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Synergy of the human spine in neutral postures

Abstract The neutral position of the spine is the posture most commonly sustained throughout daily activities. Previous investigations of the spine focused mainly on maximal exertions in various symmetric and asymmetric postures. This report proposes a new synergetic approach for analysis of the spine in neutral postures and evaluates its performance. The model consists of passive components, the osteoligamentous spine, and active components, the spinal muscles. The muscle architecture includes 60 muscles inserting onto both the rib cage and lumbar vertebral bodies. The passive spine is simulated by a finite element model, while kinematic constraints and optimization are used for resolution of a redundant muscle recruitment problem. Although the passive spine alone exhibits little resistance to a vertical load, its load-bearing capacity in neutral posture is significantly enhanced by the muscles, i.e., the passive spine and its muscles must be considered as a synergetic system. The proposed method is used to investigate the response of the spine when the T1 vertebra displaces 40 mm anteriorly and 20 mm posteriorly from its initial position. The sacrum is fixed at all times and the T1 displacements are achieved by the action of muscles. The results suggest that relatively small muscle activations are sufficient to stabilize the spine in neutral posture under the body weight. The results also indicate that muscles attaching onto the rib cage are important for control of the overall spinal posture and maintenance of equilibrium. The muscles inserting onto the lumbar vertebrae are found mainly to enhance the stability of the spine. The proposed method also predicts forces and moments carried by the passive system. Flexion moments ranging from 8000 Nmm to 15,000 Nmm, corresponding to decreases in lordosis of 6° and 7.5° respectively, are found to be carried by the passive spine at the thoracolumbar junction when the T1 vertebra is 40 mm anterior to its initial position.

Key words Synergy · Passive spine · Muscles · Equilibrium · Stability

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Introduction

An understanding of mechanisms for the load-resisting capacity of the human spine and load sharing between the passive ligamentous spine and the active muscle tissues in neutral posture is essential for investigation of spinal functioning in normal and pathologic conditions. Several approaches have been taken in previous numerical models for static analysis of the lumbar spine: maximum moment-generating capacity models [7, 38], transverse section equilibrium models [36], and stability criterion models [5, 10, 16]. Experimental measurements of, for instance, electromyographic activities in spinal muscles and

deformations of the passive spine or disk pressures can be used for an independent verification of numerical results or as input data to handle the redundancy in muscle activation [9, 16, 17, 38]. Although the maximum moment-generating capacity models are well elaborated [7], the equilibrium models for submaximal effort incorporating the spinal stability criteria [9] require still further development.

Muscles acting on the spine can be divided into local and global systems. Muscles originating from the pelvis and inserting at the lumbar vertebrae belong to the local system. Global muscles transfer loads directly between the thoracic cage and the pelvis [5]. Previous studies have suggested that the group of local extensor muscles running from the lumbar vertebrae to the pelvis – the iliocostalis, multifidus, longissimus thoracic, and quadratus lumborum muscles – is most important for maintaining the stability of the lumbar spine [10]. In static equilibrium models, which estimate maximum resistance to external loads, decoupling of forces carried by osteoligamentous spine and moments carried by the muscles is often assumed [21]. Optimization procedures are then employed to determine activations in the muscles, with the objective of reaching the maximum sagittal, lateral, or axial moment at the lumbar (L1–L5) [7] or T12 [38] vertebrae. The resultant moment-resisting capacity becomes a function of the maximum contractile stress in the mammalian muscle assumed to be between 0.4 and 0.8 MPa [5, 7, 26]. Changes in muscle orientations associated with different spinal configurations can further affect the potential of the lumbar spine to resist external moments [25, 33]. The inclusion of passive bending resistance of the individual lumbar motion segments is reported to significantly enhance the maximal moment-bearing capacity of the spinal system as compared with ball and socket joints [38].

To assess the effect of muscle stiffness on the mechanical stability of the lumbar spine, a simplified model of the muscle in an isometric condition has often been used [5, 10, 38]. These studies have quantified effects of muscle stiffness on maintaining the stability of the lumbar spine.

Although the available modelling techniques can provide a good insight into spinal behavior in conditions of maximal exertion, development of new techniques is necessary for analysis of the spine in the neutral posture under physiological load. Previous studies [20] have shown that, in a relatively simple model of the thoracolumbar spine under a physiological load of 400 N, only minimal muscle activations are necessary to support and stabilize the passive spine. This physiological load is much greater than the critical load of 20 N applied at T1, which the osteoligamentous spine, unassisted by the muscles, can resist before the hypermobility is observed [22]. The numerical findings of minimal muscle activations during standing neutral posture under the body weight are corroborated by in vivo experiments on normal subjects [32].

