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SYNERGY OF HUMAN SPINE IN NEUTRAL POSTURE

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SYNERGY OF HUMAN SPINE IN NEUTRAL POSTURE

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To my mother, in Christ

RÉSUMÉ

Les désordres de la colonne vertébrale humaine sont la source majeure des malaises des travailleurs et ont un impact socio-économique important. Dû aux changements dans le style de vie et l'environnement de travail, de nouvelles tendances de traumatismes du rachis apparaissent. Ces désordres sont souvent la conséquence de postures prolongées, en absence de chargements externes, plutôt que de la manutention de charges lourdes. La venue de causes cumulatives d'endommagement de la colonne vertébrale nécessite le développement d'approches nouvelles. L'approche proposée dans cette étude combine les deux parties de colonne vertébrale humaine: la colonne vertébrale ostéoligamentaire et les muscles rachidiens . Un algorithme de contrôle décrit l'interaction entre ces deux parties. L'analyse du rachis ostéoligamentaire est effectuée par un module passif, basé sur le logiciel d'analyse structurale ABAQUS. Le module actif analysant les activations des muscles est construit à l'aide d'un assemblage de routines de FORTRAN et de C qui exécutent la transformation géométrique et l'optimisation.

Dans le modèle proposé pour la posture neutre, la rigidité des segments vertébraux reste linéaire. Par conséquent, le comportement des disques est simulé par des poutres. Le tronc, renforcé par la cage thoracique est considérablement plus rigide que le rachis lombaire, est simulé par une augmentation de la rigidité des disques thoraciques. Bien que les rotations inter-segmentales soient petites, l'effet cumulatif du déplacement de tous les disques ne peut

pas être négligé et l'analyse en grands déplacements est nécessaire pour quantifier les effets d'instabilité. Dans l'analyse d'équilibre, les ressorts virtuels sont utilisés pour contrôler la configuration vertébrale et l'interaction entre la colonne vertébrale passive et les muscles. Par conséquent, le potentiel des composantes passives et actives de la colonne pour résister au chargement est utilisé de façon efficace. Dans l'analyse, les muscles sont classifiés en deux groupes: les globaux qui lient le bassin pelvien à la cage thoracique et les locaux qui lient le bassin aux vertèbres lombaires. Les lignes d'action des muscles sont considérées rectilignes, et leurs points d'insertion suivent les vertèbres dans un mouvement de corps rigide. La redondance du problème musculaire est résolue par une minimisation du chargement compressif produit par les activations musculaires.

Les activations ainsi déterminées peuvent être utilisées pour estimer la stabilité mécanique de la colonne. Une formule simple pour la rigidité musculaire attribue une rigidité à chacun des muscles suivant sa force et sa longueur (obtenues dans l'analyse d'équilibre). Les propriétés correspondantes à la rigidité requise sont attribuées aux éléments barres utilisés dans cette étape, et les forces de tension trouvées dans analyse d'équilibre sont induites par des changements de température. Donc, on obtient la même distribution du chargement, des forces internes et de la configuration géométrique que dans le cas de l'analyse d'équilibre avec les ressorts virtuels. Cet état de la colonne est analysé par une perturbation linéaire pour une amplitude du chargement additionnel qui peut causer l'instabilité.

L'approche proposée examine le rôle du changement de configuration de la colonne contrôlée par des mécanismes tels que le positionnement de T1 qui affecte la mobilisation de la résistance passive. La rotation pelvienne et les forces musculaires sont ensuite analysées par rapport à la viscoélasticité des disques. La réponse de la colonne au chargement physiologique est aussi évaluée pour les cas d'architecture musculaire simple et détaillée.

Les résultats d'un modèle avec une architecture musculaire simple indiquent que les amplitudes des forces musculaires sont reliées à la configuration de la colonne et à la distribution du chargement contrôlé par le positionnement de T1 et la rotation pelvienne. La rotation pelvienne en conjonction avec le positionnement de T1 est efficace dans le contrôle de lordose lombaire, ainsi que dans l'activation de la résistance passive et la stabilisation de la colonne vertébrale. En présence d'un comportement viscoélastique des disques, la rotation pelvienne et les forces musculaires changent avec le temps. L'amélioration de l'architecture musculaire a pour conséquence une distribution des activations réalistes. En général, les activations musculaires avec leurs effets sur la colonne passive se rapprochent du cas d'architecture simple. Les résultats indiquent que les muscles attachés à la cage thoracique sont efficaces dans le contrôle de la configuration de la colonne et de la distribution du chargement. Les muscles attachés sur les vertèbres lombaires jouent un rôle majeur dans la stabilisation de la colonne.

Les observations démontrent que le potentiel de la colonne vertébrale à supporter des chargements est basé sur des principes de synergie. La résistance en compression de la colonne vertébrale, est exploitée par un maintien des modes supérieurs de flambement. Les forces nécessaires pour maintenir ces modes supérieurs, pendant son application sur les points caractéristiques, sont minimales. Une des fonctions de la multiplicité d'architecture musculaire lombaire dans la posture neutre est de protéger la colonne d'une transition en hypermobilité quand elle se déforme selon les modes supérieurs de flambement.

ABSTRACT

Disorders of the human spine account for a major part of disabilities in the workforce and have a significant socioeconomic impact. Due to the changes in the lifestyle and the working environment, new trends of spinal disorders have emerged. Often, these disorders result from long sustained postures under the body weight rather than from manipulating heavy loads. The present study makes an attempt to develop a new hybrid kinematic approach for analysis of the human spine in the neutral posture. The model combines two essential parts of the human spine: the passive osteoligamentious spine and spinal muscles with both parts interacting according to a proposed control algorithm. The passive module for analysis of the osteoligamentous spine utilizes ABAQUS structural analysis software. The active module analyses activations in the muscles and is constructed as an assembly of FORTRAN and C routines performing geometric transformation and optimization.

Since the model focuses primarily on neutral posture with a limited range of motion, stiffness of the spinal motion segments is considered to be linear, and beams are used to simulate their elastic behavior. The trunk, reinforced by the rib cage and significantly more rigid than lumbar spine, is simulated by increasing the stiffness of the thoracic disks. Although the intersegmental rotations are small, the cumulative effect of displacements at all disks levels on the load configuration cannot be neglected and large displacement analysis is necessary to quantify the instability effects. In the equilibrium spinal analysis virtual springs are used to control the spinal configuration and to provide for interaction between

the passive spine and the muscles, thereby to utilize load resisting potential of passive and active components in an efficient manner. In the analyses, muscles are classified in two groups, global running from the pelvis to the rib cage and local running from the pelvis to the lumbar vertebrae. The lines of muscle action are considered to be straight, and their insertion points follow the vertebrae in a rigid body motion. Transfer of forces between the passive spine and the muscles is controlled by five constraints at each vertebra. The redundancy of the muscle problem is resolved by minimizing compressive load resulting from muscle activations. Thus determined muscle activations can be subsequently used to assess the mechanical stability of the spine. A simple formula for muscle stiffness is used to assign stiffness to each muscle based on its force and length determined from the equilibrium analysis. Constraints are replaced by the truss elements originating and inserting into the same points as muscles with corresponding stiffness properties and tension forces. Thus, the same distribution of load and forces and geometric configuration of spine is obtained as in the case of equilibrium analysis with the virtual springs. This state of spine is subsequently analyzed in a linear perturbation step for a magnitude of an eigenvalue which corresponds to the load added onto the arms.

The proposed approach examines effect of changes in spinal configuration controlled by mechanisms such as T1 positioning on mobilization of the passive spinal resistance. The role of the pelvic rotation and muscle forces are then further investigated during the time dependent viscoelastic changes (creep) of the disks. Response of the spine under the postural load is evaluated also in cases with a simple and a detailed muscular architecture.

Results from the model with a simple muscle architecture indicate that magnitudes of muscle forces are strongly dependent on the spinal configuration and load distribution controlled by the T1 position and pelvic rotation. Pelvic rotation in conjunction with the T1 positioning are found to be efficient in controlling the physiological load distribution and activation of the passive spinal resistance thereby resulting in stabilization of the spine. When considering creep of the disks, values of pelvic rotation necessary to stabilize spine vary with time. Improvements in muscle architecture result in a more realistic pattern of muscle activations, at the same time, overall muscle activations with their effect on the passive spine remain close to the case with a simple muscle architecture. Muscles originating from the pelvis and inserting onto the rib cage are found to be efficient in controlling the spinal configuration and the load distribution. Muscles running from the pelvis and inserting into the lumbar vertebrae contribute primarily to the stability of spine.

Findings indicate that the spinal load-bearing potential is enhanced by the mechanisms based on synergetic principles. The compressive resistance of the spinal structure considered as a vertical column is exploited by maintaining the higher buckling modes. Muscle forces necessary to maintain higher modal shapes when acting at characteristic buckling points are only minimal. One of the functions of multiplicity of the lumbar muscular architecture in the neutral posture is to prevent spine from transition into hypermobily when the spine deforms in the higher buckling modes.

CONDENSÉ

La posture neutre et droite dans le champ gravitationnel terrestre est un état naturel de la colonne vertébrale maintenu pendant de longues périodes de temp, pendant les activités journalières de la vie humaine. Le maintien propre de cette posture est essentiel pour le bien-être d'un individu. La colonne vertébrale est un organe très spécial qui soutient le corps humain comme une structure biomécanique et sert en même temps pour le transfert de chargement et la locomotion, i.e., combinant la rigidité et la flexibilité. Le rachis assume sa fonction de transfert de chargement graduellement pendant les premières années de la vie humaine, contrairement aux autres structures biomécaniques du corps. La forme triplement courbée de rachis (lordose cervicale, cyphose thoracique, lordose lombaire) et sa rigidité axiale se développent en réponse à la posture droite marchante à partir de la forme cambrée propre à un enfant nouveau-née. La résistance rachidienne aux chargements axiaux origine de deux sources: le rachis passif (les vertèbres avec les disques et les ligaments de support), et le rachis actif (les muscles agissant comme les actuateurs et générateurs de force). Les ajustements de la colonne d'un enfant au chargement de gravité dans une posture droite représente une période d'entraînement neuromusculaire et de développement de la forme de la colonne passive. La physiologie et le développement adapté du système vertébral pour la posture droite dans le champ terrestre, suggèrent un fonctionnement synergétique: "*Le comportement de système complet est imprévisible par le comportement de ses parties considérées séparément*". Le rachis humain est une structure synergétique adaptée pour faire face à l'environnement terrestre. Le développement de la capacité de la colonne à résister aux

chargements est impensable en l'absence d'un champ gravitationnel approprié. Ceci se presénte sur les stations orbitales ou chez une personne confinée au lit depuis l'enfance et peut aussi affecter les autres fonctions rachidiennes. En plus des changements de longueur immédiats dus au chargement appliqué, la colonne subit aussi des changements avec le temps dûs au fluage de ses composantes passives molles (principalement des disques). Dans un cycle circadien, pendant la période des activités journalières, le rachis perd une partie de sa longueur, due au fluage des disques. Ce raccourcissement est récupéré durant la période de sommeil, lorsque les charges axiales sur la colonne sont minimales. Bien que le phénomène de fluage des disques affecte l'équilibre de la colonne, il est nécessaire pour le transport des substances biologiques (les nutriments et les produits de métabolisme) à l'intérieur et au dehors des disques.

La colonne vertébrale, dans une posture droite, doit fournir une résistance aux chargements et en même temps une stabilité suffisante. La résistance en compression est définie par les conditions d'équilibre. Néanmoins, la définition de la stabilité est sujette à interprétation dans la communauté biomécanique. Pour les fins de l'étude présentée, la stabilité est considérée celle définie pour toutes structures mécaniques: la structure est stable lorsque' elle est chargée avant sa transition dans l'hypermobilité. La théorie linéaire et nonlinéaire de stabilité peut être utilisée pour analyser la stabilité structurale. La première décrit la structure dans des termes de chargements critiques seulement, la dernière fournit l'histoire du chargement et du déplacement.

Dénudé de musculature, le rachis thoracolombaire humain ne peut résister qu'à des charges axiales minimales. De plus, les mesures électromyographiques effectuées sur les individus en posture droite et en chargements modérés indiquent que les activités dans les muscles rachidiens sont faibles. Puisque ni les muscles ni la colonne passive ne possèdent aucune résistance au chargement substantielle, quelques autres mécanismes doivent être à la base du comportement rachidien. Plusieurs mécanismes ont été proposés dans la littérature, e.g., rotation pelvienne, pression abdominale ou tension du fascia thoracolombaire.

La colonne est essentielle au positionnement de la tête qui loge les principaux organes nerveux, proprioceptifs et sensoriels. Les expériences quantifiant l'exactitude du contrôle proprioceptif de la posture droite indiquent que la personne peut centrer sa tête avec une précision de quelques millimètres dans la direction sagittale et latérale. Cette observation donne un paramètre plausible pour le contrôle de la posture humaine. Néanmoins, ce mécanisme paraît perdre son efficacité en l'absence du champ gravitationnel terrestre, due à sa liaison avec l'appareil vestibulaire.

Le présent travail développe une approche nouvelle, hybride, pour l'analyse de la colonne vertébrale en intégrant un programme commercial d'analyse structurale générale (ABAQUS) avec quelques modules d'optimisation et de transformation géométrique développée *"in-house"*. La performance de cette approche est évaluée premièrement sur un modèle avec une architecture musculaire conceptuelle dans les conditions statiques, et par la suite dans des conditions statiques avec des effets viscoélastiques. Une fois les avantages et les limitations de cette approche déterminés, elle est appliquée sur un modèle avec une architecture musculaire plus réaliste dans des conditions statiques. La colonne passive utilisée dans tous les modèles est basée sur la même géométrie, correspondant à une personne de sexe féminin ayant une masse corporelle de 68kg et une taille de 170cm. Les constantes élastiques pour les poutres représentants les disques intervertébraux ont été trouvées dans la littérature comme étant E=7MPa et G=3MPa. La partie thoracique, renforcée par sa cage est considérée rigide. Les propriétés mécaniques de la colonne passive, incluant les caractéristiques du fluage pour les disques, sont basées sur les données trouvées dans la littérature. Pour quantifier les effets d'instabilité une analyse à grands déplacements a été exécutée.

Les muscles, composantes actives de la colonne, sont divisés en deux groupes, locaux et globaux, suivant leurs points d'attache, soit à la colonne thoracique renforcée par sa cage, ou à la colonne lombaire flexible. Les forces au niveau des disques lombaires, résultant du chargement physiologique gravitationnel, sont supportées par la résistance passive des segments vertébraux et des muscles. La configuration de la colonne est importante pour le contrôle de la répartition de la résistance entre les composantes rachidiennes passives et actives. Dans cette étude, la configuration de la colonne vertébrale est contrôlée par des contraintes (ressorts virtuels) qui appliquent les forces nécessaires à la stabilisation sur la colonne passive. Ces forces de stabilisation étant exercées par les muscles, elles sont accompagnées par la pénalité de compression. Cette approche se propose de déterminer les forces musculaires, et la configuration déformée de la colonne passive. De plus, il est nécessaire de résoudre la redondance du problème musculaire et de trouver les valeurs appropriées des contraintes qui contrôlent la configuration vertébrale et la répartition de la résistance rachidienne entre ses composantes passives et actives.

Chaque disque fournit cinq conditions d'équilibre statique, aucune contrainte étant imposée dans la direction du mouvement axial. Avec un nombre de muscles attachés à chaque vertèbre excédant le nombre de conditions d'équilibre, deux procédures peuvent être employées pour résoudre le problème musculaire. La première approche utilise un groupement des muscles avec des fonctions similaires ou une prescription des relations entre leurs activations, réduisant ainsi le nombre de conditions d'équilibre requises. La deuxième approche utilise des schémas d'optimisation pour sélectionner les muscles les plus efficaces, afin de résister aux forces de chargement à chaque vertèbre lombaire, leur nombre étant égal au nombre de conditions d'équilibre.

Dans la solution de ce problème musculaire, la force de compression résultant d'activité musculaire a été minimisée à chaque vertèbre lombaire. Dans les deux cas considérés, les changements de directions des muscles dus au changement de configuration de la colonne ont été considérés.

La sélection des valeurs de rigidité pour les ressorts virtuels est liée à la stabilité générale du rachis. L'augmentation de ces rigidités contribue à la stabilité de la colonne, accompagnée par l'augmentation de la pénalité de compression. Par conséquent, la diminution de ces rigidités conduit à une participation plus importante de la colonne passive dans la génération de résistance au chargement axial, avec une stabilité et pénalité de compression modérée. La demande de stabilité de la colonne est liée à la sévérité du dommage potentiel, par conséquent, on peut pensesr que la marge de sécurité augmente avec le chargement externe. L'équilibre neutre comprend seulement le chargement minimal soutenu par la colonne dans la posture droite, dans le champ gravitationnel terrestre. Ainsi, les rigidités des ressorts virtuels dans l'équilibre neutre peuvent être basses, mais dans les limites de stabilité suffisante. L'autre façon d'estimer les valeurs des rigidités des ressorts est de procéder à une étude paramétrique. L'essai des différentes rigidités pour les ressorts permet de déterminer l'effet de leur variation sur les résultats de l'analyse

Une approche cinématique hybride est appliquée pour examiner le rachis humain dans la posture neutre statique et dans les conditions suivantes:

Modèle 1 réponse rachidienne passive+représentation musculaire conceptuelle,

Modèle 2 réponse rachidienne passive avec fluage+représentation musculaire conceptuelle,

Modèle 3 réponse rachidienne passive+représentation musculaire réaliste.

La différence entre le premier et le troisième modèle réside dans la complexité d'architecture musculaire, ainsi que dans la solution du problème musculaire. Dans le deuxième modèle, on applique la méthode développée dans le premier modèle pour un comportement viscoélastique (fluage des disques). Les lignes d'action pour l'architecture musculaire conceptuelle (quatre directions pour chaque point d'insertion) sont sélectionnées dans les confins du tronc. Pour ce cas particulier, les ressorts virtuels contrôlent les mouvements des points d'insertion dans le plan horizontal et donnent seulement deux conditions d'équilibre pour solutionner le problème musculaire avec quatre muscles. Le critère de sélection des deux directions pour les muscles actifs, parmi les quatre, est basé sur la direction de la force horizontale dans les ressorts virtuels en réduisant le problème musculaire statique. La procédure de solution est répétée de façon itérative, avec la pénalité de compression recalculée à chaque étape, jusqu'à ce que l'équilibre du système complet soit atteint.

L'architecture musculaire pour le troisième modèle est construite en liant les points d'insertion, situés sur les vertèbres, avec les points d'origine correspondant sur le bassin pelvien. Une vertèbre standard, positionnée dans l'axe de la colonne, est utilisée pour identifier les coordonnées des points d'insertion des muscles. Le bassin pelvien, ajusté proportionnellement à la taille de la colonne, est utilisé pour identifier les coordonnées des points d'origine. Le mouvement de chaque vertèbre est contrôlé dans les cinq directions (deux déplacements horizontaux et trois rotations), ce qui donne cinq équations d'équilibre pour solutionner le problème musculaire à chaque vertèbre. Néanmoins, le nombre de muscles attachés à chaque vertèbre est au moins égal à dix, surpassant le nombre de conditions d'équilibre. Une méthode d'optimisation linéaire, ayant une compression due aux activations dans les muscles minimisée à chaque vertèbre lombaire est utilisée pour déterminer les forces musculaires. La comparaison des activations musculaires des modèles un et trois montre leur accord général. L'avantage d'une musculature détaillée réside principalement dans une meilleure résolution des activations musculaires ainsi que dans un meilleur contrôle de stabilité. De plus, le mécanisme d'interaction entre la colonne passive et les muscles est similaire.

Le deuxième modèle est l'extension du premier dans le domaine viscoélastique. Le fluage des disques a un effet significatif sur la configuration rachidienne. Le déplacement vertical de T1 peut atteindre jusqu'au 25mm après une heure d'application du chargement physiologique, en utilisant uniquement les muscles globaux. Ce déplacement est réduit de

50% avec l'addition de muscles locaux. Cette réduction importante de la déflexion de T1 est obtenue par la diminution des moments sagittaux, ainsi que par le fluage en rotation des disques. Bien que cela ait pour conséquences quelques augmentations de la pénalité de compression, ces effets sont surmontés par la diminution des moments fléchissant.

