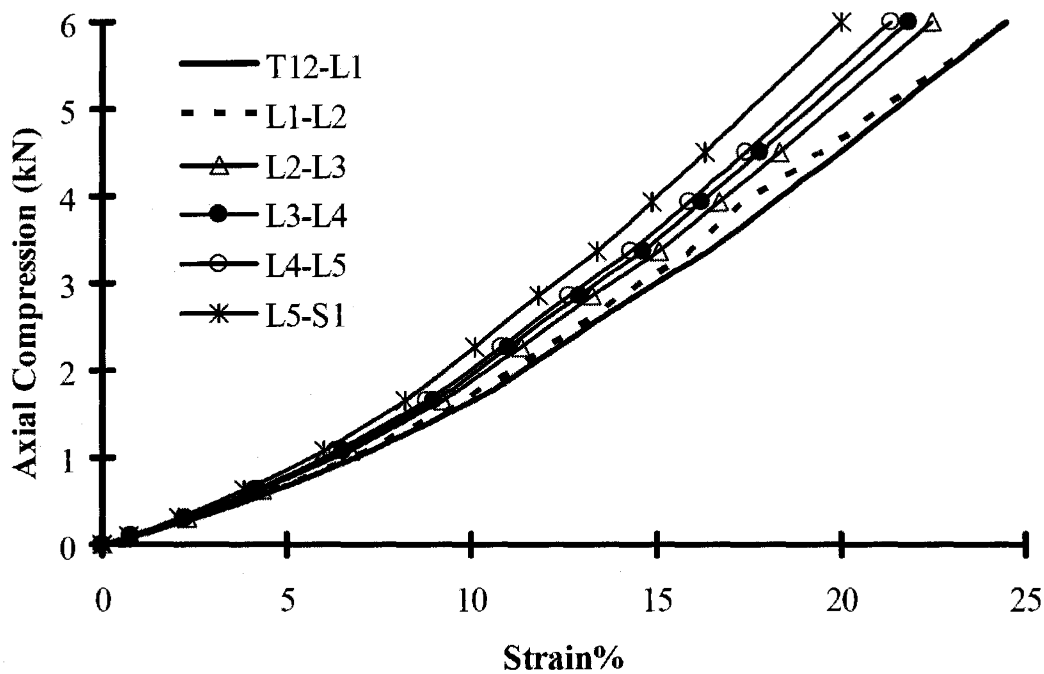

Figure A.5 Load-displacement properties for different lumbar motion segments.

A.3 Trunk Muscles

A sagittally symmetric muscle architecture with 46 local (attached to the lumbar vertebrae) and 10 global (attached to the thoracic cage) muscles is used (Fig. A.6). Anatomy of these muscles including insertion points, and physiological cross sectional area (PCSA) in the present study were taken mainly from the extensive study of Stokes and Gardner-Morse (1998) and have been given in Table A.5.

Table A.3 Trunk segments mass, mass moment of inertia and mass center locations

Level	% TM[*]	% BM	Ixx	Iyy	Izz	CG - z	CG – x
Head-Neck	--	6.94	27.18	29.34	20.13	597.60	-10.00
Upper Arms	--	2*2.8	12.63	11.30	3.80	447.38	30.00
Lower Arms	--	2*1.6	6.45	5.99	1.20	426.85	30.00
Hands	--	2*0.6	1.31	0.88	0.50	405.81	30.00
T1	3.59	1.28	6.70	2.00	8.70	467.60	-8.00
T2	3.88	1.38	3.40	2.40	9.10	447.38	-12.00
T3	4.15	1.47	8.40	3.20	11.50	426.85	-20.00
T4	4.46	1.58	8.30	3.40	11.70	405.81	-28.00
T5	4.72	1.68	8.00	3.50	11.50	384.14	-33.00
T6	5.03	1.78	7.80	3.90	11.60	361.70	-39.00
T7	5.29	1.88	7.40	4.10	11.50	338.40	-43.00
T8	5.60	1.99	7.20	4.40	11.60	314.12	-45.00
T9	5.91	2.10	7.20	4.70	11.80	288.94	-48.00
T10	6.17	2.19	8.90	6.20	15.00	262.94	-48.00
T11	6.47	2.30	9.00	6.20	15.20	235.30	-46.00
T12	6.74	2.39	11.00	7.20	18.10	204.56	-44.00
L1	7.04	2.50	11.10	6.50	17.50	171.07	-37.01
L2	7.30	2.59	10.90	6.00	16.80	135.03	-29.00
L3	7.61	2.70	10.70	5.50	16.10	97.55	-17.00
L4	7.87	2.79	11.20	5.30	16.40	58.90	-10.00
L5	8.19	2.91	12.20	5.60	17.70	20.57	-6.00
S1	0.00	0.00	0.00	0.00	0.00	0.00	0.00
Pelvis	--	11.00	75.00	30.00	80.00	-89.00	0.00