This indicates that the passive spine and its accessory muscles in neutral posture exhibit synergy, i.e., the "behavior of the whole system is unpredicted by the behavior of its parts taken separately" [15]. Although the passive osteoligamentous and the active muscular components of the thoracolumbar spine are physiologically distinct structures, in neutral postures they should not be decoupled into individual subsystems. In accordance with the proposed synergetic hypothesis of the human spine in the neutral posture, the objectives of this work are:

1. To develop a synergetic spinal model combining its passive components (osteoligamentous spine) and active components (muscles)
2. To utilize the synergetic model in the analysis of the spine in neutral postures under the body weight
3. To identify the essential conditions for maintaining the equilibrium and stability in a normal spine during submaximal efforts

A modified forward static optimization method [2] is used in our present model to analyze the lumbar spine, with a muscle architecture consisting of 60 muscle fascicles [23–25]. In addition to values of the muscle forces, this method computes deformations with related internal forces resisted by the passive lumbar spine in the upright standing position when subjected to physiological gravity load.

Method

Passive spine

The behavior of the passive lumbar osteoligamentous spine is simulated by a finite element (FE) model using Abaqus structural analysis software [1]. Five vertebrae, L1–L5, are represented by rigid bodies and six disks, T12-S1, by deformable beams running along the centerline of the lumbar spine (LS), as reported in our previous study [20]. The upper torso, from T12 to T1, is modelled as a rigid body, and accounts for varying gravity moments from the upper body due to the postural changes. The model is fixed at its caudal end (S1). Large deformation analysis is used to quantify the effects of changes in the load configuration and the directions of muscle actions. The points of muscle attachments located on the vertebrae are created by defining additional secondary points on each vertebral body.

Muscle architecture

The muscle model consists of 60 muscle fascicles, 12 attached onto the rib cage (global system) and 48 attached onto the lumbar vertebrae (local system). The global system [5], despicted in Fig. 1A, includes the rectus abdominis (RA), external oblique (EO), internal oblique (IO), iliocostalis lumborum pars thoracic (ICt), longissimus thoracic pars thoracic (LTt), and spinalis thoracic (ST). The local system [5], shown in Fig. 1B, includes the iliocostalis lumborum pars lumborum (IC), iliopsoas (IP), longissimus thoracic pars lumborum (LT), multifidus (MF), and quadratus lumborum (QL) [7, 23–25]. The trajectories of the thoracic erector spinae – ICt, LTt, and ST – are modeled as straight lines [38]. Coordinates of insertions points for all global muscles (Table 1) are
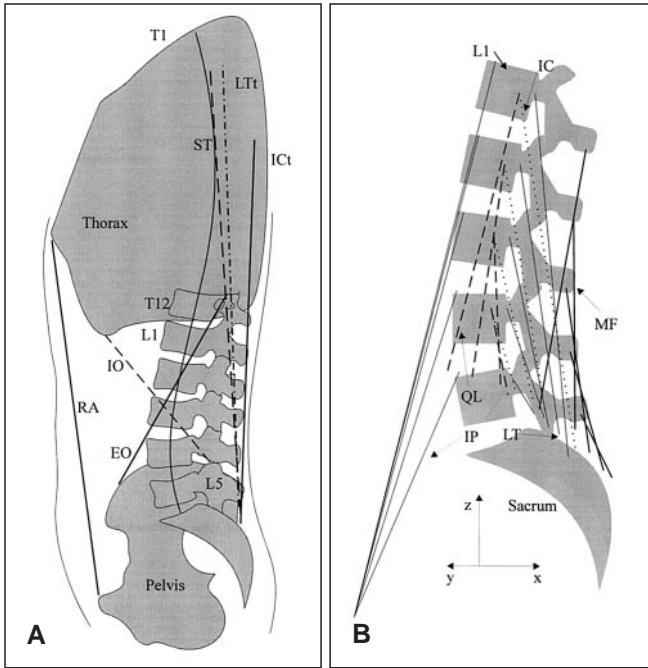
Fig. 1 A Muscles attaching onto the rib cage in a sagittal view. Thoracic erector spinae: *ICt* iliocostalis thoracic pars thoracic, *LTt* longissimus thoracic pars thoracic, *ST* spinalis thoracic. Abdominal muscles: *IO* internal oblique, *EO* external oblique, *RA* rectus abdominis. Rigid parts (thorax, lumbar vertebrae and pelvis) are indicated by *shaded areas*. **B** Muscles attaching onto the lumbar spine at each vertebral level. *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. Vertebrae are indicated by *shaded areas*

Table 1 Coordinates of insertions and origins of fascicles for muscles in the global and local groups (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. Lumbar muscles: *IC* iliocostalis lumborum pars lumborum, *IP* iliopsoas, *LT* longissimus thoracic pars thoracic, *MF* multifidus, *QL* quadratus lumborum)