La rotation pelvienne peut balancer les chargements agissant sur le rachis et varie avec l'amplitude du chargement appliqué et le temps. Les variations de la rotation pelvienne dues au fluage des disques surmontent celles en l'absence de fluage durant l'application graduelle du chargement physiologique. Le positionnement de T1 est un des autres paramètres qui affectent la configuration rachidienne avec le fluage des disques. Dans les positions ou le bassin est fixe pour la rotation sagittale (ou T1 ne peut pas bouger librement) une variation dans les forces musculaires est nécessaire pour maintenir l'équilibre rachidien sur le chargement physiologique.

Dans le troisième modèle, les activations des muscles globaux sont contrôlées par les cinq contraintes (trois rotations, k_{mx}, k_{my}, k_{mx}, et translations horizontaux $k_h = k_{fx} = k_{fy}$) imposées sur le mouvement de la cage thoracique lié avec la vertèbre T12. Pendant les changements de posture avec petits mouvements sagittaux (±40mm à partir de la position initiale de T1) les amplitudes des forces musculaires dépendent essentiellement de la contrainte horizontale à la T12, adjacente à la jonction thoracolombaire et on observe peu de sensibilité aux contraintes rotationeles. La valeur de la contrainte horizontale k_{fx}=5Nmm^{-1}, est aussi la valeur minimale pour ce paramètre, en dessous de laquelle le problème musculaire ne peut pas être résolu. Ceci indique, qu'une force horizontale, execée par les muscles, en plus de moments est toujours nécessaire pour le maintien de l'équilibre rachidien. Avec la valeur la plus élevée pour la contrainte horizontale, k_{fx}=100Nmm^{-1} qui permet peu de déplacement (contrainte presque rigide), la réponse délimitée par les valeurs de contrainte k_h utilisées représente aussi les limites du comportement pour le modèle (du rachis passif et des muscles) en présence des muscles globaux. Les variations des activités musculaires de système global suivent pendant le mouvement sagittal de la T1 des chemins "*close-to-linear*" à l'exception d'une région de transition proche de la position initiale de la T1. L'Iliocostalis thoracic et l'Oblique interne sur le côté gauche et droit sont actives dans les positions antérieures de T1. Le Rectus abdominis avec les Obliques externes sont nécessaires pour les positions postérieures de la T1.

L'activation des muscles locaux dans les positions antérieures a montré peu d'efficacité dans l'actuation du rachis, le retour du déplacement vers la position initiale de T1est accompagné par une pénalité de compression disproportionnelle. L'efficacité des muscles locaux diminue aussi avec l'augmentation des contraintes horizontales en T12. Avec la valeur de contrainte en T12 la plus élevée, le déplacement de la T1 due à l'activation des muscles locaux se renverse et devient antérieur.

Les trois modèles indiquent un rôle significatif du système passif dans la résistance aux chargements axiaux sur le rachis. Les segments vertébraux ont un bon potentiel pour supporter les forces de cisaillement et quelques moments fléchissant. Dans la réponse du rachis complet (considéré comme un assemblage des segments) sans la contribution des muscles, seulement une fraction de la résistance en flexion des segments peut être exploitée. Avec l'aide des composantes horizontales et des moments générés par les forces musculaires (leur compression étant un facteur déstabilisant comparable à celui du chargement axial), ou par la rotation pelvienne, les effets du chargement physiologique sur le système passif peuvent être contrôlés. Ainsi la résistance des segments peut être utilisée de façon plus économique. Dans les positions de T1 proches de sa position initiale, des moments fléchissants, allant jusqu'à 6000Nmm sont supportés par le rachis lombaire, et peuvent atteindre jusqu'à 8000 Nmm avec le fluage des disques. Le mouvement antérieur de T1 de moins de 40mm peut augmenter ces moments jusqu'à 12000Nmm.

L'estimation de la stabilité des positions d'équilibre est déterminée par la solution d'un problème aux valeurs propres en utilisant un modèle simple pour les muscles, décrit par un seul paramètre soit le coefficient de la rigidité musculaire, "q". La stabilité de la colonne est très peu affectée par le niveau d'activation des muscles globaux. L'augmentation d'activation des muscles locaux affecte significativement la stabilité de la colonne. À chaque niveau d'activation des muscles locaux, l'amplitude du chargement additionnel sur les bras, avant la perte de stabilité, dépend linéairement de la valeur du coefficient "q". La valeur minimale requise de "q" augmente linéairement avec l'augmentation d'activation des muscles locaux à partir de 8.5 jusqu'à 10.5. Le chargement additionnel le plus bas, avec les muscles globaux activés seulement et avec le coefficient "q" à sa valeur minimale 8.5, est 58N. Celui-ci est environ égal à 15% du chargement physiologique gravitationnel (380N) ou à 44% de la pénalité de compression (129N) dans cette configuration équilibrée particulière. Pour l'activation maximale des muscles locaux considérés, avec le coefficient "q" minimal à ce niveau (10.5), le chargement additionnel est de 80N (21% du chargement physiologique et 58% de la pénalité de compression). Avec des niveaux élevés d'activation des muscles locaux, la proportion du chargement additionnel et de sa pénalité de compression augmente: l'effet déstabilisant des muscles surmonte celui de la pénalité de compression adverse. La stabilité du rachis, ainsi quantifiée, prouve que les états d'équilibre obtenus par la méthode proposée peuvent résister à des charges sur les bras allants de 80N jusqu'à 120N ajoutées sur les bras. Néanmoins, une valeur critique existe pour chaque niveau d'activation des muscles locaux, valeur au-dessous de laquelle les configurations du rachis ne sont pas stables.

Chaque analyse de la colonne vertébrale doit satisfaire les conditions nécessaires (d'équilibre) et suffisantes (de stabilité) pour le maintien d'une posture propre. L'interaction entre le rachis passif et les muscles, basée sur l'équilibre (conditions cinématiques), peut donner au rachis une capacité en compression et une stabilité suffisante dans la posture neutre. La méthode proposée utilise les conditions d'équilibre du rachis pour déterminer les activations dans les muscles et le comportement élastique du rachis passif. Ensuite, elle quantifie la stabilité mécanique des configurations ainsi déterminées. Un modèle simple, décrit par le coefficient d'activation musculaire "q" est utilisé pour simuler la rigidité musculaire. La valeur minimale du coefficient "q" nécessaire pour le maintien de la stabilité, est environ de 9.

Les diagrammes de distribution des moments fléchissants sagittaux pour les composantes passives et actives du rachis indiquent une dépendance de lordose sur les conditions de chargement. La courbure lordotique est reliée à la position de la jonction thoracolombaire, un point important pour le contrôle des forces exercées sur le rachis lombaire par la partie thoracique. De façon similaire, la courbure lordotique est un indicateur important de la mobilisation de la résistance élastique des disques intervertébraux et des ligaments rachidiens au chargement appliqué. Lorsque la posture droite est sujette à la gravité, la rotation pelvienne est un contrôleur efficace de la courbure lordotique.

La méthode proposée crée une base pour l'implantation de différents critères d'activation des muscles et pour la modélisation du comportement des tissus passifs. La relation obtenue pour les forces musculaires, et leurs longueurs, indique que les celles-ci sont liées aux changements de configuration rachidienne, dans une transformation de coordonnés, avec un changement de valeur de référence pour chacun des muscles.

Bien que les considérations de fonctionnement des muscles et la modélisation des tissus passifs soient importantes dans la méthode proposée, la prédiction majeure est celui de la synérgie structurale du rachis: Le contrôle sélectif des muscles peut maintenir la colonne dans les modes du flambement plus élevés, ce qui donne par conséquent une posture efficace au niveau de l'effort musculaire. La complexité de la musculature rachidienne suggère son bon potentiel pour le contrôle des transitions entre les modes de flambement.

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Forces:

$\{F^c\}$	constraint forces in virtual springs
$\{F^l\}$	forces resulting from the external load acting at the disks
$\{F^m\}$	forces resulting from muscle activations acting at the disks
$\{F^p\}$	forces generated by the passive tissues acting at the disks
$\{\Delta P\}$	compressive penalty resulting from muscle activations acting at the disks
$\{F\}$	forces resisted by the passive and the active spinal components
$\{S\}$	forces in muscle fascicles
$\{P_i^{crit}\}$	buckling load corresponding to the "i-th" mode
R_h, R_v	horizontal, vertical reaction

Stiffnesses:

$[K^p]$	stiffness matrix for the passive components
$[K^c]$	stiffness matrix for the virtual springs
$[K]$	stiffness matrix including passive and active spinal components

Geometry:

$[A]$	matrix describing the muscle geometry
$\{B\}$	vector of penalty coefficients
n_i	muscle normal for the "i-th" muscle
α_{ij}	angle between the i^{th} global axis and the j^{th} muscle
r_i	muscle lever arm at the disk level around the i^{th} global axis
l_i	muscle length
h_i	disk height
φ_i	i^{th} mode shape

Mechanical properties:

A	disk area
I	disk modulus of gyration
J	disk torsion modulus
E	Young's modulus
G	Shear modulus
ν	Poisson ratio
k	muscle stiffness

Geometry - points:

T1-T12	Thoracic vertebrae
L1-L5	Lumbar vertebrae
S1	Sacrum
O_i	muscle origins
I_i	muscle insertions

CHAPTER 1

INTRODUCTION

1.1 MUSCULOSKELETAL DISORDERS

Musculoskeletal disorders, MSD, constitute a major portion of disabilities in the contemporary North American population. The causes of MSD can be classified in the following categories:

i acute or cumulative trauma (damage to bone, ligaments, muscles or nerves),

ii infections, i.e., acute or chronic disease (viral or bacterial infection),

iii natural ageing factors (desiccation of cartilaginous tissues, loss of muscle tissue, loss of bone minerals, neural degeneration),

iv congenital defects (genetic or acquired),

v idiopathic disorders.

From these five causes, the first two are preventable. The other causes cannot be directly eliminated, nevertheless, surgical and/or pharmaceutical treatment together with the appropriate physiotherapy and supportive devices can partially compensate for the defects and improve the quality of life of the afflicted individual. There can be a significant overlap between the mentioned factors for causes where one or more precursors might lead to pronounced symptoms such as decrease in mobility, constriction of the neural ducts or joint instability.

The MSD phenomenon becomes manifest through a defective behavior in one or both of its physiological constituents, muscles or skeleto-cartilaginous components. Neuro-muscular disorders can be caused either by: a neural condition affecting the central and peripheral nervous system, such as Cerebral palsy, Polio, Parkinson's disease, or a muscle degeneration such as Muscular dystrophy. The skeleto-cartilaginous disorders affect the osseous and cartilaginous tissues, such as: osteoporosis, osteopaenia, degeneration of the articular cartilage (osteoarthritis), or gradual loss of water in the cartilage (Schauff et al., 1990).

Certain environmental MSD can be prevented by elimination of their causative factors. This, in the majority of cases means protection against injury or infection. The protection against the environmental factors is double fold: protective and prophylactic. Various legally binding codes specify the environmental regulations in a workplace, traffic or sport events. Thus, an optimal interaction between the worker/driver/athlete and environment in which he performs can be achieved. Protective measures include design of protective gears, ergonomic optimization of the workplace or safer automobiles. Last century, with an advent of antibiotics and vaccination witnessed also a dramatic decrease of MSD caused by viral and bacterial infections (polio, meningitis). This decrease is mainly century, with an advent of antibiotics and vaccination witnessed also a dramatic decrease of MSD caused by viral and bacterial infections (polio, meningitis). This decrease is mainly attributable to application of efficient vaccines and drugs, nevertheless, overall improvement in nutrition of population is also important.

The natural ageing phenomena during the course of human life affect the physical performance of every individual, nevertheless, their effects should not be overestimated. The consequences of an increase in the age include loss of the muscle mass, loss of the bone mineral, desiccation of the ligamentous tissues, and changes in the composition of neural tissues. It should be emphasized, that these gradual natural hormonal/degenerative changes are treatable. Homeopathic lifestyle, i.e., healthy nutritional habits, and exercise are the main factors in prevention of the age related MSD (Schauff et al., 1990).

1.2 FUNCTIONAL ANATOMY OF THE TRUNK

1.2.1 General Anatomy

Designer of the human body endowed it both with unique aesthetic and functional properties. It is impossible to draw a clear dividing line between the aesthetic and functional in the human body, since the aesthetic in the human body is an expression of the ultimate functional perfection and vice versa. Creator formed human body with four movable extremities serving locomotive and prehensile functions and a head attaching onto the trunk (Netter, 1989). The relative movement of various parts of human body is made possible by a system of osseous structures serving as lever arms (able to withstand both tension and compressive loads), and muscles serving as tension force generators and actuators (able to withstand tensile force only). The trunk is the most substantive part of the human body and has a fourfold purpose: a base and reference frame for the movement of the head and the four extremities, a structure for the load transmission, a container for most of the vital organs and a structure allowing for high mobility. Whereas the first three functions would presuppose a physical body with a high rigidity, such as found in some organism possessing an exoskeleton, they are at a complete variance with the fourth requirement. Creator achieved this through the ultimate design based on principles of synergy (Fuller 1985). The vertebral column, VC, runs vertically in the posterior part of the trunk and consists of the bony vertebrae connected by intervening flexible disks and ligaments. The VC is the only osteoligamentous structure capable of transmitting vertical compressive loads (physiological and external load plus muscle forces) throughout the whole trunk, Figure 1.1. In addition to its load bearing and mobile function, the VC protects the neural cord, a high velocity transmission channel, connecting the peripheral sensory organs to the central nervous system, CNS, housed in the cranium. The VC, extending from the head to the sacrum, consists of three anatomically distinctive regions: a cervical spine, a thoracic spine and a lumbar spine. In a sagittal view, these three regions correspond to the cervical lordosis, thoracic kyphosis and lumbar lordosis, where lordosis represents an anterior curvature and kyphosis a posterior curvature. One of the advantages of the S-shape of the vertebral column is attenuation of the vertical shocks transmitted on the cranium in the erect posture.

The vertebra, a basic osseous structure of the VC, consisting of one cylindrically

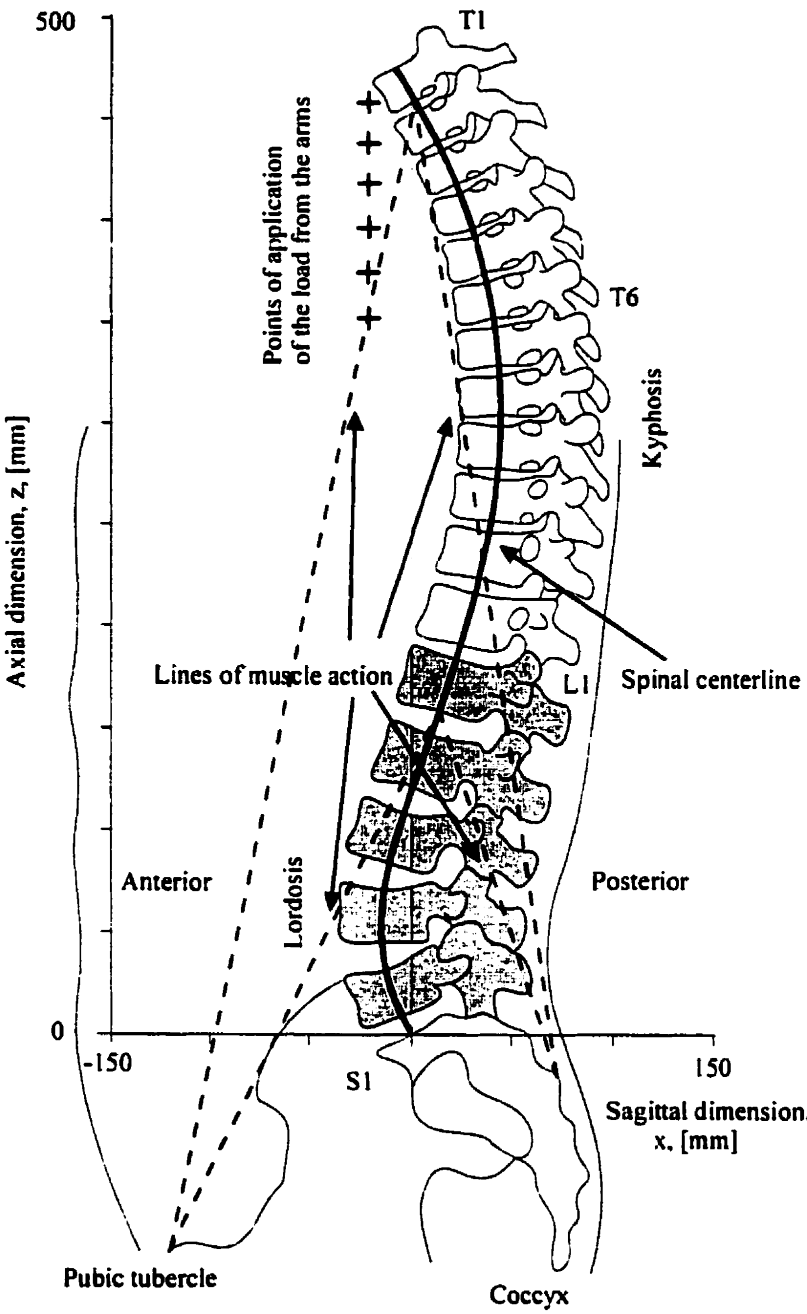

Figure 1.1 Torso with the skeletal elements of the spine and the pelvis. Lines of the muscle action for a simplified muscle model are indicated by the broken lines.

shaped vertebral body with a posteriorly placed vertebral foramen formed by the lamina and the pedicles, Figure 1.2. Laterally from the foramen are running two transverse processes and posteriorly a spinous process. Vertically, two superior and two inferior articular processes are connected on the posterior aspect of the vertebral foramen and provide articulation with the adjacent superior and inferior vertebral. Downward enlargement of the vertebral bodies (from the T1 to L5) reflects the cumulative increase in the physiological gravity load. The vertebra is filled with a spongy osseous tissue, covered by a cortical shell, in the case of the upper and lower surface of the vertebral body cortical bone forming the vertebral endplates. Intervertebral disk, a flexible structure intervenes between the endplates of the superior and inferior vertebral bodies.

The cervical region of the VC, with seven vertebrae, C1-C7, running through the neck, is highly mobile, allowing for an efficient multidirectional adjustments of the head which is a carrier of the most important sensory organs. The loads transmitted through the cervical spine are usually small, and this together with demands on an accurate motion is reflected in a fine design of the cervical vertebrae. In addition to protection of the spinal cord, transverse processi of the cervical vertebrae have the transverse foramina which form channels for protection of the right and left vertebral arteries.