[*]TM: Trunk mass, BM: Body mass, *Ixx*, *Iyy*, *Izz*: Mass moments of inertia respectively in anterior-posterior, transverse and longitudinal directions ($Kg \, . \, m^2 * 10^{-3}$), CG-z: height of the centers of mass with respect to the S1 (*mm*), CG-x: anterior-posterior distance from corresponding vertebral centers with negative indicating anterior position (*mm*). Upper arms, lower arms, and hands centers of mass are considered posteriorly at T2, T3, and T4 vertebral levels, respectively.

Table A.4 Buttocks' nonlinear stiffness properties

Force (N)	Deflection (m)	Stiffness (kN/m)
0.0	1.0	0.0
0.0	0.0	0.0
-12.5	-0.001	12.5
-62.5	-0.002	50
-1232.5	-0.02	65
-7232.5	-0.05	200

Since the number of muscles at each level exceeds the number of equilibrium equations, the following optimization problem is solved to resolve the remaining redundancy at different spinal levels:

$$Minimize(\sum_{i=1}^{i=m}\left(\frac{F_i}{A_i}\right)^3) \qquad \text{(A.1)}$$

Subject to:

$$\sum_{i=1}^{i=m} r_i F_i - M = 0 \qquad \text{(A.2)}$$

$$F_p \leq F_i \leq F_p + 0.6\times10^6 \times A_i \qquad \text{(A.3)}$$

Where F_i denotes the unknown force in the muscle i (N), A_i is the physiological cross sectional area (m^2) of the i^{th} muscle, m is the total number of muscles attached to a vertebral level, r_i is the moment arm of the i^{th} muscle (m) and M is the resultant moment (Nm) in sagittal plane at the spinal level under consideration.