Muscle	Insertions (mm)			Origins (mm)		
	x	y	z	x	y	z
Global muscles						
RA	−135	50	258	−80	34	−80
EO	38	45	208	−60	122	29
IO	−80	96	167	30	107	51
IC	64	57	360	60	7	−10
LT	30	23	431	58	14	7
ST	17	3	429	60	5	0
Local muscles L1						
IC	22	28	159	63	44	−7
IP	−1	10	174	−40	79	−88
LT	21	10	159	58	42	−9
MF	39	−9	131	39	35	14
QL	22	28	159	9	77	32
Local muscles L2						
IC	13	31	127	49	49	12
IP	−10	12	141	−40	79	−88
LT	12	12	126	54	44	2
MF	31	−5	98	56	30	4
QL	13	31	127	18	67	30
Local muscles L3						
IC	6	32	95	44	53	18
IP	−18	17	105	−40	79	−88
LT	6	14	94	49	45	5
MF	28	−3	68	61	25	−7
QL	6	32	95	28	56	26
Local muscles L4						
IC	7	35	64	37	58	23
IP	−19	21	69	−40	79	−88
LT	7	17	63	47	46	10
MF	34	−1	43	65	16	−16
QL	7	35	64	28	47	21
Local muscles L5						
IC	14	37	31	30	62	20
IP	−12	25	31	−40	79	−88
LT	14	19	30	42	47	12
MF	44	0	15	67	10	−19

based on parametric anatomical studies [4, 8, 12, 13, 26, 29]. The straight lines of action, shown in Fig. 1A, are assumed also for the muscles inserting onto the lumbar vertebrae. Insertion points for the local muscles are located on the lumbar vertebrae at the following landmarks: IC, transverse process; LT, accessory process; MF, spinous process; QL, transverse process; IP, sideways in-between the vertebrae [6, 23, 35]. The coordinates of insertions on the vertebrae are obtained by positioning an idealized lumbar vertebra into its corresponding location along the LS centerline (Fig. 1B). A three-dimensional geometric model of the pelvis [30, 31], scaled according to the anthropometric data [31] to correspond to the present spinal geometry (female, body weight 68 kg, height 170 cm) [20], is used to locate coordinates of muscle origins identified by their attachment areas [7, 23, 30, 35]. During the analysis, the muscle origins at the pelvis are fixed, while their vertebral/thoracic insertions change positions with the motions of the vertebrae and the rib cage, thus allowing for an update of the vectors of muscle action. The muscle geometry exhibits small lateral asymmetry due to the initial lateral deviation of the spinal centerline.

Equilibrium study

A forward static optimization method [2] is adapted for investigation of the spinal response in submaximal exertions close to neutral posture. The analysis is performed by two modules: a passive module, accounting for elastic deformations of the spine due to the external and the muscle loads and an active module incorporating

the muscle architecture and solving for the activations in the muscles modelled as simple force generators, grouping together passive and active components of muscle force [11].

In the passive module a set of virtual springs attached to each vertebra allows for the transfer of antero-posterior shear force, F_x, lateral shear force, F_y, flexion-extension moment, M_y, lateral moment, M_x, and axial moment, M_z, between the passive and the active spinal components at each level. The axial compression, F_z, resulting from both external load and muscles is carried entirely by

the passive spine. Thus, the load sharing between the passive spine and the muscles is controlled by the actions of the virtual springs. The forces in the springs (i.e., stabilizing component of muscle forces) constrain the passive spine to maintain a stable position.

The constraint forces and the updated insertion points are, at each increment, subsequently transferred from the passive module to the active module and used to calculate muscle forces and additional compressive load acting on the spine. Since the number of constraint forces/moments or equations (i.e., maximum of five) at each vertebra is less than the number of attaching muscles, optimization procedure minimizing the compressive load exerted by the muscles [11] is used to solve the redundant muscle problem at each of the T12–L5 levels. The vertical compressive load from the muscles is then transferred into the passive module and applied to each vertebra as an additional external load. An iterative procedure, where passive and active modules are utilized alternately [20], is used to minimize the unbalanced portion of the vertical load from the muscles.

The active displacement constraints by the virtual springs can be chosen from the complete set of u_x, u_y, φ_x, φ_y, φ_z resulting in corresponding constraint forces F_x, F_y, M_x, M_y, M_z respectively. The number of non-zero positive muscle forces for a chosen set of constraints is equal to the number of constraints. In the case when no displacement constraint is applied at a particular degree of freedom (DOF), the net force from muscles in that free DOF has to be evaluated and applied externally along with the axial compression. Moreover, assigning a zero stiffness to a spring at a DOF (i.e., no load carried by the muscles in that direction) represents a case in which the sum of muscle forces in that particular DOF becomes zero [20].