In the thoracic region of the torso, a barrel like shaped thoracic cage forms a protective casing for the vital organs, the lungs and the heart which are separated from the abdomen by a dome shaped muscle, the diaphragm. The semicircular ribs articulate posteriorly with the vertebra and anteriorly with the sternum. The thoracic spine, with 12

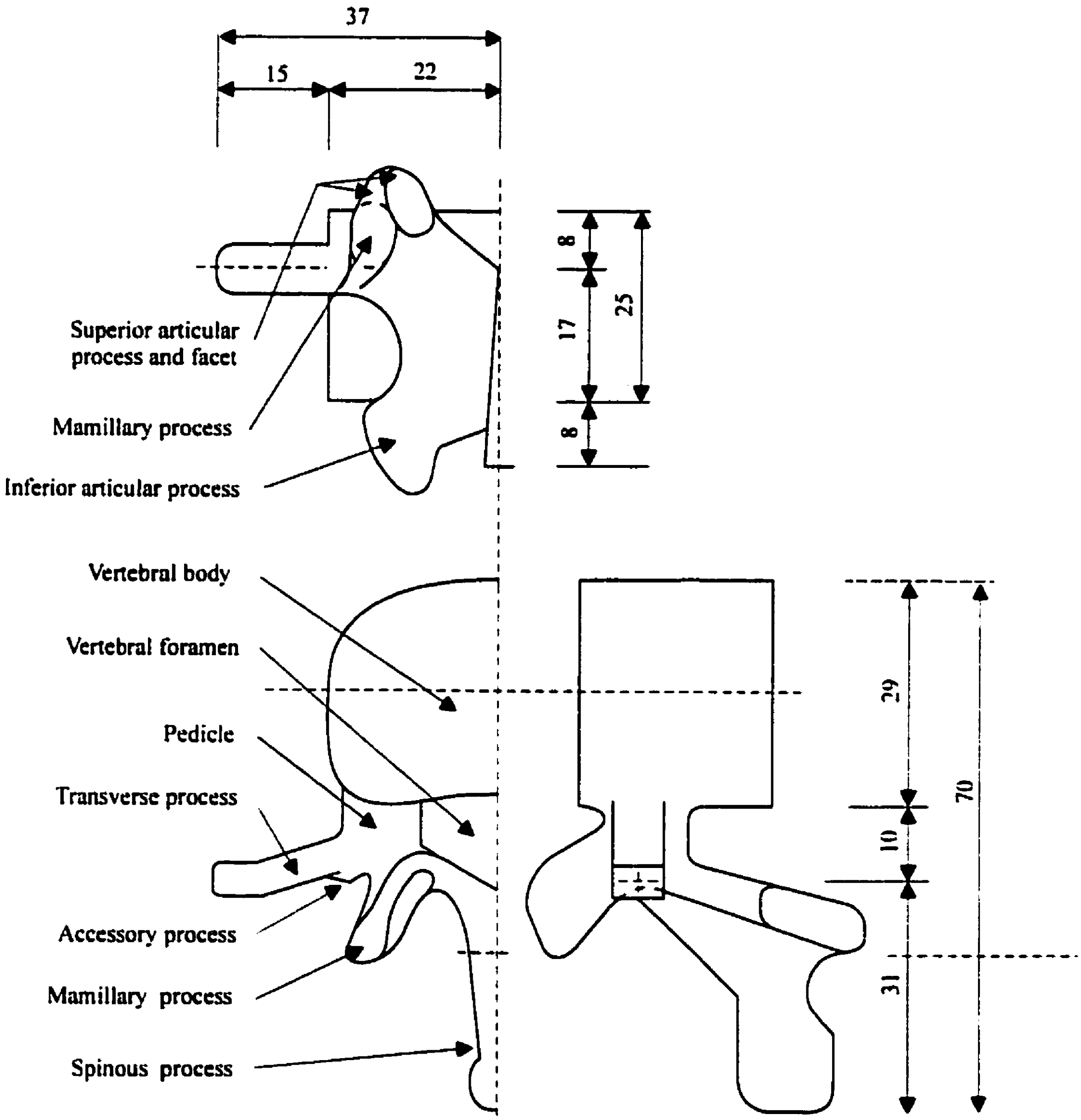

Figure 1.2 Model of a standard lumbar vertebra, Dimensions indicated in milimeters.

vertebrae, T1-T12, Figure 1.1, passes posteriorly through the upper torso and forms a posterior part of the rib cage. Each rib in its posterior aspect articulates vertically with the two successive vertebrae at the superior and the inferior costal facets and with the inferior vertebra at the transverse costal facet. This articulation spanning across the intervertebral disk and held in the place by the transverse costal facet, increases the stiffness of the intervertebral space. In addition to a local stiffening of the thoracic spine by the articulating ribs, the rib cage acting as a barrel reinforces the whole thoracic spine. Below the thoracic cage, spanning the space between the upper torso and the pelvis, in the abdomen, is located the lumbar spine, LS, Figure 1.1. The abdominal cavity located anteriorly to the LS contains vital organs such as liver, kidneys, stomach, spleen, intestines, etc. Apart from LS, no other bony structures are found in the abdominal region, and LS alone, besides the partial shielding of the abdominal organs must provide for transfer of the vertical compressive load from the upper torso onto the pelvis. In addition to the load-bearing function, the lumbar spine is a highly mobile structure. The lumbar spine consists of five vertebrae L1-L5 with intervening disks. The transverse processes, serving at the thoracic vertebrae for articulation with ribs, in the lumbar region increase in size, and serve as an attachment site for the lumbar muscles. Intervertebral disks increase in thickness in comparison with their thoracic counterparts, to allow for larger intersegmental rotations needed for a high mobility of the lumbar spine. Likewise, the intervertebral facets adapt their structure and form to the specific demands of the lumbar spine.

A vertebral segment is a unit consisting of the two adjacent vertebrae, the disk in-between the superior and inferior endplates and the ligaments. This unit allows for a relative motion of the vertebrae controlled by the visco-elastic behaviour of the disks and the ligaments. In addition to the disks and the ligaments, the intervertebral motions are also affected by the facets. Facets limit the range of intervertebral motions.and when in contact, guide the relative movement of the vertebra and participate in transfer of the load.

Ligaments connect both the body and the posterior structures of the vertebra. In the thoracolumbar region, anterior and posterior longitudinal ligaments run continuously on the respective surfaces of the vertebral bodies and the supraspinous ligament runs on the posterior aspects of the spinous processi. Ligamentum flavum connects the lamina and interspinous ligament the spinous processi of the successive vertebra. A ligamentous capsule encloses the zygapophyseal joint (the articulation between the superior and inferior facets) to protect the joint and to prevent its excessive opening.

1.2.2 Muscles

Spinal muscles can be divided into two systems, the global muscles that act on the spine through the thorax (the rib cage) controlling the overall response, and the local muscles that are attached directly to the lumbar spine controlling locally the response of the lumbar spine (Bergmark, 1989). The global muscles in addition to some control of the breathing and digestive functions (Schauff et al.,1990) actuate and stabilize the trunk by transmitting forces between the thorax and the pelvis (Hollingshead et al., 1981; Netter 1989). In the anterior aspect they include flat abdominal muscles, trunk flexors: Rectus abdominis, RA, External oblique, EO, Internal oblique, IO, and Transversus abdominis, TA, Figure 1.3. Rectus abdominis originates from the pubic arch on the pelvis and inserts into the cartilaginous costal margin between the 8^{th} left and right ribs. Anatomically it consists of two strips of vertically running muscle fibers, Figure 1.4, enveloped in the rectus sheath with the linea alba in between. The External oblique, Internal oblique and Transversus abdominis are flat sheet muscles superimposed on top of each other, forming the body wall on the sides of the abdomen. The three muscles originate posteriorly from the aponeurosis (running vertically next to the lumbar spine), superiorly from the lower costal margin inferiorly from the iliac crest and insert anteriorly into a vertically running aponeurosis formed by the rectus sheath running parallel to Rectus abdominis. Fibers composing the three flat muscles follow three general directions: External oblique runs anteriorly downward, Internal oblique anteriorly upward and Transversus abdominis runs horizontally. The abdominal muscles are innervated by the ventral ramus of the Spinal nerve. All abdominal muscles, together with the Diaphraghm are important for breathing and digestion. These muscles also control the abdominal pressure.

In the posterior aspect, close to the spinal centerline, the global muscles include Erector spinae pars thoracic, ESg, part of the Quadratus lumborum running from the 12^{th} rib to the pelvis, QLg. Erector spinae, the most powerful of extensor muscles, with its three divisions (Iliocostalis lumborum, ICg, Longissimus thoracic, LTg, and Spinalis thoracic, ST) originates from the posterior coccygeal surface of the pelvis and inserts the posterior aspects of the vertebrae. It runs parallel to the VC, and is enveloped by the thoracolumbar fascia.

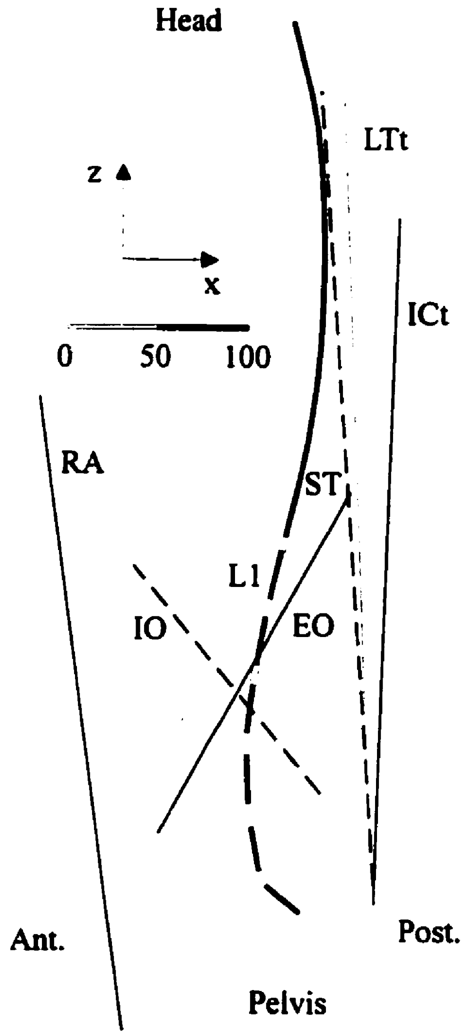

Figure 1.3 Muscles attaching onto the rib cage in a sagittal view. Thoracic erector spinae: ICt-Iliocostalis thoracic pars thoracic, LTt-Longissimus thoracic pars thoracic, ST-Spinalis thoracic. Abdominal muscles: IO-Internal oblique, EO-External oblique, RA-Rectus abdominis

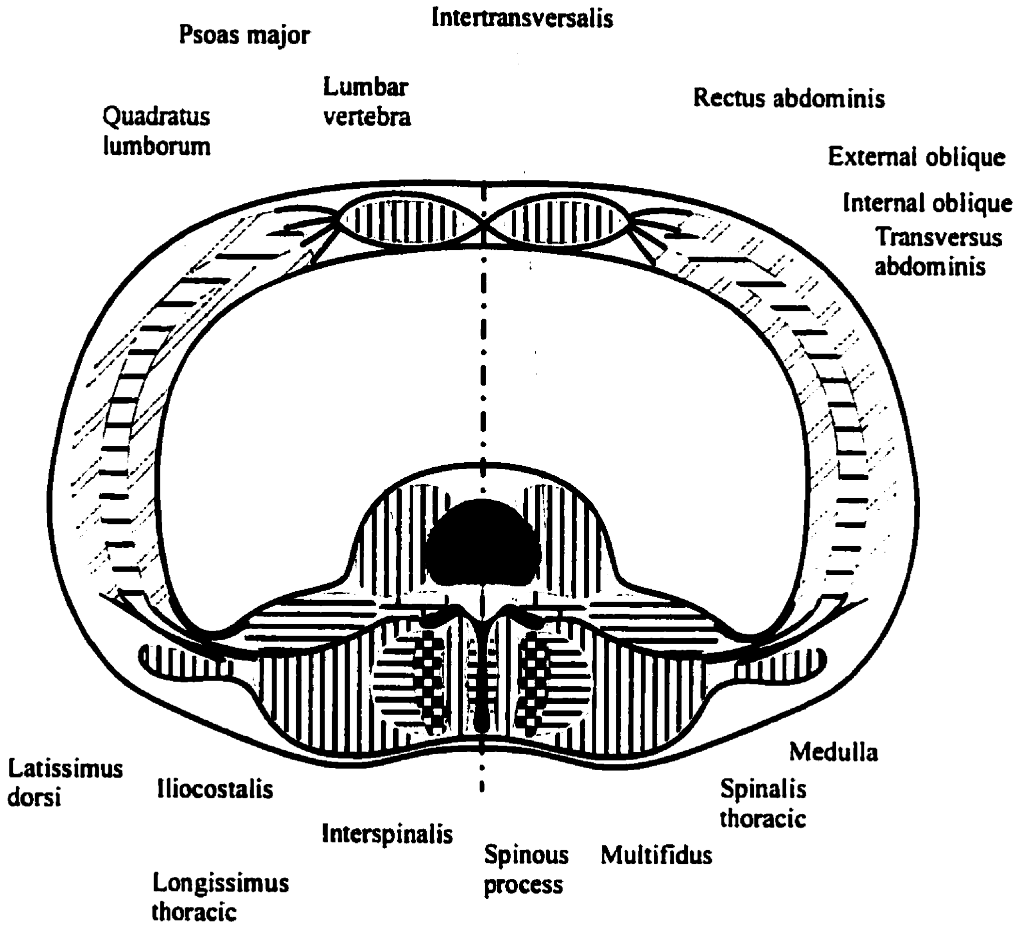

Figure 1.4 Cross section of the trunk in the lumbar region. The global and some local muscles shown.

Erector spinae is innervated by the dorsal ramus of the Spinal nerve.

The local muscles, Figures 1.4 and 1.5, insert onto the lumbar vertebrae and are located beneath the mass of the global Erector spinae. The first group of local muscles, originating from the pelvis and inserting onto the lumbar vertebrae, includes: Iliocostalis lumborum pars lumborum, IC, Longissimus thoracic pars lumborum, LT, Multifidus, MF, Quadratus lumborum (lumbar attachments), QL, and Iliopsoas, IP. The name Erector spinae pars lumborum can be attributed to the grouping of lumbar parts of two global muscles Iliocostalis lumborum pars thoracic and Longissimus thoracic pars lumborum and two local muscles Iliocostalis lumborum pars lumborum and Longissimus thoracic pars lumborum. The second group of local muscles, running from vertebra to vertebra includes: Intertranverse, Interspinalis, and some fibers of Multifidus.

Insertion points for the local muscles originating from the pelvis are located at the characteristic locations on the vertebrae, Figure 1.2, IC - Transverse process, LT - Accessory process, MF - Spinous process, QL - Transverse process, Iliopsoas - sideways in-between the vertebrae. The points of origin for the local muscles on the pelvis are located on the posterior aspect of the coccyx the iliac tuberosity and on the iliac crest.

The morphology and innervation of both local and global muscles indicate that they are not controlled on the level of anatomical muscle entities, but rather on a more detailed level of subunits such as individual fascicles. This allows for a selective muscle activation and a very accurate control of the spinal actuation.

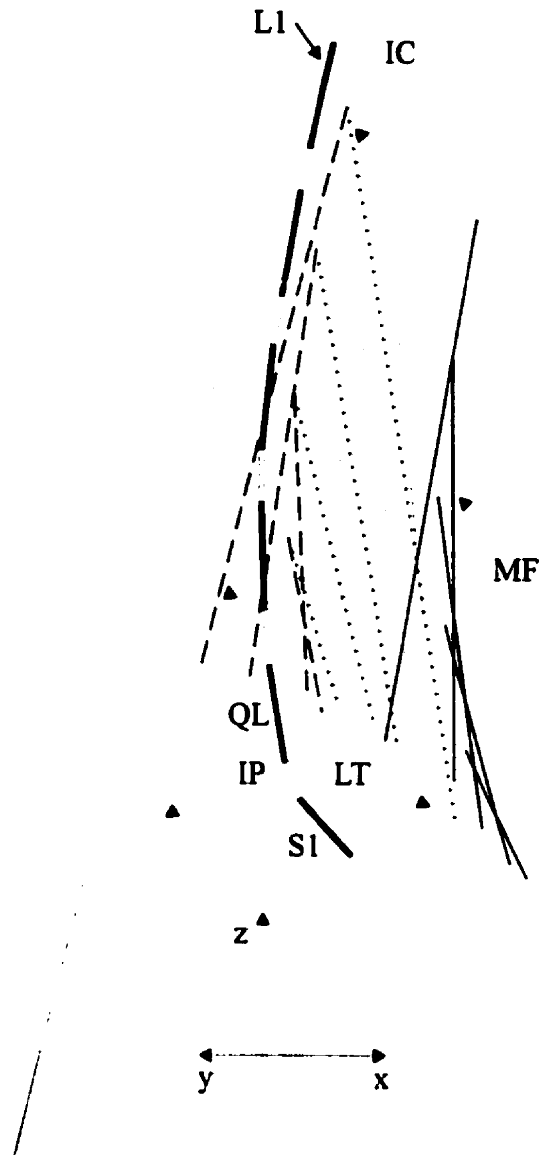

Figure 1.5 Muscles attaching onto the lumbar spine at each vertebral level. The left half in a view rotated 30 axially from the lateral. IC-Iliocostalis lumborum pars lumborum, IP-Iliopsoas, LT-Longissimus thoracic pars thoracic, MF-Multifidus, QL-Quadratus lumborum.

1.3 LITERATURE REVIEW

1.3.1 Synergy

Efficient functioning of living organisms, the highest branch of living creatures, rests upon the principles of synergy. Development of sciences from their inception in the Classical Hellenistic science incorporated principles of synergy (Kahn, 1979; Sorabji, 1988), nevertheless, due to subsequent fractionization of *Science* into the different fields (specialization), the interrelatedness of seemingly independent aspect of nature became obscure. Fractionization of *Science* into the diverse fields and establishing of borders between fields of exploration gave also rise to different *basic laws* governing each of these fields. The concept of energy is presently used as a unifying principle to relate the individual split branches of science (Borisov et al., 1993).

Out of the four elementary interactions present in Nature (electromagnetic, weak, strong, and gravitational) energistic approach allowed science to describe governing mechanisms of the first three, the last one still avoids any rational scrutiny. As the *Heisenberg's Principle of Uncertainty* indicates, energistic approach beyond certain limits becomes a theory of explaining the undetectable (Heisenberg, 1959).

Synergy presents an alternative view of *Nature* in comparison with energistic approach. The great thinker of the 20th century, architect and promulgator of synergetic principles, Robert Buckminster Fuller defines synergy: "*Synergy means behavior of the whole systems unpredicted by the behavior of their parts taken separately*" and further "*Synergy means behavior of integral, aggregate, whole systems unpredicted by behaviors of any of their components or subassemblies of their components taken separately from the whole*" (Fuller 1985). Synergy and energy are related according to him in a following manner: "*Energy relates to differentiating out subfunctions of nature, studying objects isolated out of the whole complex of Universe-for an instance, studying soil minerals without consideration of hydraulics or of plant genetics. But synergy represents the integrated behaviors instead of all the differentiated behaviors of nature's galaxy systems and galaxy of galaxies.*"

In an interdisciplinary field, such as biomechanics synergetic approach is necessary for comprehension of principles enabling a complex mechanical behavior of organic structures. In Fuller's words: "*Synergetics discloses the excruciating awkwardness characterizing present-day mathematical treatment of the interrelatioships of the independent scientific disciplines as originally occasioned by their mutual and separate lack of awareness of the existence of a comprehensive, rational, coordinating system inherent in nature*".

1.3.2 Spinal geometry, loads

Large quantities of radiographs of the spine obtained during the medical treatments of the spine are a convenient source for statistical evaluation of the spinal geometry (curve passing through centroids of disks and vertebrae), Figure 1.1, characterized by spinal anthropometric data. The most common spinal characteristics are lordotic angle and kyphotic angle. For an adult with a normal spine in neutral posture these values are: kyphosis 36°±10°(SD), lordosis 44°±12°(SD), (standard Cobb measurement) (Voutsinas and MacEven, 1984; Bernhardt and Bridwell, 1989). Any significant departure from these values might indicate a defective state of the spine. The other potential sources for evaluation of spinal geometries are computer tomography scans. Although more demanding on equipment and the resources, they can reconstruct a detailed 3-D morphology of vertebral structures. Such geometries can be used in detailed FE models of the lumbar spine (Shirazi-Adl and Parnianpour, 1996b).