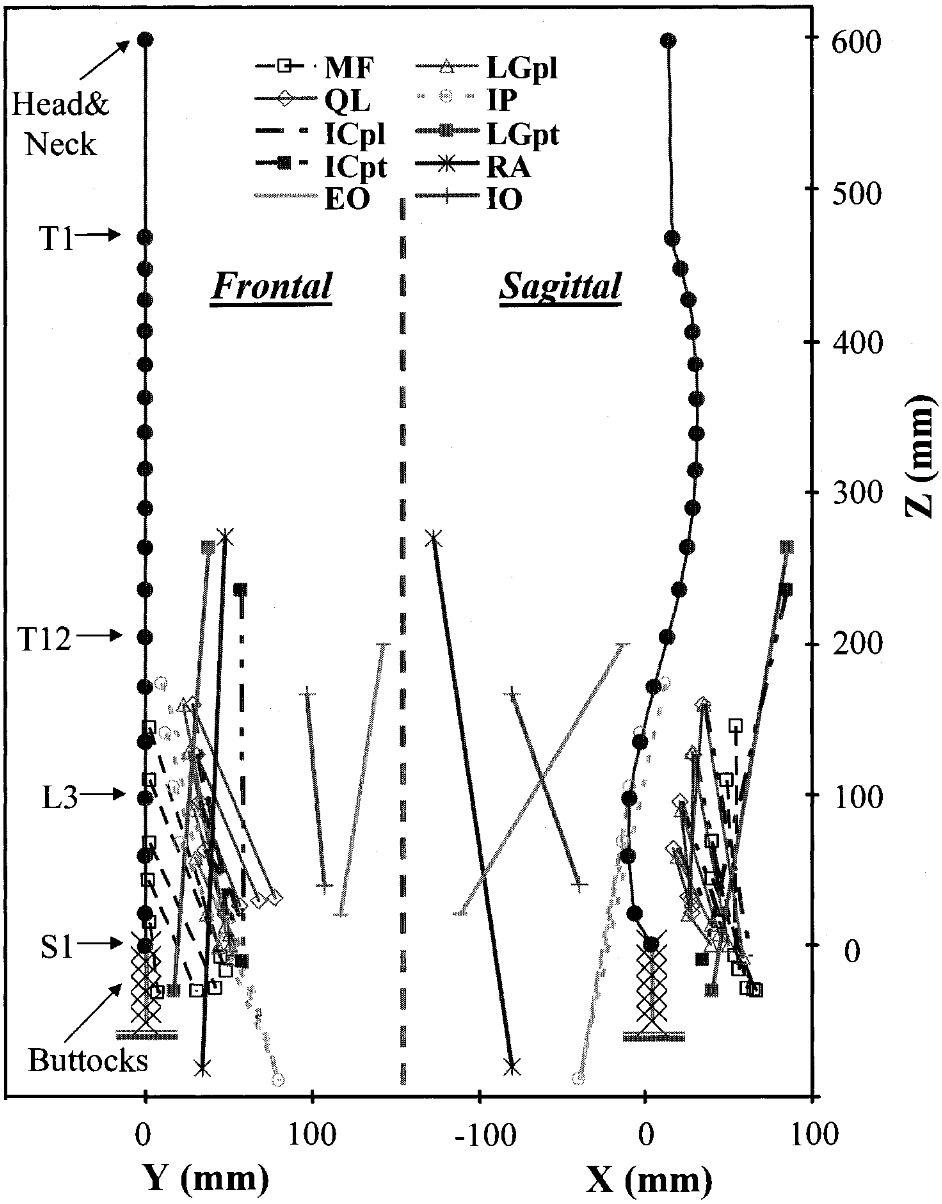

Figure A.6 Representation of the spine including global and local musculatures in the sagittal (on the right) and frontal (on the left, fascicles on one side are shown) planes. ICpl: Iliocostalis lumborum pars lumborum, ICpt: Iliocostalis lumborum pars thoracic, IP: Iliopsoas, LGpl: Longissimus thoracis pars lumborum, LGpt: Longissimus thoracis pars thoracic, MF: Multifidus, QL: Quadratus lumborum, IO: Internal oblique, EO: External oblique, and RA: Rectus abdominus.

Table A.5 Coordinates of origins and insertions of the trunk muscles (initial unloaded geometry) along with physiological cross sectional area for muscles on each side

Level	Muscle Name*	PCSA (mm^2)	Origin			Insertion		
			X(mm)	*Y(mm)*	*Z(mm)*	*X(mm)*	*Y(mm)*	*Z(mm)*
T12	***ICpt***	660	84.9	57	235.3	34.9	58	-10
	IO	1200	-80	96	167	-40	107	40
	RA	567	-126.7	47.3	269.8	-80	34	-80
	EO	1576	-13	141.5	200	-111	116	20
	LGpt	1345	85.6	37.8	262.9	40	17	-30
L1	***MF***	96	54.9	2.5	145	54	45.2	-7.4
	QL	88	34.8	28.3	159.3	26	77	32
	IP	252	12	10	174.3	-40	79	-88
	ICpl	108	35.7	28.3	159.3	63	52	-7
	LGpl	79	35	22.1	159.6	59	51.3	-8.4
L2	***MF***	138	48.8	2.5	110	56	47.9	-16.5
	QL	80	28.6	30.5	126.9	27	67	30
	ICpl	154	29.8	30.5	126.9	49	52	12
	LGpl	91	28.4	26	128.2	50	50.7	0.1
L3	***MF***	211	40.6	2.6	68.4	61	41.6	-28.3
	QL	75	21	32.3	95.3	28	56	26
	ICpl	182	21.9	32.3	95.3	44	55	18
	LGpl	103	22	28.9	90.1	44	49.6	7.3
L4	***MF***	186	40.6	1.5	43.4	65	30.4	-29.5
	QL	70	17.1	35.1	63.9	28	47	21
	ICpl	189	19.3	35.1	63.9	37	58	23
	LGpl	110	19.3	30	58.6	39	47.1	13.3
L5	***MF***	134	43.9	2.3	15.2	67	7.6	-30.4
	LGpl	116	25.7	36	20.6	39	42.9	0

* ICpl: Iliocostalis lumborum pars lumborum, ICpt: Iliocostalis lumborum pars thoracic, IP: Iliopsoas, LGpl: Longissimus thoracis pars lumborum, LGpt: Longissimus thoracis pars thoracic, MF: Multifidus, QL: Quadratus lumborum, IO: Internal oblique, EO: External oblique, and RA: Rectus abdominus. x, y, and z represent anteroposterior, transverse (lateral), and longitudinal directions respectively

To qualitatively compare predicted muscle forces with measured normalized EMG activity, the total muscle force in each muscle is partitioned into active and passive components with the latter force being evaluated base on a length-force (passive tension) relationship. In the current study, we defined muscles length-force relationship based on recent in vivo study of Davis et al (2003) (Fig A.7) whose data are in the range of the all reported values (Woittiez et al 1984, Deng and Goldsmith 1987, McCully and Faulkner 1983). A parabolic curve was fitted for use in our model study.

$$F_P / F_{Max} = 15.05(L/L_0)^2 - 30.238(L/L_0) + 15.194, R^2 = 0.9943 \quad \text{(A.4)}$$

in which L_0 is the optimal muscle length (m), L is the instantaneous muscle length, F_p is the passive muscle force (N), and $F_{M\,ax}$ is the maximum active muscle force (N).