Stability study

To investigate the stability of the spinal system (passive components together with muscles) in a deformed configuration under static load and in the absence of an afferent feedback, the constraints for the muscles from the equilibrium analysis are replaced by the springs [19] originating and inserting into the same locations as the muscle fibers [16]. A simplified muscle model assuming that the muscle stiffness is directly proportional to the muscle force by a coefficient "q", and inversely proportional to muscle length, $k = qF/l$ [5] is used to find stiffness for the muscles. Thus, the springs represent each active muscle in terms of its geometry and elastic properties, with their stiffness based on the forces and lengths obtained from the equilibrium analysis and a coefficient "q". The FE model of the spinal system, with either constraints (equilibrium analysis) or springs for the muscles (stability analysis), possesses an identical deformed configuration and force distribution. For each equilibrium configuration a minimal value of the coefficient "q", "q_{crit}", and its eigenvalue corresponding to a metastable state of the system are found [5, 16]. Previous studies [5] report values of "q_{crit}", both under body weight and under heavy loads, close to 40; however, more recent studies [16] indicate values of "q_{crit}" during maximum effort to be around 5. Overall, the values of "q" reported in the literature range from 0.5 to 42 [16].

Considered parameters

The gravity load, a total of 380 N, is applied at its physiological locations [20]. First, the effect of T1 horizontal displacement in a sagittal direction 40 mm anterior and 20 mm posterior to its initial position [20, 28] is investigated with S1 fixed. The T1 position is controlled by the action of the global muscles, with the local muscles inactive. A parametric study with three levels of horizontal constraint at T12, corresponding to spring stiffnesses (k_h) of 5, 10, and 100 Nmm^{-1}, is performed to evaluate the effect of sagittal po-

sitioning of the thoracolumbar junction on the spinal response. The rotational constraints for sagittal and lateral directions at T12 are the same for all three values of the horizontal constraint and correspond to spring stiffnesses (k_{mx} and k_{my}) of 60,000 Nmmrad^{-1}. This value is close to their critical stiffnesses, with the upper trunk above the T12 considered as an inverted pendulum. The critical stiffness k_{crit} can then be calculated as $k_{crit} = P \cdot l$ [5, 39], where $P \doteq 280$ N is a compressive load above the T12 disk and $l \doteq 180$ mm is its corresponding height above T12. The rotational stiffness k_{mz} is zero, since no significant amount of torsion occurs in the neutral posture.

The effect of the local muscles is examined with all three levels of horizontal constraint at T12. The local muscles are activated in a configuration equilibrated by the global muscles only, when the T1 vertebra is placed 40 mm anterior to its initial position. The overall activation of the local muscles is quantified in terms of the sagittal moments, m_y^{loc}, which they exert at each lumbar level, thus eliminating the need for the sagittal rotational springs. The activation increases up to 500 Nmm of the extension moment exerted by the local muscles at each lumbar vertebra [18]. The lateral translational spring constraint (k_{fy}) at the L1–L5 levels is set to zero and no displacement constraint is imposed on sagittal translation, since the passive spine in the neutral posture has a sufficient capacity to transfer sagittal shear forces. The lateral flexion stiffness (k_{mx}) and axial rotational stiffness (k_{mz}) are also set to zero, since moments in those directions are small and do not significantly affect distribution of muscle forces based on equilibrium conditions.

Results

The compression penalty from muscle exertions in the presence of the global muscles only reaches its minimum with the T1 displacement in proximity to its initial horizontal position (Fig. 2). In a region close to the initial T1 position, when the moment from the external load changes from flexion in the anterior position to extension in the posterior, the compression penalties for all three
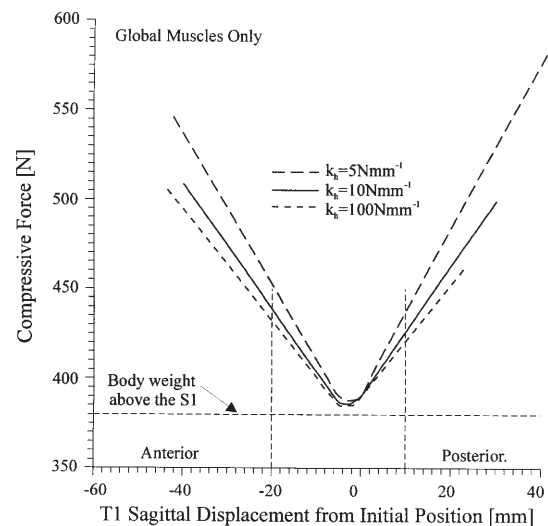