The trunk is an anatomical structure with a complex geometry and has a nonhomogenous spatial mass distribution with cavities. Since the spine, a major structural element of the trunk, is a multilink flexible chain, load distribution to individual segments is important for its stability. In the neutral standing posture, the center of mass at various vertebral levels has been found to be located up to 30mm anterior from a corresponding vertebral center (King-Liu and Wickstrom, 1973), Figures 1.1 and 1.6. This physiological

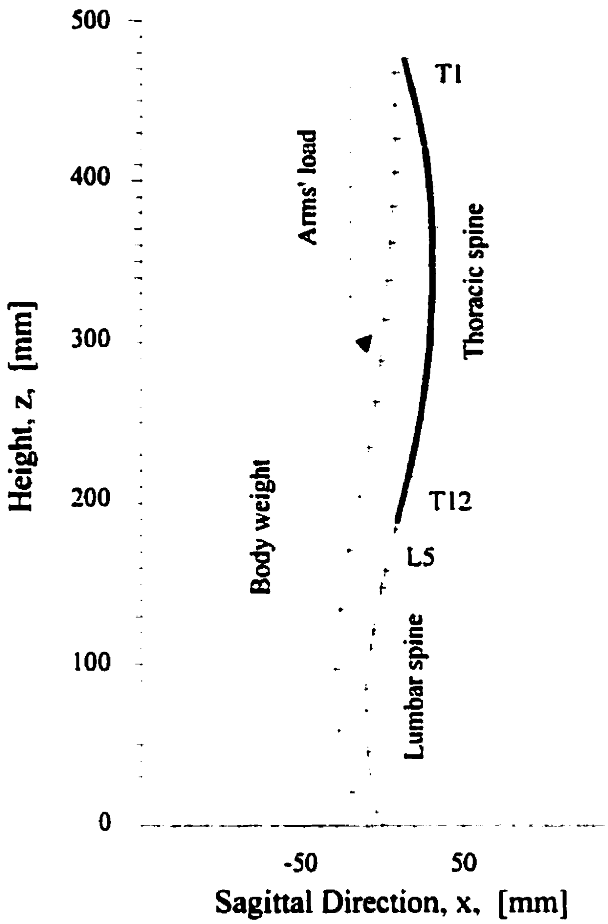

Figure 1.6 Sagittal profile of the thoracolumbar ligamentous spine. Thoracic spinner reinforced by the rib cage is modeled as a rigid body. The centroid of the body weight, and load from the arms run vertically to spinal centerline.

anterior off-centred position of a gravity load generates, in addition to axial forces, sagittal flexion moments on the TLS.

A classical attempt to measure variations of the trunk mass and mechanical properties was made by King-Liu and Wickstrom (1973). The frozen cadaveric torso was cut through the disks by a tape-saw into the transverse trunk slices. The slices were then weighed, the center of gravity was found by hanging the slice from the two points on its periphery and the gyration tensor was determined by spinning it along selected axes. The major disadvantage of this classical study is the changed nature of the postmortem anatomical structures and tissue as compared with the living one. Correlating mass properties with characteristics obtained from the γ-ray scanning (Duval-Beaupère and al., 1992) presents a non-invasive alternative for measuring mass properties and can be applied onto the living subjects. The noninvasive approach assures that the measured characteristics correspond to a realistic spinal condition, with a result being dependent on the accuracy of correlation between measured physical properties and desired mass properties.

The load effect of structures peripheral to the trunk on the spine can be considered by their distribution into the points of attachments. Anthropometric statistical evaluations give proportions between masses of body structures (head, neck, arm, torso, leg) as a fraction of the total body weight correlated with sex, age and external morphology (Zatsiorsky and Seluyanov, 1983). Attachment of the arms onto the trunk through the ribs can be identified from the radiographs, and provides a basis for distribution of arms loads onto the spine.

Changes of the spinal configuration occur as a response to the load distribution, to defects in the tissues or can be induced by control factors such as T1 spatial positioning or pelvic rotation. Experimental works on the positioning accuracy of the head in the free standing posture indicate that the T1 vertebra can be centered horizontally within 10mm (McGlashen et al., 1990). Studies on asymptomatic subjects (Jackson and McManus, 1994; Voutsinas and MacEven, 1984) and low-back patients (Keegan 1951) suggest correlation between the sacral slope (angle between the plumb line and the line parallel to the back of the proximal sacrum), lordosis, kyphosis, and sagittal position of the T1 vertebra in both groups. The decrease in lordosis is accompanied by a decrease in sacral slope, thus preserving the horizontal sagittal distance of the T1 vertebra with respect to the sacrum (Jackson and McManus, 1994). Correlation between the sacral slope and the lordosis is postulated to be an important factor in an efficient standing position (Duval-Beaupere at al., 1992). Postural adjustments are also observed in astronauts during the flights in microgravity conditions, and these involve flattening of the lumbar spine and a decrease in the sacral slope (Massion et al., 1993). Moreover, the lumbar lordosis is reported to increase when external loads are added in hands while in erect posture (Gracovetsky, 1988; Parnianpour et al., 1994).

1.3.3 Mechanical properties of the spine

Mechanical properties of the passive spinal components (disks, ligaments, bone) can be obtained by tests on specimens extracted from the postmortem tissues. The mechanical behaviour of the spinal segment becomes complex with involvement of the zygapophyseal joints, nevertheless, it was found that during the intersegmental angular changes less that 1.5° in the extension, 3° in flexion and lateral bending and 1° in axial rotation, the spinal response is affected mainly by the ligaments and the disks (Markolf, 1970). The relation between the applied moments and rotations for this range of motions can be approximated as linear. Behavior of the lumbar spinal segments (normal/defective) has been investigated both in experimental studies (Markolf, 1970; Panjabi et al., 1984; Tencer et al., 1982) and by numerical modeling (Shirazi-Adl et al., 1984; Spilker et al., 1984). Experimental studies performed on the postmortem spinal segments, with posterior structures of the vertebra removed (eliminating effects of facets) indicate variation of segmental stiffness in different loading conditions according to the sex, age and degree of degeneration (Koeller et al., 1986). Values for the disk stiffness derived from experimental values correlated with the complex FE segmental models can provide input for the lumped stiffness spinal models (Shirazi-Adl and Parnianpour, 1993; 1996a). For deformations of the lumbar spine close to the neutral posture, the elastic response of the segment can be described by a beam characterized by the cross sectional properties and associated moduli (E=7MPa, G=3MPa) (Shirazi-Adl and Parnianpour, 1993; 1996a).

The rib-cage serves as an attachment site for the major spinal muscles and plays also an important role in stiffening of the TLS. Experimental studies show that axial rotations in the thoracic region, with the trunk rotated axially to its limits, are substantial, in spite of the stiffening effect of the costal articulations (Gregersen and Lucas, 1967). Numerical simulation of the rib-cage can be performed by taking into account mechanics of the costovertebral joints and the rib cage (Andriacchi, et al., 1974) when detailed response of the trunk to the horizontal load application such as a car crash is needed. The comparison of the numerical predictions by Andriacchi, et al., 1974, and experimental observations of Gregersen and Lucas, 1967, indicates a need for more appropriate modeling of the costovertebral articulations during maximal trunk rotations. Nevertheless, for conditions under a moderate vertical load with spinal motions confined within a limited range, articulations of the rib-cage with the vertebrae appear to stiffen the thoracic spine. An increase of the Young's modulus in the thoracic spinal segments can be used to simulate the behavior of a complex thoracic model (Andriacchi et al., 1974; Shirazi-Adl and Parnianpour, 1996).

The creep shortening of the TLS, whose approximately 30% of length consists of the intervertebral disks, is a natural phenomenon observed throughout the diurnal cycle. The normal axial shortening of the TLS under the body weight during the diurnal cycle ranges from 0.5 to 2% of the total body height (Eklund et al., 1984) and is fully recuperated during the night in the absence of the vertical load (Adams et al., 1987). Workplace automation has led to an increased number of occupations requiring prolonged seated or standing postures (Eklund et al., 1983). Epidemiological studies have identified these prolonged static postures as a risk factor in the low back disorders (Anderson, 1981; Frymoyer et al., 1983).

The creep response of the spinal motion segment was investigated both in experimental (Burns et al., 1984; Kazarian, 1975; Li et al., 1995) and numerical studies, e.g., a detailed FE model used for a creep analysis of the spinal L2-L3 motion segment (Wang, 1995). Although the existant work provides a good insight into the mechanics of the spinal motion segment in the creep conditions and the diurnal variation of the body height, the stability requirements of the spinal column in prolonged postures require further elaboration (Adams et al., 1987, Fathallah et al., 1995).

1.3.4 Spinal muscles

Anatomy of the human spinal muscles can be described in terms of their origins, insertions, and cross sectional areas. Dissection studies (Macintosh and Bogduk, 1987; 1991) as well as CT scans (Chaffin et al., 1990; Han et al., 1992; Moga et al., 1993; Tracy et al., 1989) provide reliable data for construction of muscle architectures for the numerical models.

The complex anatomy of spinal muscles can be classified according to their insertions as muscles attaching either onto the rib cage or lumbar vertebrae. Muscles originating from the pelvis and inserting onto the rib cage are termed as global muscles, Figures 1.3 and 1.4. Function of the global muscles is to actuate and control distribution of weight (physiological and external load) on the spine (Bergmark, 1989). Muscles attaching onto the lumbar vertebrae are termed as local muscles. They include muscles spanning from the pelvis and attaching onto the lumbar vertebrae and muscles both originating and inserting onto the lumbar vertebrae, the intersegmental muscles, Figures 1.4 and 1.5. The two muscle groups and the passive spine, with its double curved shape, cooperate in a synergetic manner (Broberg 1981; Bergmark, 1989. The global system (Bergmark, 1989) includes Rectus abdominis, RA, External oblique, EO, Internal oblique, IO, Ilio costalis lumborum pars thoracic, ICt, Longissimus thoracic pars thoracic, LTt, and Spinalis thoracic, ST. The local system (Bergmark, 1989) includes Iliocostalis lumborum pars lumborum, IC, Iliopsoas, IP, Longissimus thoracic pars lumborum, LT, Multifidus, MF, and Quadratus lumborum, QL (Bogduk et al., 1992; Macintosh and Bogduk, 1987; Macintosh et al., 1991; Macintosh et al., 1993). The lines of action of the thoracic erector spinae, ICt, LTt and ST, can be modeled as straight lines (Bergmark, 1989; Stokes and Gardner-Morse, 1995). The straight line approach neglects stabilization of the spine by the constraining effect of the thoracolumbar fascia. Several parametric anatomical studies (Chaffin et al., 1990; Dumas et al., 1988; 1991; McGill et al., 1988; Moga et al., 1993; Nussbaum and Chaffin, 1996) provide coordinates of insertion points for the global muscles.

The data on spatial orientation (direction angles) of the muscles obtained from a large number of individuals (dissection or tomography) is useful for statistical anthropometric description of human anatomy. Nevertheless, to provide data for biomechanical models, muscles should be characterized in conjunction with their insertions and origins on the osseous structures (vertebrae, pelvis). With respect to the lumbar vertebrae, insertions of the local muscles can be located at the following landmarks: IC - Transverse process, LT -

Accessory process, MF - Spinous process, QL - Transverse process, Iliopsoas - sideways in-between the vertebrae (Bogduk et al., 1992; Macintosh and Bogduk, 1991; Santaguida and McGill, 1995), Figures 1.2, 1.4 and 1.5. A geometric model of a standard lumbar vertebra can be used to obtain insertion coordinates by fitting it into the spinal geometry of the FE model, Figure 1.2. Similarly, a 3D geometric model of the pelvis (Netter, 1989; Nussbaum and Chaffin, 1996) can be scaled according to the anthropometric data (Nussbaum and Chaffin, 1996) and the spinal geometry of the model to locate coordinates of muscle origins identified by their attachment areas (Bogduk et al., 1992; Macintosh and Bogduk, 1987; Netter, 1989; Santaguida and McGill, 1995). Available biomechanical models use straight lines of action for the muscles inserting onto the lumbar vertebrae.

Electromyography (EMG) tests record the electrical activity in muscles by superficial electrodes placed on the skin or by invasively implanted silver electrodes. EMG experiments with electrodes implanted inside the muscles allow for good resolution in detecting activities in the individual muscle fibers. In the upright erect posture under physiological load, majority of activities at all lumbar levels is observed in the Multifidi muscles, minor activities are observed also in the Iliocostalis lumborum pars lumborum and Quadratus Lumborum (Anderson et al., 1996; McGill et al., 1996). Psoas major, as indicated by numerical studies and confirmed by the experimental EMG studies with implanted electrodes in neutral posture does not exhibit substantial activities (Bogduk et al., 1992). Similar to findings obtained by the implanted electrodes, the EMG measurements with the surface electrodes record relatively low muscle activities while maintaining the erect posture with and without weights up to 223N carried in each hand (Parnianpour et al., 1994). Other findings indicate that the activations in spinal muscles in response to small sagittal changes in posture vary linearly (Raschke and Chaffin, 1996) with the distance from the initial T1 position.

1.3.5 Spinal load-bearing capacity and stability

Human spine consists of the passive (osteoligamentous spine) and active (muscular) subsystems (Broberg, 1981; Bergmark, 1989), and simultaneously transfers the loads trough the trunk and positions it in space, Figures 1.1, 1.3 and 1.5. Studies of spinal stability can quantify effect of loads on the passive spine with or without effect of muscles. The in-vitro experiments on a human cadaveric thoracolumbar osteoligamentous spine, TLS, extending from the T1 to S1 and devoid of musculature, indicate that it can carry in the lateral plane only up to 19.5N of the vertical load applied at the T1 with the T1 free to move and up to 170N with the T1 prevented to move horizontally (Lucas and Bresler, 1961). Numerical simulations of the behavior of the osteoligamentous spine indicate buckling loads similar to those obtained from experiments (Andriacchi et al., 1974; Scholten et al., 1988). These simulations allow to go beyond limitations of experiments and to perform an effective simulation of the rib cage presence or variations in properties of the spinal components. The experimental and numerical results indicate that the passive spine is able to withstand only a fraction of the physiological gravity load when unaided by the muscles or other mechanisms enhancing the load distribution such as the pelvic rotation.

Even if the passive spine can carry only very small loads, the EMG measurements by the surface electrodes during the in-vivo loading experiments on volunteers with asymptomatic spine measure relatively low superficial muscle EMG activities while maintaining the erect posture with and without weights up to 223N carried in each hand (Parnianpour et al., 1994). With the muscular activities in neutral posture very low and surface electrodes measuring electromyographic signal with cross-talk between the numerous spinal muscles, the only potential way to measure the activities with a better resolution would be using implanted electrodes. Apart from the ethical considerations, the measured electromyographic signal would be still only a qualitative indicator of muscle activities, since there are some indications that there might be no direct correspondence between the EMG signal in the muscle and its force (Feldman, 1986).

The alternative to experimental examination of muscle activities is presented by a numerical modeling. Since the muscle is a unit containing a contractile element in series with an elastic element, a number of parameters is necessary to accurately describe its behavior (Feldman, 1986; Kaufman et al., 1989; 1991; Kenton, 1989; Schultz et al., 1991; Winters and Stark, 1988; Woittiez et al., 1984). For static the loads and muscles close to their isometric state, equation describing the muscle in terms of its stiffness can be simplified to contain a single stiffness parameter "q":

$$k = q \frac{S}{l} \tag{1.1}$$

where k is the current muscle stiffness, S is the muscle force, q the experimentally derived coefficient and l the muscle length (Bergmark, 1989; Gardner-Morse et al.,1995). During the equilibrium analysis, muscles are often modeled as simple force generators, grouping together passive and active components of muscle force (Bergmark 1989; Dietrich et al., 1990). In the numerical models with a realistic muscle architecture the muscle problem becomes redundant, i.e., number of muscles significantly exceeds number of equilibrium conditions available to determine their forces (Allard et al., 1991; Kaufman et al., 1991). A simple and efficient way for solution of the muscle problem is to simplify the muscle architecture by grouping the muscle with similar effects (Bergmark, 1989) until the number of equilibrium conditions is equal to the number of muscles groups - muscle problem becomes statically determinate. More complex methods try to duplicate control functions embedded in the central neural system, CNS, and the spinal reflex which govern the pattern of muscle activations. This can be achieved by optimization (Allard et al., 1991, Kaufman et al., 1991), neural networks (Pellionisz and Llinas, 1979; 1980; 1982; Pellionisz, 1983), lambda model (Feldman, 1986) or correlation with experimentally obtained data such as EMG (Cholewicki, 1993).

Several models for static analysis of the spine can be found in the literature: maximum load-bearing capacity models (Stokes and Gardner-Morse, 1995; Bogduk et al., 1992), transverse section equilibrium models (Schultz and Andersson, 1981), and stability criterion models (Bergmark, 1989; Gardner-Morse et al., 1995; Crisco and Panjabi, 1991).

The experimental measurements such as EMG activities, deformations or disk pressures can be used for independent verification of numerical results or as the input data to decrease redundancy in muscle analyses (Gracovetsky and Farfan, 1986; Cholewicki and McGill, 1995).

In the models for the maximum load resisting capacity the muscle architecture with associated physiological cross sections for individual muscle fascicles is obtained from dissections (Macintosh and Bogduk, 1987; 1991) and the CT scans (Chaffin et al., 1990; Han et al., 1992; Moga et al., 1993; Tracy et al., 1989). The resultant moment resisting capacity becomes a function of the maximum contractile stress in the mammalian muscle assumed to be between 0.4 to 0.8 MPa (Bergmark, 1989). Changes in the muscle orientations associated with changes in spinal configuration effect the potential of the lumbar spine to resist external flexion moments (Macintosh et al., 1993). The inclusion of the passive bending resistance of an individual lumbar motion segments significantly increases the maximal moment bearing capacity of the spinal system as compared with ball and socket joints (Stokes and Gardner-Morse; 1995).

In the static equilibrium models for maximum resistance to external load, decoupling between compressive load and moments in sagittal and lateral planes is often assumed, (Ladin et al., 1989). An optimization procedure is then used to determine activations in the muscles with an objective to reach the maximum sagittal, lateral or axial moment at the L1-L5 (Bogduk et al., 1992) or T12 (Stokes and Gardner-Morse; 1995) vertebra.

The study of spinal stability by Bergmark (1989) uses a statically determinate model of the TLS, with muscles divided into the local and the global systems, to derive conditions for spinal stability in sagittal and lateral directions under sagittally symmetric loading in terms of minimal muscular activities. The size of the muscle problem is reduced to a number of available equilibrium conditions by grouping of individual fascicles and prescribing fixed relations between activations in functionally similar muscle fibers. The minimal values of the coefficient "q" (assumed the same for all muscles) necessary for maintaining of the spinal stability (Bergmark, 1989) is found to be around 40. More recent studies (Gardner-Morse et al., 1995) use a linear programming algorithm to resolve of the muscle problem during the maximal exertions, and report values of the muscle stiffness coefficient, q_{crit}, necessary to maintain the spinal stability, to be around 5.

A qualitative study of functionally different muscle fibers attached onto the lumbar spine using a simple muscle model suggests that the group of muscles running from the vertebrae to the pelvis, e.g., the Erector spinae, Multifidus, Iliopsoas and Quadratus lumborum muscles, is more efficient for maintaining the stability of the lumbar spine than the intersegmental muscles (Crisco and Panjabi, 1991).

An optimization method used on a simple, continuous beam model, of the TLS with selected muscles suggests the stabilizing potential of lateral moments in frontal plane in the lumbar region when carrying 300N load in one hand (Hjalmars, 1988). The effect of the horizontal support and a flexion moment applied at the L1 increases the load-bearing capacity of the lumbar spine up to 400N with minimal horizontal displacements (Shirazi-Adl and Parnianpour, 1993). Although these works (Hjalmars, 1988; Shirazi-Adl and Parnianpour

1993; 1996ab) recognize importance of the various constraints and moments in the compression load-bearing capacity of the TLS, they do not identify an exact mechanism by which they are generated.

Two approaches are available for solution of a static biomechanical optimization problem involving an elastic structure combined with a redundant muscle problem. When a final geometry is known from an experiment, this can be used as an input in a solution of the muscle problem (Backward static optimization, BOS). Unknown final geometry requires a simultaneous solution both for the muscle forces and the geometry (Forward static optimization, FOS) (Allard1995).

Muscles crossing a joint maintain its stability by their coactivation (Hogan 1984). The stiffness of a stable human ankle joint, stabilized in the plantar flexion and extension by coactivation of the calf muscles, during the neutral posture is reported to be only slightly above its critical value (Bergmark 1989).

1.4 OBJECTIVES AND PLAN OF THE THESIS

Neutral posture is a condition of the spine sustained throughout the daily activities and as such, deserves attention equal to the states of spine with close to maximal loadings. The objective of this work is to examine some mechanisms affecting spinal stability in the neutral posture related to the interaction between the passive (osteoligamentous) and active (muscular) components of the spine. The approach for investigation of the neutral equilibrium phenomena should be computationally efficient to perform a large number of analyses, and at the same time it should incorporate important mechanisms of the spinal behavior. The approach should also take advantage of already available commercial software and have a flexible modular structure allowing for addition of new and modification of its components.

The main objectives of this study can be summarized as:

(i) to investigate effects of mechanisms already reported in the literature such as pelvic rotation, T1 sagittal positioning or effects of constraints from muscle activations on the spinal behavior,

(ii) to create a numerical model capable of accommodating these selected mechanisms of spinal control,

(iii) to develop an interface for interaction between the passive (osteoligamentous) and active (muscular) spinal components,

(iv) to test the proposed approach in the static conditions with a simple muscle architecture without/with time dependent behavior (creep),

(v) to extend the simple model into a model with a realistic muscle architecture,

(vi) to identify conditions governing patterns of muscle activation in neutral posture, estimate effect of muscle activations on the mechanical stability of the spine.

The main part of this work is contained in three publications attached in the appendices. Chapter 1 reviews origins of the musculoskeletal disorders, some aspects of the functional anatomy of the trunk and reviews literature relevant to biomechanical investigations of the human spine. Chapter 2 is a compendium of the three publications attached in the appendices. The results and conclusions of the publications are presented in a general context. The common idea underlying this study and its continuity is examined with respect to the publications. Chapter 3, the last chapter, discusses limitations and originalities of the proposed approach. The predictions of the approach are compared with the previous works and some future developments are outlined.

Three publications submitted in a partial fulfillment of the requirements for the Ph.D. degree are attached as Appendices A, B, C. Performance of the beam elements from the ABAQUS element library in a pure compression is presented in Appendix D. Appendix E outlines some concepts of adaptive control of a flexible column with modal transitions. Appendix F contains details of the numerical approach for the spinal analysis employed in this study. Appendix G presents some of the most frequently asked questions which arose during the work on the publications.

CHAPTER 2

COMPENDIUM OF ARTICLES

2.1 INTRODUCTION

Sustaining of the upright neutral posture in the gravitational field of the Earth is a natural state of the spine endured for prolonged periods of time in daily activities throughout the human life. Proper maintenance of this condition is essential for the individual's well-being. The spine is an elaborate organ providing human body with a biomechanical structure for both an efficient load transfer and locomotion, i.e., combining rigidity and flexibility.

The spine assumes its load-bearing function only gradually during the first years of the human life, in contrast to some other biomechanical structures such as limbs. The triple curved shape of the flexible spine (cervical lordosis, thoracic kyphosis, lumbar lordosis) and its axial rigidity develop as a response to the effects of gravity during transition to an erect walking posture from an initial arched shape in a newborn infant (Taylor and Twomey, 1985).

The spinal resistance to vertical loads originates from two sources: passive spine (vertebrae with disks and supportive ligaments) and active spine (muscles acting as actuators and force generators). The adaptation of the infant's spine to the gravity load in an upright posture represents a period of neuromuscular training and shape developments of the passive spine. The physiology and adaptive development the spinal apparatus for the upright posture in the gravitational field of the Earth, suggest that it operates on a synergetic level: "behavior of the whole system is unpredicted by the behavior of its parts taken separately" (Fuller, 1975). Human spine is a synergetic structure adapted to cope with physical properties of the surrounding Terrestrial environment. The proper development of the load-bearing function of the spine is unthinkable in the absence of an appropriate gravity field, such as on the orbital stations or in a person confined since infancy to bed, and can adversely affect other spinal functions.

In addition to the effects of the static load, the spine is subjected also to time dependent changes resulting from the creep of intervertebral disks. In the circadian cycle (24 hrs), the spine during the period of daily activities in standing and sitting positions loses part of its length due to the creep of disks which is recovered by the swelling of disks during the sleep period, when the axial loads on the spine are minimal. Although, the creep of the disks affects the spinal configuration, it is a necessary mechanism for transport of biological substances (nutrients, products of metabolism) into and out of the disks.

Spine in the upright posture must provide for both sufficient load-bearing capacity and stability. Load-bearing capacity is governed by the equilibrium conditions. The definition of the spinal stability can be based on the clinical or mechanical criteria (Bergmark, 1989). For the purpose of present study, spinal stability is considered in the same terms as stability of any mechanical structure: structure is stable under a specific load prior to its transition into the hypermobility. Both linear and nonlinear theory of stability can be used to investigate the structural stability, the former describing structure in terms of the critical loads only, the latter provides a full load-displacement history.

When devoid of musculature, the human thoracolumbar spine, TLS, can sustain only minimal vertical loads. At the same time, EMG measurements on normal individuals in the upright erect posture indicate low activities in spinal muscles. Since neither the muscles nor the passive spine alone can account for any substantial part of spinal load-bearing capacity, some other factors are necessary for maintaining of the spinal stability in the neutral posture.

Since the head, whose position is controlled by the spine and its accessory muscles, houses the major neural, proprioceptive and other sensory organs, it is imperative that it possesses a superior positioning accuracy. Experiments quantifying the accuracy of human proprioceptive control of the upright posture show, that persons can center their head with several millimeters precision both in sagittal and lateral directions. This provides a plausible parameter for control of the human posture. Nevertheless, this mechanism appears to lose some of its efficiency in the absence of the terrestrial gravitational field due to its linkage to the vestibular apparatus.

This work develops a new hybrid approach for analysis of the spine by integration of multipurpose commercial program for structural analysis ABAQUS with some in-house built optimization modules. The performance of this approach is first evaluated on a spinal model with a simplified muscle architecture in static and later static time-dependent behaviour. Once the advantages and limitations of the approach are determined, it is applied to a model with more realistic muscle architecture in static conditions.

2.2 METHOD

The passive spine used in all three articles has the same geometry corresponding anthropometrically to a female person with a body weight 68kg and height 170cm (Figures 1.1, 1.6 and Appendix C). Elastic constants for the beams representing the intervertebral disk were found from the literature as E=7Mpa and G=3Mpa. The thoracic part of the spine, reinforced by the rib cage, was considered to be stiffened by increasing the Young's modulus for the thoracic disks. Mechanical properties of the passive spine, including creep parameters, were based on data available from the literature (Appendix A). To quantify the instability effects, large deformation analysis was performed.

Muscles, the active spinal components, are divided into global and local groups, according to their points of attachments, respectively, onto relatively rigid thoracic spine, stiffened by the thoracic cage, and a flexible lumbar spine.

For the spine, the static equilibrium between the load, $\{F^l\}$, active, $\{F^m\}$, and passive, $\{F^p\}$, joint forces can be expressed as:

$$\{F^l\}=\{F^p\}+\{F^m\} \tag{2.1}$$

The loads acting on a spine are resisted by forces generated by the passive tissues, and forces generated by the muscles. Position of a point on the spine is controlled by constraints (springs) with stabilizing forces $\{F^c\}$ Figure 2.1. Since these stabilizing forces are produced by the muscles, they must be associated with a compressive penalty $\{\Delta P\}$ as:

$$\{F^m\}=\{F^c\}+\{\Delta P\} \tag{2.2}$$

The load-resisting forces are associated with stiffness matrices for the passive tissues, $[K^p]$, and spring constraints, $[K^c]$, and their common vector of displacements, $\{u\}$, as:

$$\begin{gathered}\{F^l\}+\{\Delta P\}=([K^p]+[K^c])\{u\}\\ \{\bar{F}\}=[\bar{K}]\{u\}\end{gathered} \tag{2.3}$$

where loads acting on a vertebral segment and compressive penalty form the vector $\{\bar{F}\}$ and passive stiffness and constraint forces form the matrix $[\bar{K}]$. The general equation (2.3) can be then solved for both unknown displacements and muscle forces. Two other ingredients are necessary for solution of Eq.(2.3), namely: solution of the muscle problem, Eq.(2.2) and selection of the constraint springs, $[K^c]$.

The muscle problem (calculation of muscle forces) with given constraint forces can be formulated as:

$$f^c_i = a_{ij} s_j \tag{2.4}$$

where f^c_i are the constraint forces which are a function of displacements at a disk level and of the muscle forces s_i. The magnitudes of the muscle forces s_i are related to the constraint forces f^c_i by coefficients of geometric transformations a_{ij}. After resolution of the muscle problem, the compressive penalty ΔP at the constraint point becomes a function of muscle forces s_i and coefficients b_i (Appendix F):

$$\Delta P = s_i b_i \tag{2.5}$$

Selection of the constraint springs, $[K^c]$, is related to the overall stability of the spine. Higher stiffnesses of the constraint springs have a potential for producing higher stability, although, at the same time, accompanied by an increase in the compressive penalty. Lower values of constraints result in smaller compressive penalty and a higher participation of the passive spine in generating of the load resisting capacity. The stability margin of the spine (magnitude of the added load prior to transition into hypermobility) in the neutral posture is related to the severity of a potential damage by the sustained load. The higher the load - the more severe the potential damage for the spine. Thereby, in the neutral posture under the physiological load, constraint spring stiffnesses for maintaining of the equilibrium can be kept low, at the same time providing a sufficient stability margin. In addition to these considerations, appropriate values for the constraint springs can be estimated by a parametric study. Trial of different constraint spring stiffnesses allows to determine effect of their variation on the results of the analysis.

The hybrid kinematic approach was applied to examination of the static behaviour of the human spine in the neutral posture in the following conditions:

publication 1	passive spinal response + conceptual muscle representation,
publication 2	passive spinal response with time-dependent behavior + conceptual muscle representation,
publication 3	passive spinal response + realistic muscle representation.

The difference between the first and third publication resides in the complexity of muscle architecture and thereby in a solution of the muscle problem. The second publication applies the method developed in the first publication to a time dependent behavior, creep, of the disks. The lines of action for the conceptual muscle architecture are selected within the confines of the trunk, Fig2.1. In the case of this model, the constraints in the virtual springs control motions at the insertion points in the horizontal plane and yield only two equilibrium conditions for solution of the muscle problem (2.5) with four muscles. A criterion for selection of the two directions for the active muscles out of the possible four, based on a direction of the horizontal constraint force, reduces the muscle problem to a statical determinacy (Appendix F).

The steps described by Eqs. (2.4), (2.5) and (2.6) are repeated in an iterative manner, with a compressive penalty updated in each step Figure 2.2 until the equilibrium of the whole system is reached.

The muscle architecture for the third publication is constructed by connecting the points of the insertion of the muscles on the vertebrae with corresponding points of origin on the pelvis. A standard vertebra, Figure 2.1, positioned in the vertebral centerline, is used to identify coordinates of landmarks for insertion of muscles. Pelvis, antropometrically scaled to fit the spinal geometry, is used to obtain the coordinates for the origins. The movement of each vertebra is controlled in five possible directions (horizontal displacement + three rotations), thus yielding five possible constraint equations for solution of the muscle problem. Nevertheless, the number of muscles attaching onto each vertebra is not less than ten, thus exceeding the number of constraint conditions. An optimization method (linear programming), with the compressive load resulting from the muscle action as a cost function, is used at each vertebra to determine the active muscles and their forces, Appendix F.

2.3 RESULTS

Comparison of muscle forces in the global group (muscles attaching onto the rib-cage) predicted in publication 1 (Figure 11) and publication 3 (Figure 5) shows a good agreement in terms of the overall activation quantified by the compressive penalty. Advantage of the more

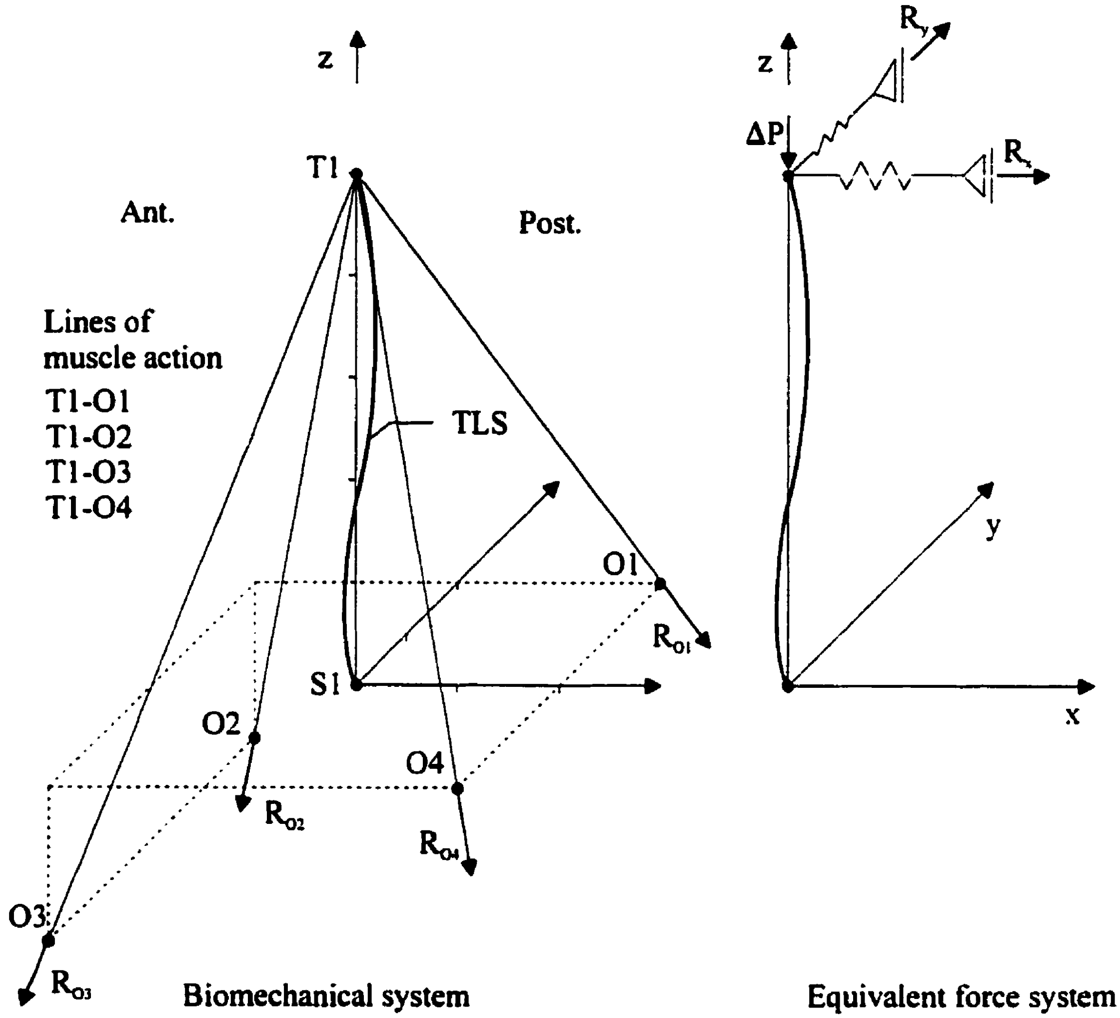

Figure 2.1 Simplified muscle model. Only the insertion point at the T1 with its points of origin at the pelvis (O1, O2, O3, O4) is shown. Springs at the point of insertion in the equivalent force system used to evaluate muscle activations and allow vertical movement only.

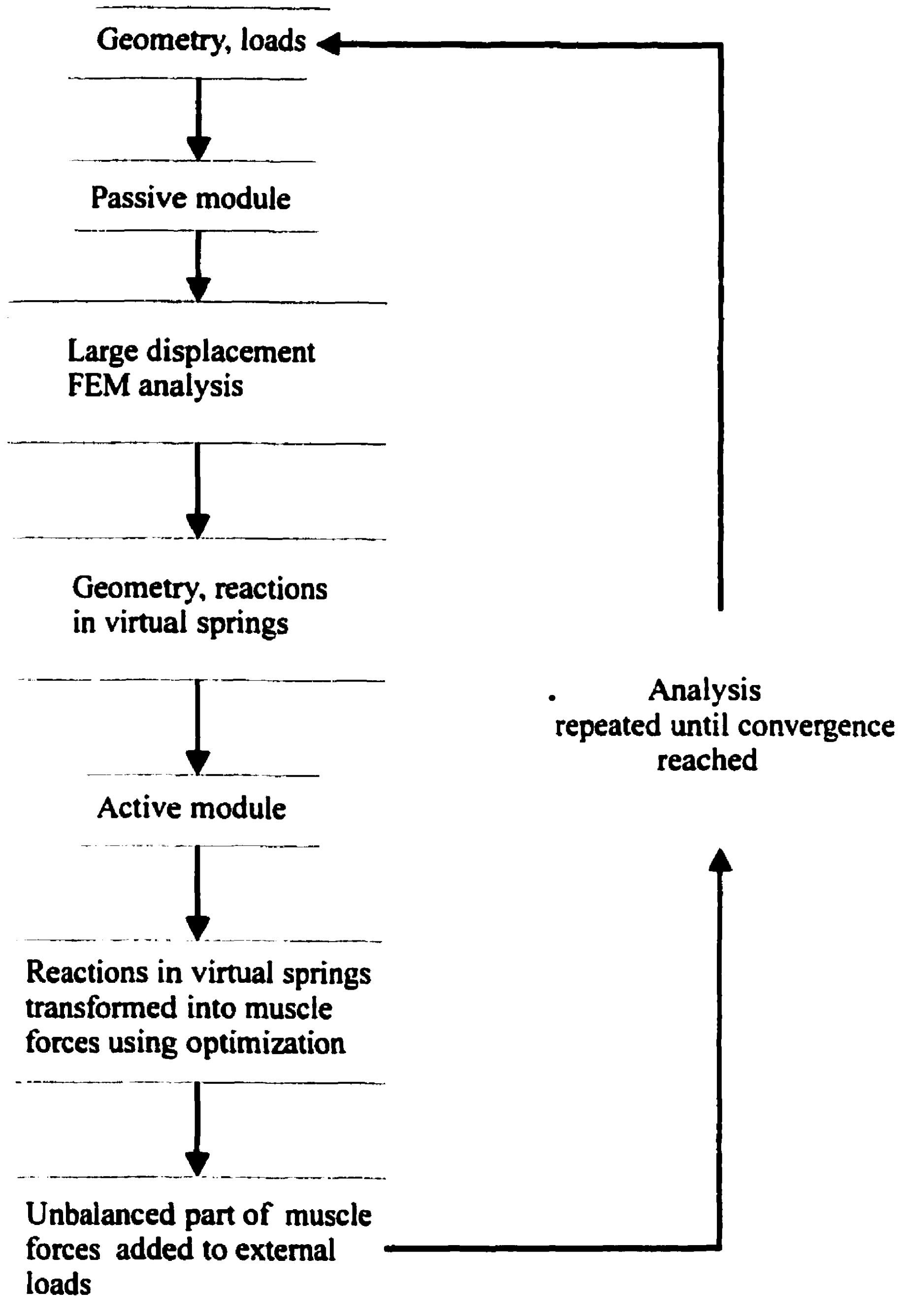

Figure 2.2 Flow chart illustrating the numerical procedure with interaction between the passive and the active modules.

complex muscle anatomy for the global muscles resides mainly in the improvement of the resolution of muscle activations and thereby a better control of the stability. For the local muscles, considering the detailed muscle architecture is necessary for obtaining a realistic quantification of spinal stability. Nevertheless, for both conceptual and detailed muscle architecture, the mechanism of interaction between the passive spine and the muscles, based on equilibrium conditions, remains very similar.

The second publication extends the first one into the domain of the time dependent behavior. The creep of the disks has a significant effect on a configuration of the spine. The vertical displacement of the spine at the T1 can reach up to 25mm after one hour after the application of the physiological load with the global muscle group active only, Figure 2.3. This displacement is decreased almost by 50% with activation of the local muscles. Significant reduction in the T1 deflection is due to decrease in the sagittal moments and thereby the creep rotations of the disks. Although this has as a consequence some increase in the compressive penalty, it is far outweighed by the decrease in sagittal moments.

Pelvic rotation can balance the loads acting on the spine and varies both with the magnitude of the applied load and with time. The variations of pelvic rotation related to the creep of the disks (publication 2 Figure 3) significantly exceed those with a gradual application of the physiological load (publication 1 Figure 5). The T1 positioning is another factor influencing configuration of the spine with the creeping disks. In the positions with the pelvis not allowed to rotate, or the T1 not to move freely, variation in muscle forces is necessary to maintain the spinal equilibrium under the physiological load (publication 2, Figures 3, 4b).