Fig. 2 Compression penalty for different levels of horizontal constraint at T12 in the presence of the global muscles only and different T1 sagittal displacements from the initial position. Body weight above S1 is 380 N

Table 2 Effect of horizontal constraint at T12 and activation of global and local muscle groups on configurations and forces of the spine. Local muscles are activated from the configuration equilibrated by only global muscles when T1 is 40 mm anterior to its initial position

Parameter	$k_h = 5$ Nmm^{-1}		$k_h = 10$ Nmm^{-1}		$k_h = 100$ Nmm^{-1}	
	Glob	Glob + loc	Glob	Glob + loc	Glob	Glob + loc
u_x^{T1} (mm)	–40.0	–32.0	–40.0	–36.5	–40.0	–40.2
u_z^{T1} (mm)	–3.6	–2.9	–3.6	–3.4	–4.3	–4.3
u_x^{L1} (mm)	–5.6	–3.1	–3.3	–2.1	1.0	1.0
u_z^{L1} (mm)	–2.3	–2.3	–2.4	–2.3	–2.7	–2.6
$\Delta\varphi_y^{L1-L2}$(°)	–3.2	–3.2	–4.0	–4.0	–5.7	–5.7
ΔP_{mf}^{glo} (N)	114	61	129	84.9	157	121
ΔP_{mf}^{loc} (N)	–	53.3	–	53.6	–	53.6
mf_{glo} (N)	128.5	69.0	147.9	98.5	184.8	144.0
mf_{loc} (N)	–	71.4	–	71.9	–	72.0
M_p^b (Nmm)	3786	2484	2352	1526	341	787

k_h = horizontal stiffness at T12
Glob = global muscles active only
Glob + loc = global and local muscles active
u_x^{T1} = sagittal movement at T1
u_z^{T1} = vertical movement at T1
u_x^{L1} = sagittal movement at L1
u_z^{L1} = vertical movement at L1
$\Delta\varphi_y^{L1-L2}$ = rotation of the L1–L2 segment
ΔP_{mf}^{glo} = compression penalty from global muscles
ΔP_{mf}^{loc} = compression penalty from local muscles
mf_{glo} = total force in all global muscles
mf_{loc} = total force in all local muscles
M_p^b = moment carried by L5–S1 disk

Fig. 3 Deformed shapes for different levels of horizontal constraint at T12 with T1 40 mm anterior to its initial position in the presence of the global muscles only. As the horizontal constraint at the T12 increases, displacements of the lumbar spine change from the anterior to the posterior

Fig. 4 Sagittal moments carried by osteoligamentous lumbar spine for different levels of horizontal constraint at T12 with T1 40 mm anterior to its initial position in the presence of global muscles only

horizontal constraints at T12 exhibit nonlinearity. For the larger T1 displacements, the compression penalty increases linearly with a varying degree of horizontal constraint at T12. The displacements for the three levels of horizontal constraint at T12 are given in Table 2.

When T1 is displaced 40 mm anteriorly from its initial position, the sagittal displacements of the lumbar spine vary with the horizontal constraint at T12 (Fig. 3). For the intermediate constraint $k_h = 10$ Nmm^{-1}, the horizontal displacements of the LS below L1 are minimal. For $k_h = 100$ Nmm^{-1} the displacements become posterior, and for $k_h = 5$ Nmm^{-1} anterior with respect to the geometry obtained for the intermediate k_h value (10 Nmm^{-1}). The lower levels of the horizontal constraint at T12 are associated with lower constraint forces and activation in the global muscles, thereby resulting in smaller moments resisted by the LS at T12–L1 (Fig. 4). At the same time, the anterior movement of T12 results in an increase of the moment resisted by the LS at S1.