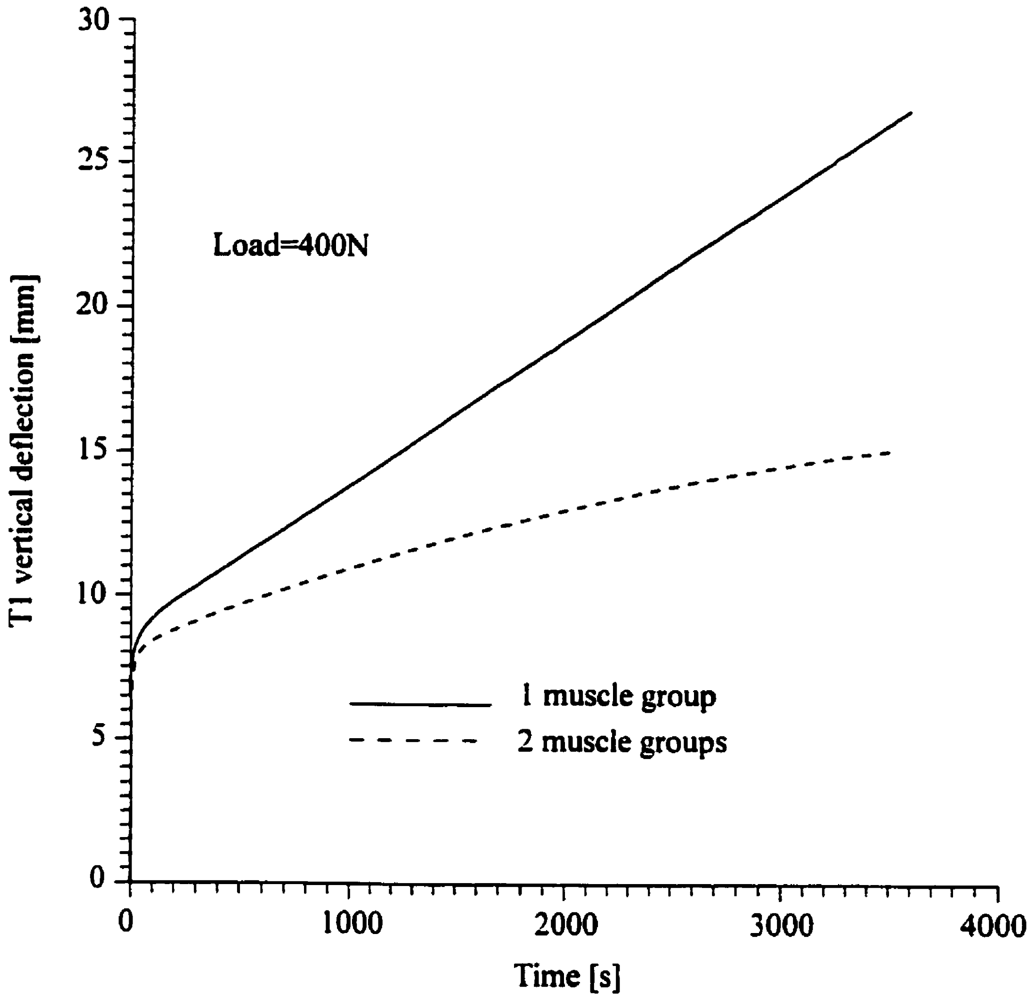

Figure 2.3 Temporal variation of the vertical translation of the T1 vertebra in presence of the muscles

The third publication indicates that during the postural changes with small sagittal movements (±40mm from the initial position at the T1), magnitudes of muscle forces depend on magnitude of the horizontal constraint at the T12 vertebra, adjacent to the thoracolumbar junction. At the same time little sensitivity to the rotational constraints is observed (Table2, Figures 4, 7). The lowest level of the horizontal constraint is determined as a minimal value for this parameter (system is stable only with parameter values higher than minimal) i.e.: some horizontal force resulting from action of the global muscles is always necessary for maintaining of the spinal stability. The highest level of the horizontal constraint allows only very small movement. Thereby, the elastic response of the TLS delimited by these two values for the horizontal constraint represents also limits for the possible deformed configurations of the spine in the neutral posture based on the equilibrium conditions. Figure 2.4. Variations of muscular activities, during the small T1 sagittal movements, follow a close-to-linear pattern apart from a switching point at the T1 initial position. Iliocostalis thoracic and the Internal obliques both on left and right sides are important in anterior T1 positions and Rectus abdominis with External obliques in the posterior positions, Figure 2.5.

Activation of the local muscles in the anterior positions of the T1 proves little efficiency in actuation of the spine. Recovery toward the initial position is accompanied by a disproportionately large compressive penalty (Table2, Figure 8). Efficiency of the local muscles further decreases with an increase of the horizontal constraint at the T12. With high values of the T12 horizontal constraint, the recovery toward the T1 initial position changes to

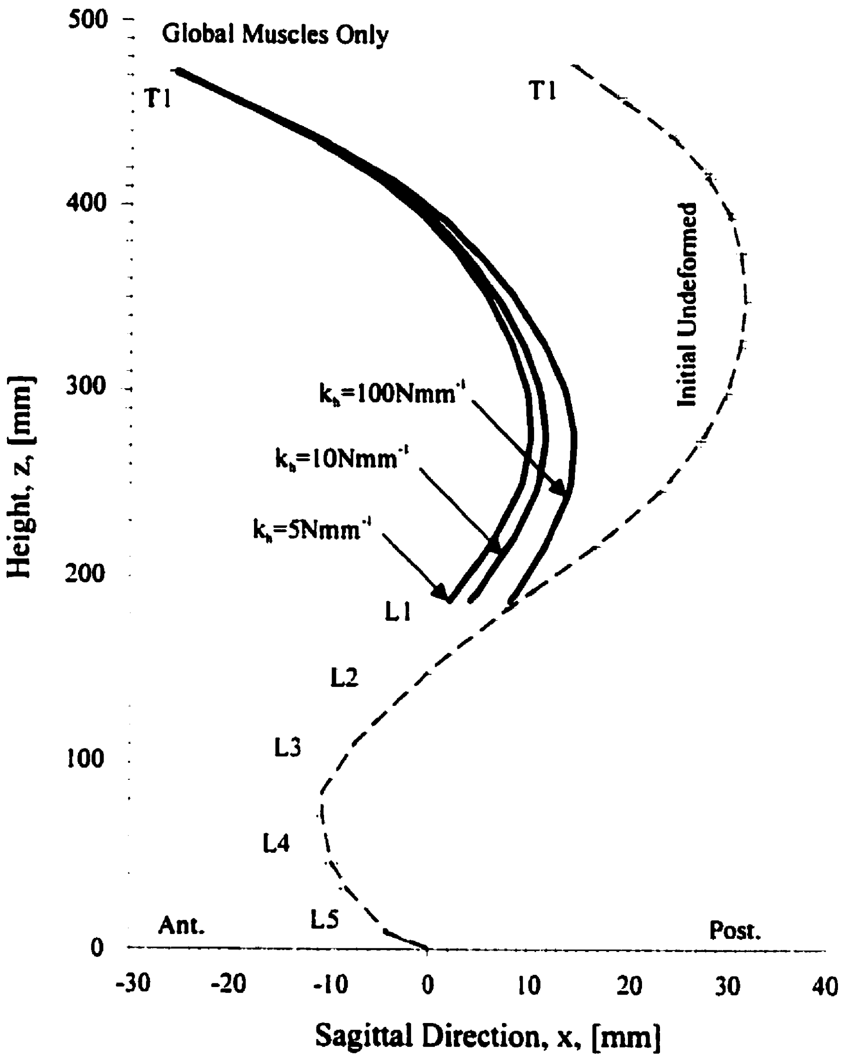

Figure 2.4 Deformed shapes for different levels of the horizontal constraint at the T12 vertebra with the T1 40mm anterior from its inital position in presence of the global muscles only.

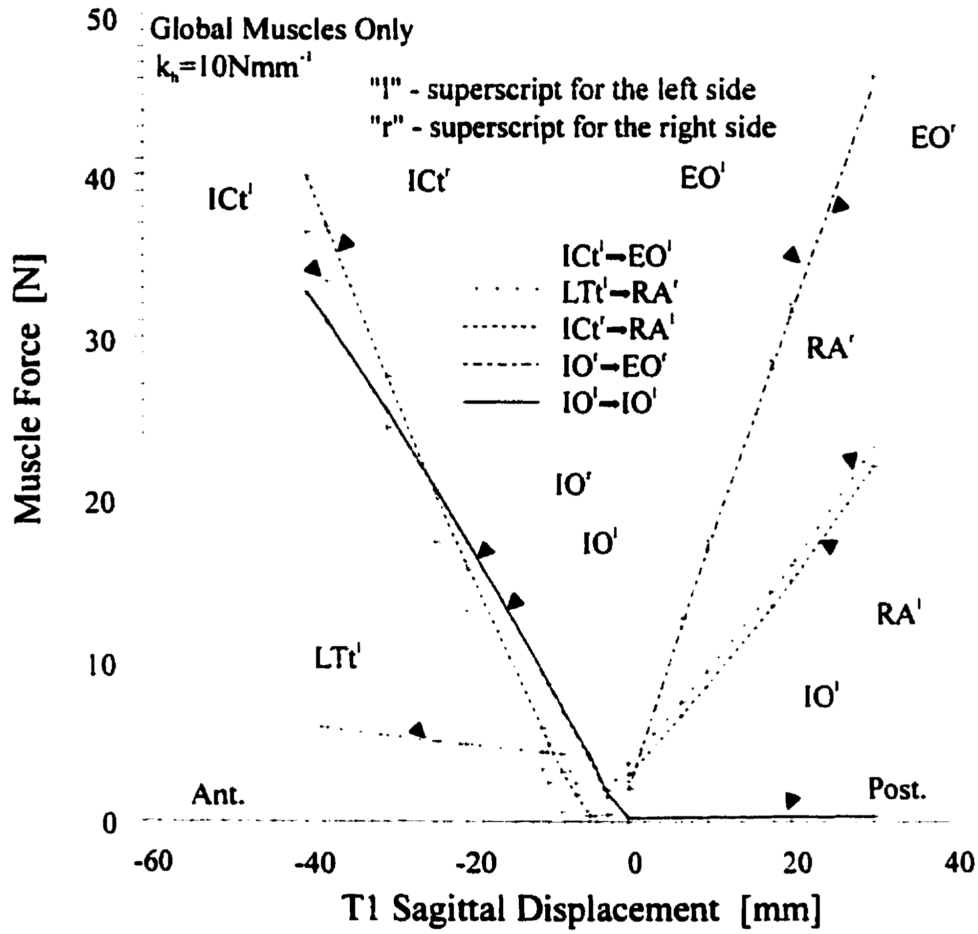

Figure 2.5 Activations in the global muscles for the T1 displacements from 20mm posterior to 40mm anterior from its initial position.

a movement further away from the T1. Among the local muscles, the most preferred are those with largest moment arms.

All three publications indicate a significant involvement of the passive spine in load bearing capacity of the spine. Spinal segments have a good potential in carrying compressive and shear forces and some moments. In the response of the whole passive spine (as an assembly of segments) without cooperation of muscles, only a fraction of potential intersegmental load resistance can be utilized, lest it loses its stability. Aided by the horizontal components or moments from muscles forces (compression resulting from muscle activations being an adverse destabilizing factor comparable to increase of the load), or a pelvic rotation, effects of the physiological load can be controlled and more resistive potential of the segments utilized. In the positions with T1 close to initial neutral position, flexion moments up to 6000Nmm are carried by the passive spine (publication 1, Figures 6, 10) increasing up to 8000Nmm with the creeping disks (publication 2, Figure 7b). An anterior T1 movement can further increase these moments up to 12000Nmm (publication 3, Figure 6).

Assessment of the stability of the predicted equilibrium states is made in publication 3 by the solution of a linear eigenproblem, Figure 2.6, using a simple muscle model described by a single parameter, muscle stiffness coefficient "q", Eq.1.1. Stability of the spine in the neutral posture is affected only slightly by a level of activation in the global muscles. Increased activations of the local muscle prove to significantly effect the overall stability of the spine (Dietrich et al., 1995). At each level of activation of the local muscles, the value of the load added onto the arms before the loss of stability depends linearly on the muscle coefficient "q".

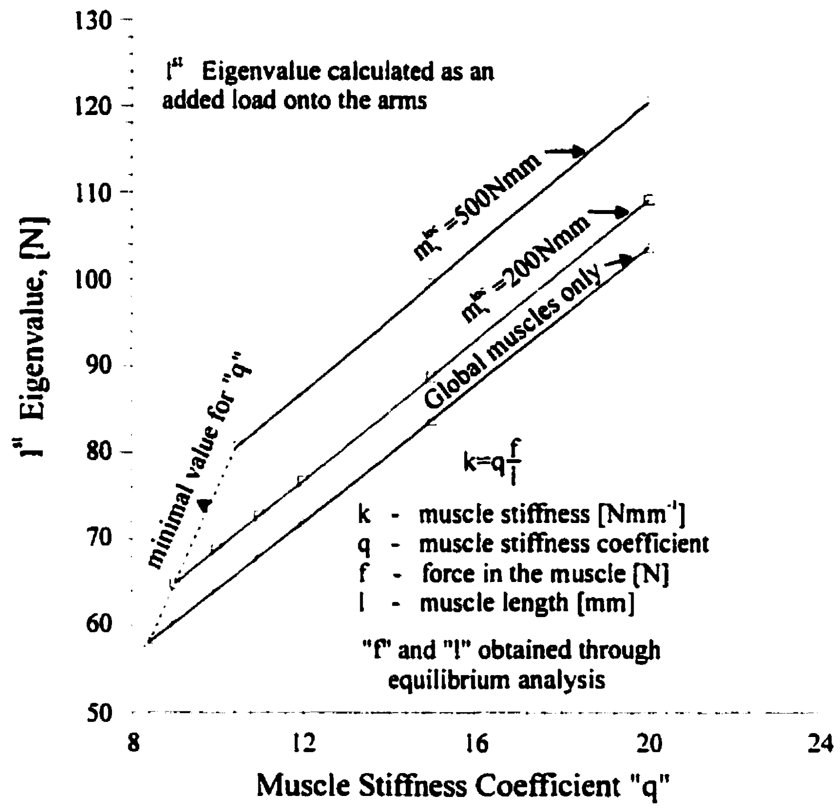

Figure 2.6 Effect of activations of the local muscles on the stability of the thoracolumbar spine evaluated by a linear eigenproblem at a final deformed configuration

The minimum value of "q" increases linearly with an increasing level of activation in the local muscles from 8.5 to 10.5 Nmm^{-1}. The lowest additional load with no activation in the local muscles and muscle stiffness coefficient at its lowest value, 8.4, is 58 N, which is about 15% of the physiological gravity load (380N) or 44% of the compressive load from the muscles (129N) in this particular equilibrium configuration. For the maximal activation of local muscle considered with corresponding minimal muscle coefficient 10.5, the added load is 80N (21% of the physiological load or 58% of the compressive penalty from the muscles). With the increasing level of activation in the local muscles, the proportion of the added load to the compressive penalty from the muscles increases, i.e., stabilizing affect of the local muscles is larger than the adverse effect of the compressive penalty. Thus quantified stability of the spine in neutral posture shows that the equilibrium states predicted for the range of considered activations in the local muscles can withstand additional loads from 6 kg up to 12 kg added onto the arms. Nevertheless, some critical value of muscle stiffness coefficient exists for each level of activation in local muscles, below which equilibrium configurations are unstable.

2.4 DISCUSSION

The proposed method examines and, in the three publications, utilizes equilibrium conditions to determine activations in the muscles and the behavior of the passive spine and subsequently quantifies the effect of some constraint (virtual springs) and a muscle stiffness parameter on its stability. Even with the minimal values of the virtual constraints, equilibrium states of the spine are stable above the critical values of the muscle stiffness coefficient which is around 9. Any analysis of the spinal behavior must conform both with the necessary (equilibrium) and the sufficient (stability) conditions for a proper spinal posture. Interaction between the passive and the active spine based on equilibrium (kinematic conditions) can provide the spine with a sufficient load-bearing capacity in neutral posture.

The diagrams for the distribution of the sagittal moments between the active and the passive components indicate that lordosis is dependent on the loading conditions of the spine. Lordotic curvature is related to the position of the thoracolumbar junction, an important parameter in control of the loads exerted on the lumbar spine from the upper torso. Similarly, lordotic curvature is an important indicator of mobilization of the elastic resistance of the intervertebral disks and spinal ligaments (passive spine) to the applied moments. In the upright posture in gravity conditions, pelvic rotation is an efficient parameter for control of the lordotic curvature.

The hybrid kinematic method allows prediction of both muscle forces and the deformations of the passive tissues. As such it can be classified in a biomechanical terminology to be a type of the forward static optimization, FOS (Allard et al., 1995). The method was not developed a priori from FOS formulation, rather an implementation of the basic synergetic principles governing the spinal behavior within the software limitations, converged to that point. The proposed method creates a framework within which different criteria of muscle activations and passive tissue modeling can be implemented. Relations between the muscle forces and muscle lengths, Figure 2.5, suggest that the muscle forces are related to the change of the spinal configuration by a coordinate transformation. This coordinate transformation is controlled by a shift in the equilibrium setpoint for each muscle, as proposed in the lambda method (Feldman, 1986).

Although considerations of the individual muscle functioning and modeling of the passive tissues are instrumental in creating the present method, its most important prediction, is that of the structural synergy. Selective control of muscles can maintain the passive spine in higher buckling modes, thus leading a neutral posture efficient in terms the body energy expenditure. The complexity of spinal musculature suggests that it has a good potential for control of the transitions between the buckling modes, Appendix E.

CHAPTER 3

DISCUSSION

3.1 HYBRID KINEMATIC METHOD

The presented study aims to develop an approach for investigation of the human spine in neutral posture with static loading. The approach considers interaction between the passive (osteoligamentous) and active (muscular) components of the spine. Several mechanisms feasible for maintaining of the spinal stability in neutral postures without/with presence of the creep in the disks are examined, such as: the realistic distribution of the upper body weight (King-Liu and Wickstrom, 1973; Takashima et al., 1979), rotation at the pelvis

(Gracovetsky et al., 1985; Jackson and McMannus, 1994; Shirazi-Adl and Parnianpour, 1996), initial positioning of the T1 (Jackson and McMannus, 1994; McGlashen et al., 1990). The role of spinal muscles is considered first in a simplified way (Bergmark, 1989), later a more realistic muscle architecture is developed. The model of the passive spine assumes a linear relation between the forces and deformations in the disks, and simulates disks by the beams with reasonable structural properties (Andriacchi et al., 1974; Shirazi-Adl and Parnianpour, 1996; Stokes and Gardner-Morse, 1995; Takashima et al., 1979). Although limited to relatively small intersegmental rotations, the model is both practical to perform a large number of nonlinear analyses coupled with constraint equations and accurate enough to determine the stability response in the upright postures under relatively small axial compressive forces. A large displacement analysis is used to quantify the instability effects. It is to be noted that, due to the lack of bifurcation phenomena, the term instability is used throughout this study to indicate large displacements or hypermobility which occurs due to loss of the global stiffness in axial compression. Moreover, the load-bearing capacity refers to the notion of stability in axial compression (Shirazi-Adl and Parnianpour, 1993; Shirazi-Adl and Parnianpour, 1996) and not the compressive strength of the system. The main advantage of the proposed approach resides in its potential to include a synergetic interaction between the passive and the active spinal components. Nevertheless, its applicability remains limited to the upright neutral posture.

3.2 EFFECT OF THE GEOMETRY AND LOAD CONFIGURATION

Although the load configuration significantly affects the load-bearing capacity of the passive thoracolumbar ligamentous spine, TLS, the magnitude of the total compression load at hypermobility remains for all configurations considered well below the upper body weight. The presence of the optimal pelvic rotations is found to substantially stiffen the TLS during the gradual application of the static physiological load. These rotations, observed also in-vivo in neutral postures, and are affected by the magnitude of the compressive load and spinal geometry (Gracovetsky et al., 1985; Parnianpour et al., 1994). The predicted results also point to a strong dependence of the optimal pelvic rotation on the spinal configuration. The pelvic rotation appears to stabilize the TLS by increasing the lordosis. In the presence of the pelvic rotation under 400N axial load the lordotic angle increased by 6.2 deg (Appendix A). Initial posterior placement of the T1 further increases this trend, while an anterior placement decreased it. The stabilizing role of the pelvic rotation is more evident in these works than in the earlier one (Shirazi-Adl and Parnianpour, 1996).

The optimal pelvic rotation is found to stabilize the spine under physiological loads when creep of the disks is considered. In this case, the magnitude of the pelvic rotation depends significantly on the T1 horizontal position (Appendix B). Results indicate that with a proper T1 positioning the sagittal displacements of the spine due to the creep of the disks can be limited to small values in a presence of an intelligent mechanism which controls rotations without any contribution from muscles. The efficiency of pelvic rotation for the stability of the spine in prolonged neutral postures is related to the horizontal positioning of the thoracolumbar junction at the L1 vertebra.

3.3 EFFECT OF MUSCLES

The muscles supporting the spine are divided into the local (insertions on the lumbar vertebrae) and the global (insertions on the rib cage) groups. The axial load in each muscle is calculated from the spring forces using the optimization procedure followed by an iterative procedure until the convergence was reached. Due to the directions of muscle forces, the stabilizing horizontal forces are accompanied by the adverse vertical forces which increase the axial load on the spine. The sequence of equilibrium states of the spine (a multijoint elastic structure with muscles), is controlled by kinematic constraints modeled as virtual springs (Flash, 1987). The virtual springs simulate the overall effect of the neural system on the control of the spinal synergy. As such, the method allows for a realistic modeling of muscle activities depending on the passive stiffness properties and geometric configuration of the spine.

At first a simplified muscle arrangement was considered, later a more realistic muscle architecture was developed. The compression load-bearing capacity of the TLS substantially increases with the addition of simplified muscles. Relatively small axial forces were generated in muscles at a 400N load, for the global group. The overall horizontal displacements are smaller in the presence of the two muscle groups, although at the expense of increased sagittal moments in the lower lumbar region. In both cases, the initial lordosis increases from 39 deg to about 43 deg. Similar to the optimal pelvic rotation, the geometry of the spine markedly affects the muscle forces required for the spinal stability. That is, there is an optimal initial positioning of the T1 that minimizes the muscle forces (publication 1 Figure 11, publication 3 Figures 2, 5).

When considering effects of the creeping disks, the presence of one muscle group in the model was sufficient only for maintaining of the short term postural stability. Two muscle groups are necessary to maintain postural stability for longer load duration, beyond 15 minutes, (publication 2). The increase in lordosis due to the creep deformations (excluding the instantaneous change of 4deg), in the case of two muscle groups attached at T1 and L1, is 4.6deg.

The realistic muscle architecture (publication 3) includes all major muscles both in local and global muscle groups, however, the intersegmental muscles in the lumbar region are not included, since their moment generating potential in the neutral posture is low (Crisco and Panjabi, 1991). Action of the global muscles only, spanning from the rib cage to pelvis, in conjunction with the passive spinal resistance is found to be sufficient for maintaining of the spinal equilibrium and stability with small sagittal displacements at the T1. The activations in the global system depend on the T1 sagittal position, with the muscle forces increasing linearly (Raschke and Chaffin, 1996) with the distance from the initial T1 position. The recruitment pattern of the global muscles shows coactivation (Hogan. 1984), (publication 3, Figure 5) when in the anterior T1 positions IC and IO, and in posterior T1 positions EO and RA are active both on the left and the right sides. The magnitude of muscle forces in the global system depends also on the degree of horizontal constraint at the T12 (publication 3, Table 2). The postures in which T12 is not allowed to move horizontally with sufficient freedom (publication 3, Figure 3) might result in a significant increase of flexion moments at L1 and L2 levels (publication 3, Figure 4). Activation of the local muscles decreases forces in the global system attached onto the rib cage and provided additional stiffness, thus increasing the overall spinal stability (Dietrich et al., 1990) Fig.11. Majority of activities at all lumbar levels is observed in the Multifidi muscles, some activities (14% of the total force in the local muscles) were observed also in the left Iliocostalis lumborum pars lumborum, LTl, at the L5 level and left Quadratus Lumborum (3.5% of the total force in the local muscles) (Andersson et al., 1996; McGill et al., 1996).

The vertical load from the upper body weight and muscles attached onto the rib cage cause variations of sagittal moments in the lumbar region due to lordotic curvature. Without efficient flexor muscles in the lumbar region, since Psoas major in neutral posture cannot exert the required flexion moments (Bogduk et al., 1992) (publication 3 Table 3), some amount of moment in the lordotic arch appears to be inherently carried by the passive spine, Figure 7.

The muscular activations in equilibrium positions in neutral posture under the postural load exhibit small variations of the recruitment pattern with varying degree of activation in global (different values of horizontal constraint at the T12) and local muscles.

Position of the thoracolumbar junction, controlled by horizontal constraint at the T12, has a marked effect on the distribution of intersegmental rotations of vertebrae in the lumbar lordosis (T12-S1) (Table 2, Figure 5) and thereby also on the magnitude of the sagittal moments carried by the passive spine (Figure 7). The results of the study corroborate qualitatively well with the EMG findings (Parnianpour et al., 1994) and give further insights into spinal mechanics in neutral postures.

The muscle recruitment strategy allows maximum utilization of the elastic resistance of the passive tissues while minimizing the required muscle force to stabilize the torso within the physiological limits of displacements. Results with both simplified and realistic muscle architecture show similar patterns of behavior with distinct effects of the local and the global groups (Bergmark, 1989; Broberg, 1981).

3.4 PELVIC ROTATION AND MUSCLES

Comparison of results, (Figures 7 and 10 publication 1), demonstrates a similar stabilizing influence of the pelvic rotation and the muscles in the spine under axial compression. Both mechanisms are found to increase the initial lordotic angle of the TLS in neutral posture. Interestingly, the lumbar spine in standing postures has been noted to have a larger lordotic angle than in stress-free cadaveric specimens (Adams et al., 1988). Similar trends have been observed in microgravity conditions (Massion et al., 1993) and sideways lying positions

(Keegan, 1951) free of axial loads, where flattening of the spine occurs as compared with the neutral standing position under gravity loads. This lordotic posture, although a function of the load and the T1 positioning, likely enhances the compression stability of the spine.

3.5 CONCLUSIONS

Experimental studies with surface electrodes provide EMG information from a volume below the electrode and do not differentiate between individual muscle fascicles. This can be overcome by using invasive wire electrodes implanted into discrete muscle locations. For both non-invasive and invasive approaches, however, there exists no exact relation between EMG signal and muscle force. The present method evaluates muscle forces in the individual muscle fascicles from kinematic and optimization criteria, with a full interaction between the passive spine and muscles (Figure 2.1).

The predictions of the recruitment pattern and magnitudes of muscle forces in this study are dependent upon the cost function requiring that compressive load from muscles be minimal at each lumbar level. Nevertheless, choice of a different criterion for solution of a redundant muscle problem in conditions of the neutral posture should not lead to a significant augmentation of the reported muscle activities, since they provide a sufficient degree of stability (publication 3).

When considering the effect of creeping disks, the muscles together with the pelvic rotation allow for an efficient control of the spinal response in creep. The flexion load-bearing potential of the passive spinal structures, TLS, mobilized through changes in spinal curvatures (i.e., kyphosis and lordosis), becomes also important in the control of the spinal stability. The nonlinear response of the spine in neutral posture subjected to creep deformations indicated that the passive TLS and the spinal muscles should not be treated as isolated systems. The effects of a prolonged load can be highly damaging when no appropriate adjustment of spinal geometry is allowed. The behavior of the spine in semi-constrained situations, such as seated position, can significantly limit the mechanism for an effective control of the spinal stability, followed by an adverse compensatory response to stabilize the system (publication 2).

Findings of this work indicate that small muscle activations and pelvic rotation fully exploit passive load-bearing potential of the TLS (Gracovetsky et al., 1985; Jackson and McMannus, 1994) by controlling its deformation modes. The trunk in free standing posture under physiological gravity loads exploits the TLS double curvature and decreases its effective buckling length by the action of muscles and pelvic rotation, thereby, increasing its compression load-bearing capacity while decreasing its horizontal displacements. To move into and maintain higher modes of deformation, only relatively low muscular forces are required when applied at critical regions along the height of the TLS (i.e., the inflection points). In contrast to the strategy for stabilization of upper and lower extremity joints with presence of high coactivations, only moderate coactivations are predicted in the trunk upright posture(Parnianpour et al., 1994), indicating that a more intelligent control strategy can be used for the stabilization of the spine. The actions of muscles and pelvic rotation are postulated to be coordinated by a neural controller with the horizontal translation at the T1 as a likely feedback parameter. The results suggest that the pelvic rotation, muscle activations, and the off-center placement of the line of gravity are exploited to stabilize the passive spinal system in the neutral postures. The evaluation of the stability of the spine at higher exertions and more complex loading conditions deserves a separate analysis.

FUTURE DEVELOPMENTS

The presented hybrid kinematic (modified static forward optimization) method develops an approach incorporating the passive components of the human spine and muscles. In its present form, the model is limited to static loading conditions in the neutral posture and calculates muscle activations from the equilibrium conditions of the spine. Similarly, the muscle model utilized is limited to the static equilibrium only. The muscle architecture is realistic enough, nevertheless, intersegmental muscles are not included. Elastic behavior of the vertebral segments is considered linear and is modeled by the beams. The rib cage is simulated by a rigid body. The interindividual variations, age or any forms of defective anatomy were not considered.

The abovementioned assumptions are made to achieve a good computational efficiency and are not inherent to the method. Predicted patterns of muscle activations, are dependent on the assumption of the cost function and represent only minimal necessary degree of coactivation, since they are based on equilibrium conditions. Since the neutral posture is considered only under a static physiological gravity load sustained in an isometric muscle state, any conclusions drawn from this study are relevant only in the specified conditions. Effect of interindividual variations of the spinal geometry and properties in the healthy individuals deserve a separate study.

Incorporation of nonlinear vertebral segment properties would allow investigation of more general postures and loadings, other than the neutral posture. This can be achieved by the use of the nonlinear lumped springs for the disks and ligaments, or by modeling detailed spinal segments including the contact problem. Development of the more detailed passive structures with complex material behavior is limited only by capacities of the ABAQUS structural analysis software.

Using the ABAQUS structural analysis software, the method can be extended also into dynamic response, nevertheless, this would require also more sophisticated modeling of the muscle behavior. This can be conveniently accommodated by the flexibility of the in-house built active module.

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Appendix A

Publication 1

On the stability of the human spine in neutral postures

Appendix B

Publication 2

Creep stability of the human spine in neutral postures

Appendix C

Publication 3

Synergy of the human spine in neutral postures

Appendix D

Coordinates for the spinal geometry

Appendix E

Beam element in compression

APPENDIX E

BEAM ELEMENT IN COMPRESSION

Beam elements were used in this work to simulate behavior of the intervertebral disks. Abaqus element libraries provide two formulations of beam elements: *Timoshenko* and *Euler-Bernoulli beams.* The Timoshenko beam is shear-flexible and allows for transverse shear deformations, Euler-Bernoulli beam does not.

Euler-Bernoulli beam, B33, is applicable to the cases where beam's cross-sectional dimensions compared to typical distances along its axis are small, the beam is slender. For beams made of uniform material typical dimensions in a cross section should be less than about 1/15 of the typical axial distances. The beam uses a cubic interpolation function.

Timoshenko beam element, B31, allows for transverse shear and deformation. They can be used for thick (stout) as well for slender beams. For beams made from uniform material, shear flexible beam theory can provide useful results for cross sectional dimensions up to 1/10 of typical axial distances. Beyond this, the approximations that allow the member's behavior to be described solely as a function of axial position, no longer provide adequate accuracy. The beam uses a linear approximation function.

The two beams were tested in a pure compression under the concentrated load up to 2000N. The properties of the beams were selected to represent a typical lumbar intervertebral disk and are shown in Figure 1. Initially, the two beams deflect almost linearly, with a slope $k=1050Nmm^{-1}$, up to the load 500N. With an increasing load, the stiffness of the B33 beam decreases, at the load 2000N becoming only 17% of its initial value. The B31 beam stiffness with the increasing load slightly increases up to 110% of its initial value at 2000N. This loading test indicates that *Euler-Bernoulli beam* is inappropriate to represent the intervertebral disk at loads higher than 500N. The *Timoshenko beam* behavior is slightly nonlinear, since it uses the real strains.

The test shows, that *Timoshenko beam* provides more accurate results for the intervertebral disk like beams in a large deformation analysis at loads more than 500N in a pure compression. The beam representing the intervertebral disk is relatively stout with a slenderness ratio around 4, with significant shear effects. When the compressive load is higher than 500N, the linear *Timoshenko beam* B31 is more appropriate for modeling of the intervertebral disks than the cubic *Euler-Bernoulli* beam B33.

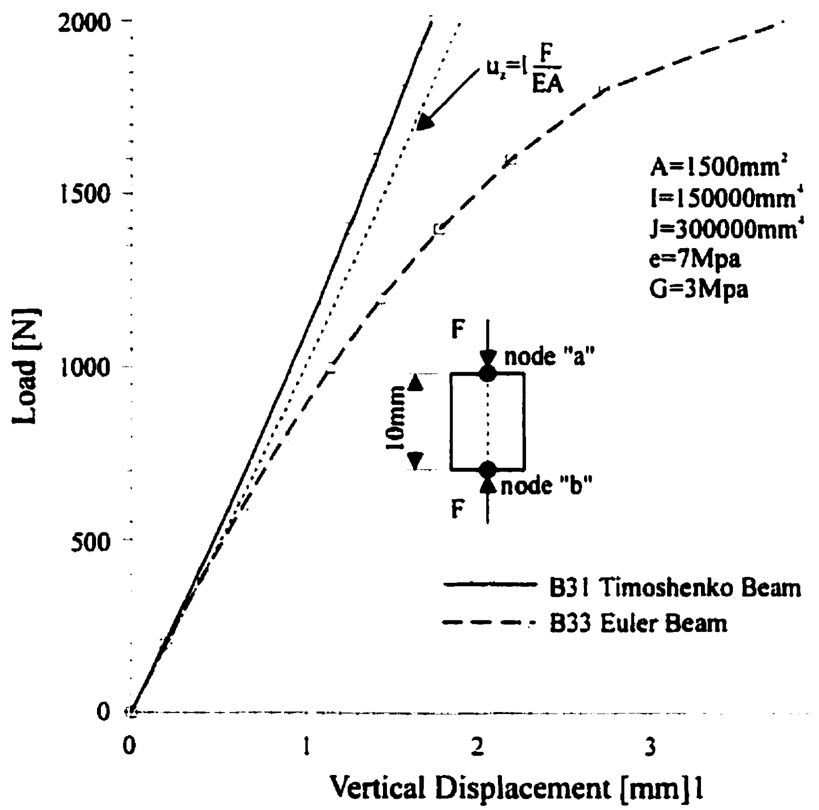


Figure 1 Compression test for Abaqus beam elements

Appendix F

Adaptive control of a flexible column

APPENDIX F

ADAPTIVE CONTROL OF A FLEXIBLE COLUMN

The buckling behavior becomes important in a case of the flexible columns with an adaptive control. A finite element analysis, using Abaqus structural analysis software, was performed to illustrate some mechanisms affecting the behavior of a flexible column with an adaptive control. The mesh consisted of ten 2 node linear *Timoshenko beam* elements B31 from the Abaqus element library. Large displacement analysis was used to quantify the load effects with the changing geometric configuration. The dimensions and properties of the column were selected to approximately represent the flexible vertebral column, Figure 1.6. An initial imperfection, with an amplitude less than 0.25% of the column's height, was calculated as a combination of the first two buckling modes, φ_1 and φ_2. Several proportions of modal combination were tested, e.g., $0.8\varphi_1+0.2\varphi_2$, $0.5\varphi_1+0.5\varphi_2$, $0.1\varphi_1+0.99\varphi_2$, $0.05\varphi_1+0.95\varphi_2$, with no appreciable effect on the column's behavior. The loading test was designed to demonstrate the transition between the first and second buckling modes in two loading steps, LDS, as:

LDS1 the column with a vertical slider at the top and a hinge at the bottom was loaded by a vertical concentrated load, P, at the top, close to the first buckling load, P_1^{crit},

LDS2 in the configuration under the load close to P_1^{crit} the top slider was replaced by a hinge with a reaction R_v, and the midpoint, M, of the column was dragged toward its initial position with the horizontal restoring force, R_h , and the vertical reaction, R_v, monitored.

The results in terms of the column's displacements and forces are shown in Fig.1. The two extreme deformed shapes with corresponding loads correspond to the buckling modes enforced by the boundary conditions. In the LS1, the vertical reaction reaches 99% of the first buckling load with a horizontal displacement around 15mm at the column's midpoint. The column is left to displace further up to 40mm horizontally at the midpoint M. In the LS2 the horizontal restorative force, R_h, reaches the maximum in the point T, with a value close to the first buckling load, P_1^{crit}, with the midpoint around 25% on the way toward its initial position. In the point T the vertical reaction, R_v, reaches a value close to 90% of the P_2^{crit}. The behavior in the LS2 can be then divided into -region 1- with M in position 30-40mm and -region 2- with M in positions 30-0mm from the initial position.

Let's now assume that some controller, such as muscle operates on the present column at the point M to preserve its stability. The stiffness of a muscle depends on a force it exerts. In a simplified form, the stiffness of a muscle can be expressed as:

$$k = q\frac{S}{l} \tag{1}$$

Where k is the current muscle stiffness, S is the muscle force, q the experimentally derived coefficient and l the muscle length.

The stability of the column in the second mode of deformation under the load close to P_2^{crit} is maintained by the stiffness of the controller at the midpoint. With horizontal displacements of the midpoint M inside -region 1- the restorative force R_h, determined from the equilibrium conditions of the system, increases with the movement of the point away from the initial position. Increasing force causes also increasing stiffness of the controller at the point M and the stability of the column can be maintained. In the -region 2-, the restorative force R_h decreases with the movement of the point M away from its initial position, thereby causing a decrease of the stiffness in the controller. When the stiffness of the controller decreases, it is not able to maintain the stability of the system anymore, and column transfers into the first mode and collapses under the applied compressive load.

It can be concluded, for the column with a controller described in Eq.(1), that it can safely operate in the -region 1-, with the restorative force R_h providing a sufficient stiffness for the controller. In the -region 2- the restorative force R_h is insufficient to provide a required stiffness for the controller, and equilibrium of the system is unstable.