Optimization procedure predicts activations in the 5 most effective muscles from the total of 12 considered, for each of the T1 positions associated with its corresponding load distribution (Fig. 5). Muscular activities, constructed as equilibrium paths, increase linearly with T1 displacement away from its initial position. In the anterior T1 po-
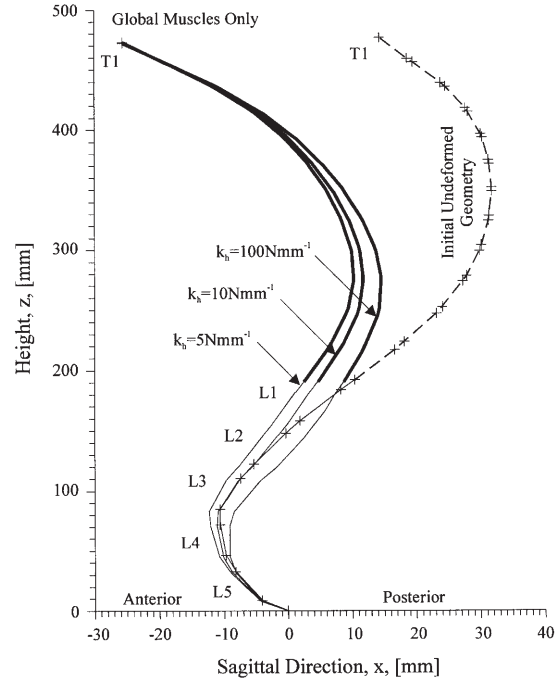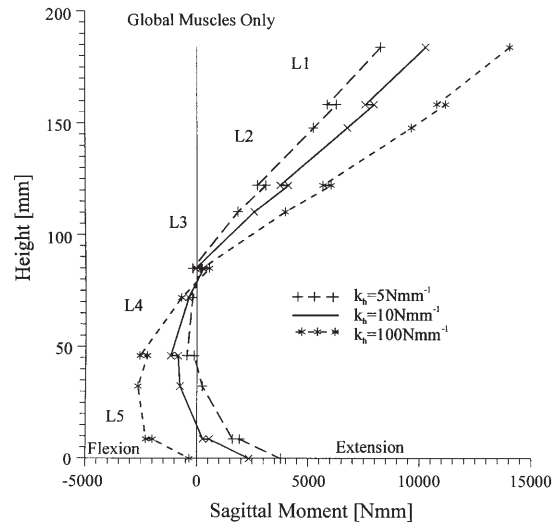
sitions, activities are observed in the iliocostalis pars lumborum (IC) on both sides, with slight asymmetry. In addition to both internal obliques (IO), small activities are observed in the left longissimus thoracic (LTt). In the posterior positions of T1, major activities are in both external
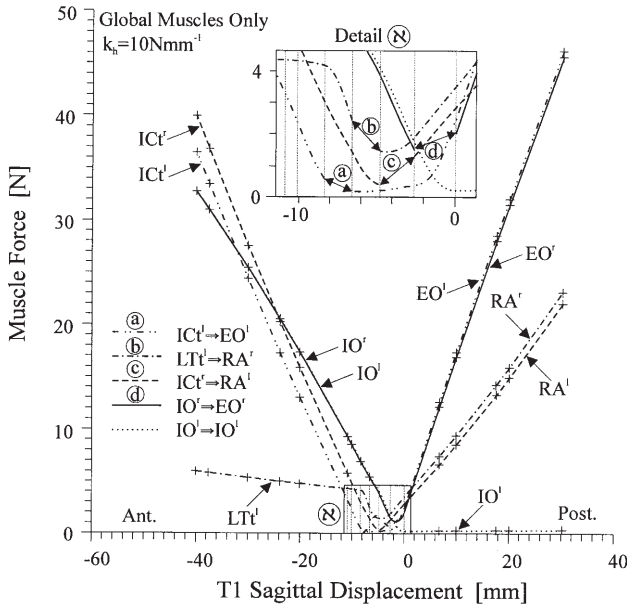
Fig. 5 Activations in global muscles for T1 displacements from 20 mm posterior to 40 mm anterior to its initial position. Switching between the muscles active in the anterior and the posterior T1 positions in the interval –10 mm; 0 mm> is shown in an *enlarged detail* ℵ. Calculated points are located at the + symbols and at the intersections of the curves with the vertical dotted lines in the detail ℵ

obliques (EO). The rectus abdominis acts as an efficient flexor of the spine, and is half as active as EO, while the left internal oblique is almost inactive. Transition between the two five-muscle sets, one active in the anterior and the other in the posterior T1 positions, happens in a region from –8 mm to 0 mm from the T1 initial position (Fig. 5, detail ℵ). The dependencies of muscle forces on the T1 positioning in Fig. 5 are related through geometric transformations to the compression penalties shown in Fig. 2. Activations of the muscles exhibit some contralateral coactivation of muscles in the anterior and posterior positions of the T1 vertebra. The lateral asymmetry in muscular activations is a consequence of the slight lateral deviations of the initial geometry of the thoracolumbar spine.

In the 40-mm anterior T1 position, equilibrated by the global muscles, local muscles are activated to exert an equivalent extension moment ($m_y^{loc} = 500$ Nmm) at each lumbar level, producing a total muscle force (mf_{loc}) of 72 N (Table 2). The major part of muscle activities at all lumbar levels (around 80% of mf_{loc}) is observed in the multifidi muscles, which are the most effective extensors of the spine (Table 3). Some activities, 14%, are observed also in the left iliocostalis lumborum pars lumborum (LTl) at the L5 level and the left quadratus lumborum, 3.5%. The rest of the local muscles account for less than 0.8% of their total activation. The activation of the local muscles causes a posterior movement of the spine (Fig. 6), accompanied by an increase in the compression penalty (Table 2). The resistance to an external flexion moment is split between the passive spine and muscles, as shown in Fig. 7. The sagittal moments carried by the muscles can be further subdivided between the global and the local muscle groups (Fig. 8). As the levels of horizontal constraint at T12 increase, the restorative displacement of T1 toward its initial position caused by activation of the local muscles decreases, so that at $k_h = 100$ Nmm^{-1} the displacement reverses to become anterior (Table 2). With activation of the local muscles, forces in the global muscles decrease (Tables 2, 3 and Fig. 8). Nevertheless, the compression penalty for a specific equilibrium T1 position with both local and global muscle groups active is higher than with only the global muscles active. The interaction between the global and the local muscle groups (Table 2) is reflected also in the proportion of extension moments exerted by each of the two muscle groups (Fig. 8).

Activation of the local muscles, although increasing the compression penalty, contributes significantly to the stability of the spine. The first critical load, corresponding to a lateral buckling mode, evaluated as an additional weight added onto the arms, increases with activations in the local muscles, as compared with global muscles active only (Fig. 9). No stable states are predicted at values of the muscle stiffness coefficient "q" less than 8.4 (to the left from the vertical line passing through Σ_0). Similarly, to the right, the stable region is delimited by a vertical line passing through the maximal possible "q" value, reported in the literature to be around 40. Increase of activations in the local muscles increases also the minimal "q" value (a value below which the structure becomes unstable), e.g., at the points Σ_0, Σ_{200}, Σ_{500} the corresponding "q" values

Table 3 Activations in fascicles for muscles (*r* right side muscles, *l* left side muscles) in the local group as a fraction of total force in local muscles (mf^{loc} in Table 2)

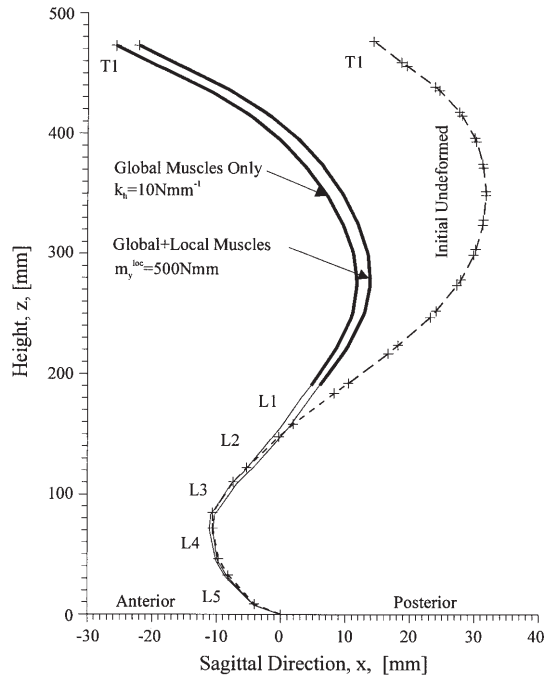| Level	Muscle force in local muscles for $k_h = 10$ Nmm^{-1} (% of mf_{loc})									
	IC^r	IC^l	IP^r	IP^l	LT^r	LT^l	MF^r	MF^l	QL^r	QL^l
L1	0.5						7.9	5.9		1.2
L2		0.1					8.7	6.6		0.9
L3					0.2		8.9	7.5		0.7
L4					0.2		8.8	7.1		0.7
L5		13.9		0.1			17.1	3.2		
Total	0.5	14.0	–	0.1	0.4	–	51.4	30.3	–	3.5

Fig. 6 Effect of activations in local muscles on the spinal configuration at the equilibrium position when T1 is 40 mm anterior to its initial position. The assumed activation for local muscles corresponds to 500 Nmm extension moment at each lumbar level, which results in a 3.5 mm posterior movement at T1
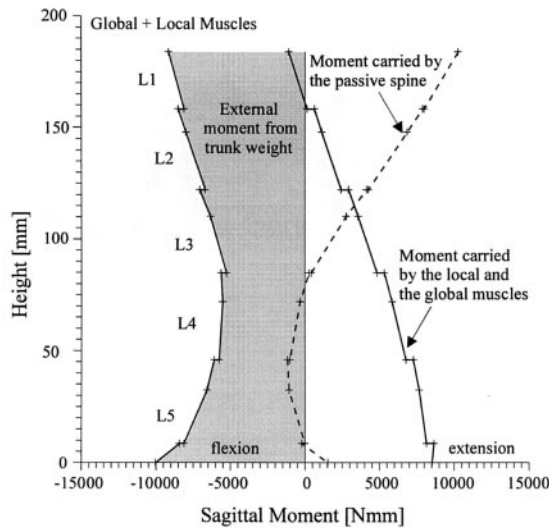