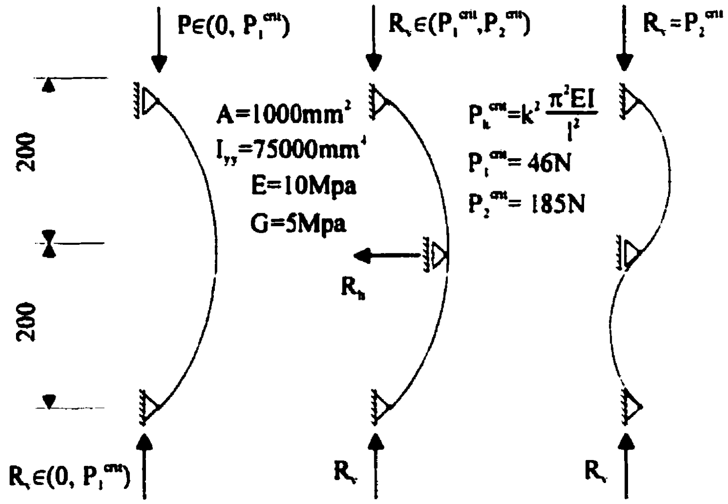


Figure 1 Test for a modal transition of a column

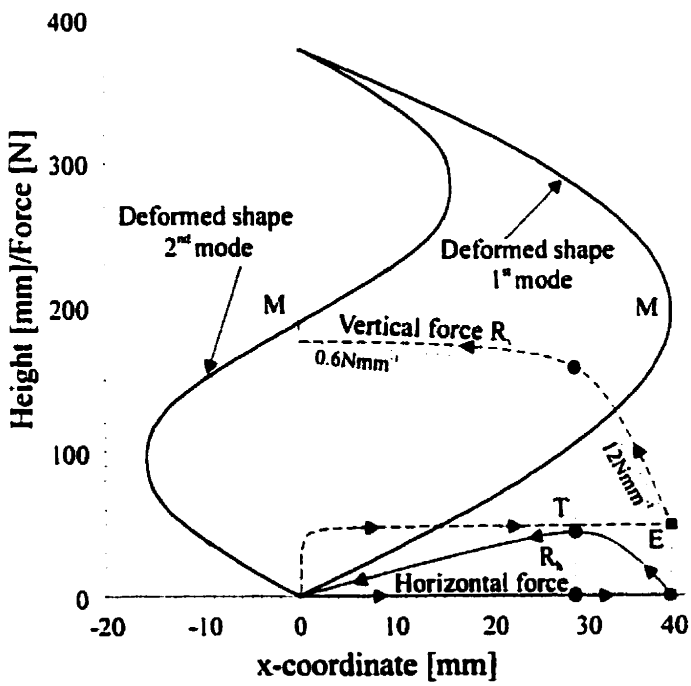


Figure 2 Equilibrium paths and displacements for a modal transition of a column

Appendix G

Hybrid kinematic method - details

APPENDIX G

METHOD

Equilibrium formulation

The passive, osteoligamentous spinal column can be described structurally as a vertical column, with the sagittal deviations from the vertical line less than 8%. This column, when devoid of muscles can sustain only minimal loads, nevertheless when functioning in synergy with muscles its resistance increases multifold. The muscles stabilize (support) the passive spine in the horizontal plane and decrease its buckling length by forces acting at the critical points. Selective activation of muscles in the critical buckling points represents an economic method for increase of the spinal load-bearing capacity. The forces from loads on the trunk resulting from loads (gravity, external) can be partitioned between the passive spine and muscles (active components) as:

$$\begin{aligned} &\{F^l\}=\{F^p\}+\{F^m\} \rightarrow \{F^p\}=\{F^l\}-\{F^m\} \\ &\{F^m\}=\{F^c\}-\{\Delta P\} \end{aligned} \tag{1}$$

The loads acting on a spinal segment, $\{F^l\}$, are resisted by forces generated by the passive tissues, $\{F^p\}$, and forces generated by the muscles $\{F^m\}$. The forces carried by the passive tissues become a difference between the external load and forces generated by the muscles. Forces in the muscles can be further subdivided into the stabilizing constraint forces generated by the springs,

$\{F^c\}$ and the compressive penalty, $\{\Delta P\}$. The matrices for relations between the external load muscle forces and forces carried by the passive tissues, $\{F^l\}$,$\{F^p\}$,$\{F^p\}$ and $\{\Delta P\}$ can be developed according to Eq.(1) as:

$$\begin{Bmatrix} F_x^m \\ F_y^m \\ F_z^m \\ M_x^m \\ M_y^m \\ M_z^m \end{Bmatrix} = \begin{Bmatrix} F_x^c \\ F_y^c \\ -\Delta P \\ M_x^c \\ M_y^c \\ M_z^c \end{Bmatrix} \qquad \begin{Bmatrix} F_x^p \\ F_y^p \\ F_z^p \\ M_x^p \\ M_y^p \\ M_z^p \end{Bmatrix} = \begin{Bmatrix} F_x^l \\ F_y^l \\ F_z^l \\ M_x^l \\ M_y^l \\ M_z^l \end{Bmatrix} - \begin{Bmatrix} F_x^m \\ F_y^m \\ F_z^m \\ M_x^m \\ M_y^m \\ M_z^m \end{Bmatrix} \tag{2}$$

The forces and displacements are related by their respective stiffness matrices:

$$\begin{aligned} \{F^p\} &= [K^p]\{u\} \\ \{F^c\} &= [K^c]\{u\} \end{aligned} \tag{3}$$

where [Kp] is a matrix of the stiffness coefficients for the beam representing the disk, [Kc] is a matrix of the constraint spring stiffnesses and {u} is a vector of the displacements. Equations (1) and (2) can be combined to obtain:

$$\begin{aligned} \{F^l\} - \{F^c\} + \{\Delta P\} &= [K^p]\{u\} \\ \{F^l\} + \{\Delta P\} &= ([K^p] + [K^c])\{u\} \\ \{\bar{F}\} &= [\bar{K}]\{u\} \end{aligned} \tag{4}$$

where loads acting on a vertebral segment and compressive penalty forms the vector $\{\bar{F}\}$ and passive disk stiffness and constraint forces form the matrix $[\bar{K}]$. Equation (3) can be developed as:

$$\begin{Bmatrix} F_x^{\,l}-k_{11}^{c}\,u_1 \\ F_y^{\,l}-k_{22}^{c}\,u_2 \\ F_z^{\,l}+\Delta P \\ M_x^{\,l}-k_{44}^{c}\,u_4 \\ M_y^{\,l}-k_{55}^{c}\,u_5 \\ M_z^{\,l}-k_{66}^{c}\,u_6 \end{Bmatrix} = [K^{p}] \begin{Bmatrix} u_1 \\ u_2 \\ u_3 \\ u_4 \\ u_5 \\ u_6 \end{Bmatrix} \tag{5}$$

and further as:

$$\begin{Bmatrix} F_x^{\,l} \\ F_y^{\,l} \\ F_z^{\,l}+\Delta P \\ M_x^{\,l} \\ M_y^{\,l} \\ M_z^{\,l} \end{Bmatrix} = \begin{bmatrix} k_{11}+k_{11}^{c} & k_{12} & k_{13} & k_{14} & k_{15} & k_{16} \\ k_{21} & k_{22}+k_{22}^{c} & k_{23} & k_{24} & k_{25} & k_{26} \\ k_{31} & k_{32} & k_{33} & k_{34} & k_{35} & k_{36} \\ k_{41} & k_{42} & k_{43} & k_{44}+k_{44}^{c} & k_{45} & k_{46} \\ k_{51} & k_{52} & k_{53} & k_{54} & k_{55}+k_{55}^{c} & k_{56} \\ k_{61} & k_{62} & k_{63} & k_{64} & k_{65} & k_{66}+k_{66}^{c} \end{bmatrix} \begin{Bmatrix} u_1 \\ u_2 \\ u_3 \\ u_4 \\ u_5 \\ u_6 \end{Bmatrix} \tag{6}$$

The muscle problem (calculation of muscle forces) with the given constraint forces can be then formulated as:

$$\begin{gathered} f^{c}_{i}=a_{ij}s_j \\ \{F^{c}\}=[A]\,\{S\} \end{gathered} \tag{7}$$

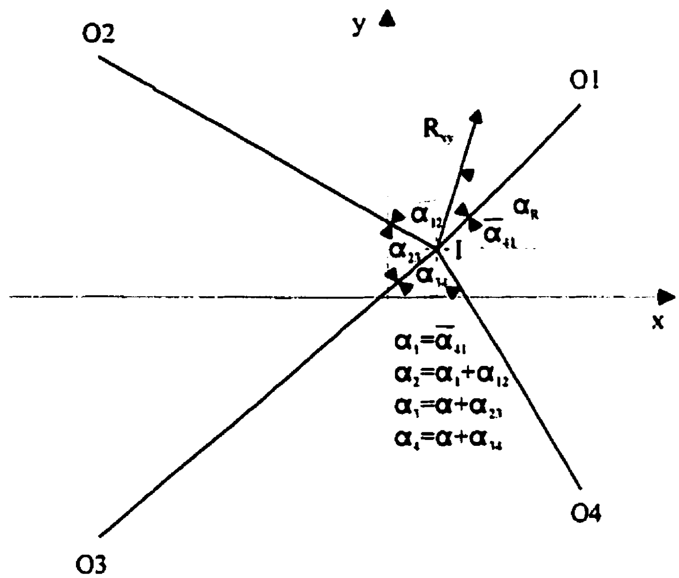
where F^c are the constraint forces which are a function of displacements at a disk level and then further a function of the muscle forces {S}. The magnitudes of the muscle forces {S} are related to the constraint forces F^c by a matrix of geometric transformations [A]. The compressive penalty at the particular disk level can be calculated as:

$$\begin{aligned} &\Delta P = s_i b_i \quad \textit{where} \quad b_i = \cos(n_i, z) \\ &\Delta P = \{S\}\{B\} \end{aligned} \tag{8}$$

The compressive penalty ΔP is a function of muscle forces {S} and coefficients {b}.

Spinal model with a simplified muscle architecture

The constraints are chosen as horizontal supports (springs) at the T1 and/or L1 corresponding to activities in the global and/or local system. In the case with support at the T1 only, choice of the stiffness for the horizontal springs does not affect the values of the muscle forces as long as the stability of the whole structure is preserved. In the case with the two supports at the T1 and L1 the effect of horizontal constraint spring is also negligible. The solution of the muscle problem is obtained by a statically determinate method, Figure 1, by resolution of the horizontal constraint force into the two out of the four possible directions. The criterion for the selection of the two directions of the muscular action requires that the muscles are in tension only, thereby, the



Angle range	Activated muscles
$\alpha_4<\alpha_R<360$	4-1
or $0<\alpha_R<\alpha_1$	4-1
$\alpha_1<\alpha_R<\alpha_2$	1-2
$\alpha_2<\alpha_R<\alpha_3$	2-3
$\alpha_3<\alpha_R<\alpha_4$	3-4

Figure 1 Force resolution for the simplified muscle model

two muscles can be selected according to the sector into which falls the direction of the constraint force, Figure 1. The equation (7) can be than written:

$$\begin{Bmatrix} f_1^{\,c} \\ f_2^{\,c} \end{Bmatrix} = \begin{bmatrix} a_{11} & a_{12} \\ a_{21} & a_{22} \end{bmatrix} \begin{Bmatrix} s_1 \\ s_2 \end{Bmatrix} \rightarrow \begin{Bmatrix} s_1 \\ s_2 \end{Bmatrix} = \frac{1}{a_{12}{}^2 - a_{11}a_{22}} \begin{bmatrix} -a_{22} & a_{12} \\ a_{21} & -a_{11} \end{bmatrix} \begin{Bmatrix} f_1^{\,c} \\ f_2^{\,c} \end{Bmatrix} \tag{9}$$

where a_{ij} are the direction cosines with first index belonging to the direction of the muscle force "s" and second to the global "x" and "y" axes, Figure 2.1. The compressive penalty can be obtained from Eq.(8). The computational procedure is iterated (Figure 3, appendix A), with the compressive penalty as a feedback parameter, until convergence is reached. In the case of the two constrained points, each is treated in a same way, and interaction between the two muscle groups is achieved by transfer of forces through the passive structure.

Spinal model with a realistic muscle architecture

The model with a simplified muscle architecture allows for an expansion by incorporation of a realistic muscle architecture. The refined model preserves all basic concepts developed in the simplified model. The accommodation of more complex muscle architecture was achieved by:

(i) maximizing the number of the control points in the lumbar region,

(ii increasing the number of constraints at each control point up to five.

For the lumbar region, one control point was assigned to each vertebra, to allow for monitoring of the relative displacements of all five spinal segments. Offsetting the insertion points on the vertebra allowed to add constraints for rotations of the vertebra around the three global axes, in addition to displacements in the horizontal direction. The number of constraints at each constraint point thus increased to five, each constraint representing an equilibrium condition, Eq.(7). The torso was treated as a rigid body, thereby allowing for allocation of only one control point with five constraints. Since the number of muscle fibers inserting into each vertebra and torso outnumbered the constraints, Eq. (7) could be not solved directly and optimization procedure had to be utilized to obtain magnitudes of the muscle forces. The linear optimization problem can be formulated as:

$$\begin{Bmatrix} F_x^c \\ F_y^c \\ M_x^c \\ M_y^c \\ M_z^c \end{Bmatrix} = \begin{Bmatrix} k_{11}^c u_1 \\ k_{22}^c u_2 \\ k_{44}^c u_4 \\ k_{55}^c u_5 \\ k_{66}^c u_6 \end{Bmatrix} = \begin{bmatrix} a_{11} & a_{12} & \cdot & \cdot & a_{1k} \\ a_{21} & a_{22} & \cdot & \cdot & a_{2k} \\ a_{31} & a_{32} & \cdot & \cdot & a_{3k} \\ a_{41} & a_{42} & \cdot & \cdot & a_{4k} \\ a_{51} & a_{52} & \cdot & \cdot & a_{5k} \end{bmatrix} \begin{Bmatrix} s_1 \\ s_2 \\ \cdot \\ \cdot \\ s_k \end{Bmatrix} \tag{10}$$

$$
\begin{aligned}
a_{1i} &= \cos(x, n_i) = \cos\alpha_{1i} \\
a_{2i} &= \cos(y, n_i) = \cos\alpha_{2i} \\
a_{3i} &= -\cos\alpha_{3i} \cdot r_y + \cos\alpha_{2i} \cdot r_z \\
a_{4i} &= \cos\alpha_{1i} \cdot r_z - \cos\alpha_{3i} \cdot r_x \\
a_{5i} &= -\cos\alpha_{2i} \cdot r_x + \cos\alpha_{1i} \cdot r_y \\
b_i &= \cos(z, n_i) = \cos\alpha_{3i}
\end{aligned}
\tag{11}
$$

where a_{ij} is a coefficient transforming force from the individual muscle fascicles into forces acting at a disk level, n_i is a direction of the "i^{th}" the muscle.

Appendix H

Frequently asked questions (FAQ)

APPENDIX H

FREQUENTLY ASKED QUESTIONS, FAQ

Q: How are the rotational and translational conditions for the equilibrium decoupled in the work of Ladin et al., (1989) in a model without any elastic components an how does this relate to the work of Stokes with a deformable spine included?

A: In a pioneer work by Stokes (1995) it is stated:"Rather than attempting to simulate particular in-vivo loading conditions, the model was used to analyze for moments individually applied at the vertebral center of the T12 in the thorax". In this model also a small deformation analysis is associated with the sagittal intersegmental rotations ±5 deg and the translations ±5mm (this being relatively large). Small displacement analysis is unable to simulate hypermobility (coupling between the forces and the changing geometric configuration) related to the buckling behaviour. Likewise, the analysis is evaluated in terms of maximal moment resisting potential rather than the maximal compressive load.

Q: Why is synergy restricted to the neutral posture?

A: The spine in the neutral posture is essentially a vertical S-shaped column, carrying a compressive load with the load bearing capacity enhanced by the moments. Structurally, the problem is idealized as buckling of a column with an initial S-shape imperfections less than 8% of its height, with the multiple supports. In a Fullerian (Fuller 1975) sense, any synergetic machine can be resolved into a tensegrity. For the spine, this can be interpreted in a following way: as long as the behavior is predominantly buckling caused by the compressive load, associated with the buckling modes, synergy is predominant. In the postures away from the neutral, where the spine behaves as a beam, the response is not governed by the compressive state, nor is a modal one.

Q: Where does the number m_y^{loc}=500Nmm come from?

A: The number is an assumption based on the parametric study. The assumed activation represents an overall level of activation of the local muscles. The purpose of activating the local muscles is to observe their effect on the spinal behavior in qualitative terms. Assumption of the overall activation of the muscles is of the same order as the assumption of the same stiffness coefficient for the muscles. (Bergmark, 1989; Crisco et al., 1991; Stokes et al., 1995). Bergmark in his work assumes an overall degree of activation for some groups of muscles. The qualitative conclusion, similarly to Dietrich et al., (1990) is that the local muscles in the presence of the global muscles serve mainly to increase the stability of the spine.

Q: How are the virtual spring related to the spinal geometry?

A: The virtual springs (constraints), provide for a feedback from the muscles toward the geometry and vice versa. That is also a reason why the analysis must be performed in an iterative manner. The initial step determines the muscle forces without interaction the muscles and the passive spine. The subsequent steps incorporate effect of the muscles on the passive spine through their compressive penalty loads and effect of the changing configuration on the external load distribution and muscle insertion points. The iteration is repeated until the muscle forces and the geometry converge (the equilibrium state is achieved).

The effect of the changing virtual spring stiffness on the global muscle is examined in a parametric study for the most significant of the constraint stiffnesses k_h and a set of spinal geometries is obtained. The variation of the k_h parameter has a significant effect on the spinal configuration, nevertheless, only a limited effect on the spinal stability. For the local muscles, virtual spring stiffness $(k_y^{loc})_i$ for each lumbar level "i", correspomding to each leval of activation m_y^{loc} can be found as:

$$(k_y^{loc})_i = \frac{(m_y^{loc})_i}{(\phi_y^{loc})_i} \qquad (1)$$

where $(\varphi_y^{loc})_i$ is a flexion-extension rotation calculated from the initial undeformed state. It has been observed that with the $(m_y{}^{loc})_i$ increasing, the $(\varphi_y{}^{loc})_i$ decreases, and this indicates that the stiffness $(k_y{}^{loc})_i$ has a tendency to increase with the increasing activation of the local muscles. The effect of $(k_y{}^{loc})_i$ or $(m_y{}^{loc})_i$ has a major effect of the stability of the spine. The effect of $(k_y{}^{loc})_i$ or $(m_y{}^{loc})_i$ on the spinal configuration decreases with increasing stiffness k_h for the global muscles. (Table 2, appendix C).

Q: To resolve the issue of coactivation for a particular joint, one must first find the minimum activation necessary to balance the external loads and the activations above this level can be considered as coactivation. Coactivation can be further quantified in terms of the additional compressive load acting on the joint.

A: In the present study, the effect of the external loads, F^l, on the spine depends on the spinal configuration, u, which in turn affects forces at the disks, F^c, and muscle forces F^m. The mutual interdependence of all parameters can be expressed as:

$$f_m = \psi(f^l, u, f^c, f^m) \qquad \textbf{(2)}$$

To provide for a feedback between the individual parameters, the analysis is preformed iteratively. If a minimum activation for some configuration could be determined, any additional recruitment (change in muscle forces) would result in change of the incumbent configuration and the reference geometry for an activation-coactivation phenomenon. This exemplifies limitation of a criteria developed for a single joint with rigid structural components which is not easily extendable into a multijoint flexible links such as the human spine.

A plausible alternative for the quantification of the coactivation is the system stability. The level of the minimal activation corresponds to the minimal required stability. Additional muscle recruitment above this level would increase the stability, thereby the system stability becomes an indicator of the coactivation. Figure 2.6 quantifies stability of the spinal system and indicates also the stability in the abovementioned sense. Interestingly, there can be no equilibrium below certain values of the "q. This raises a question about the real physiological meaning of coactivation - as a concept it should be dependent only on the activations (forces) in the muscles, but in the present model, some additional stiffness in the muscles is always needed to satisfy the minimum stability condition.

Q: Not considering the muscle cross-sectional areas, while using the optimization model minimizing the compression will lead to activation of muscles with a greatest moment arm. How does this affect the conclusions regarding the local and global muscles?

A: The forces in the muscles in the present study are less than 5% of the maximum potential of any of the considered muscles. In the optimization models, cross sectional area is used to indicate when a particular muscle's capacity is exhausted, than an additional one is mobilized without increasing the order of the muscle problem. In the optimization scheme, the muscle area becomes important only when forces in the muscles are close to their maximal values.

Q: How are the slopes of the curves "muscle force versus T1 position" related to the muscle moment arms?

A: The muscle forces f^m_i are related to the relevant constraint forces f^c_i at a particular lumbar level by a corresponding force transformation matrix [A] (Eqs 8-12 and Figure 1 in the Appendix F), containing the muscle lever arms and direction cosines. Depending on the constraint forces, f^c_i, muscles are selected through the optimization process according to their advantage in carrying the constraint moments and the constraint horizontal forces. Muscles LT and IC at the T12 level in Figure 2.5 have similar sagittal lever arms, nevertheless, slopes of their equilibrium paths are different. The opposite is true for IC and IO, lever arms are different, but slopes are similar.