Fig. 7 Sagittal moments from the trunk weight carried by the passive osteoligamentous lumbar spine and muscles. The local muscles are activated in a configuration when T1 is 40 mm anterior to its initial position. The moment generated by the physiological gravity load is indicated by a shaded area. The sum of the moments from the passive spine and the muscles is equal to the moment from the gravity load

are 8.4, 9.0, and 10.5. With the local muscles active, the stable region must always lie below a line "*l*", corresponding to a considered level of activation m_y^{loc}. Varia-
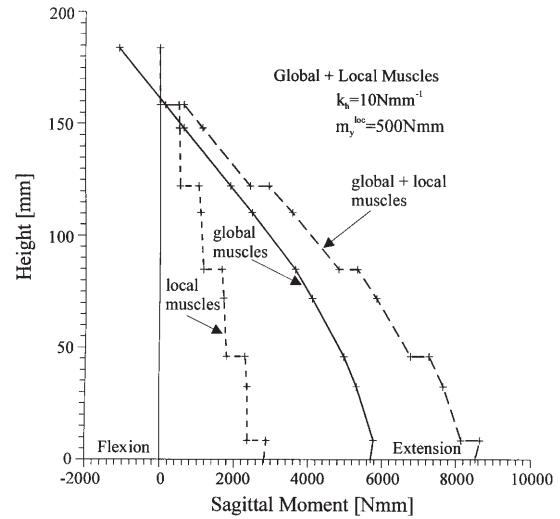

Fig. 8 Partition of the sagittal moments carried by the muscles between the local and the global groups

tion of the spring stiffness for the global muscles controlling the horizontal position at T12 does not significantly affect the first buckling load nor the minimal muscle stiffness coefficient values in the considered range of displacements.

Discussion

Although the passive spine and the muscles are two physiologically distinct entities, in neutral posture they function as a unique synergetic system [15]. In maximum exertion situations, the effect of relatively low load-bearing capacity of passive tissues is often neglected [7, 25, 26]. During submaximal activities, nevertheless, the effect of passive stiffness may become important [9, 38]. The present study investigates conditions for spinal equilibrium in neutral posture under physiological gravity load during a small sagittal movement at the T1 level in static conditions, i.e., the motions being so slow that no significant inertial effects occur. Predicted patterns of muscle activations (Fig. 5, Table 3) are dependent on the assumption of the cost function and represent only minimal necessary degree of coactivation, since they are based on equilibrium conditions. A parametric study of the function of the local muscles (Table 3) indicates that their magnitudes vary according to the required degree of stability of the spinal system (Fig. 9). It is to be stressed that the goal of this paper is primarily to further develop methodology presented in a previous issue [20]. Accordingly, the emphasis in creation of the muscular anatomy is placed on its functionality rather than on anatomical details. Interindividual variations, age, or any form of defective anatomy are not considered. Since the neutral posture is considered only under a static physiological gravity load sustained in
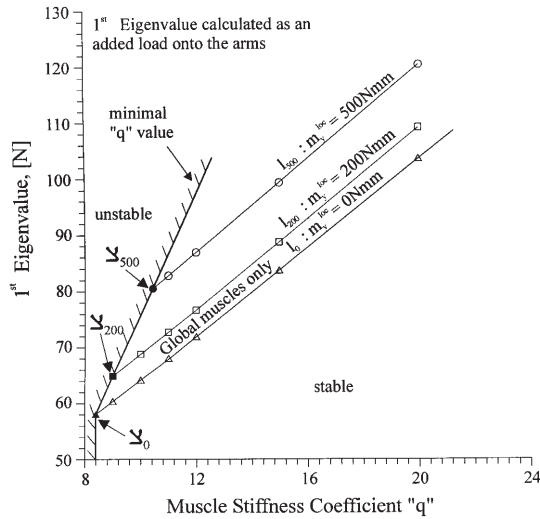
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Fig. 9 Effect of activations in local muscles on the stability of the thoracolumbar spine evaluated by a linear eigenproblem at a final deformed configuration. Stable states are confined to the area to the right of the tick-marked border and below the line "*l*", corresponding to activation of the local muscles

an isometric muscle state, any conclusions drawn from this study are relevant only under the specified conditions.

The muscle architecture includes all major muscles in both local and global muscle groups; however, the inter-segmental muscles in the lumbar region are not included, since their moment-generating potential in the neutral posture is very low. A simplified muscle model is used for stability studies with an overall muscle stiffness coefficient assumed to be the same for all muscles [5, 16].

Experimental studies with surface electrodes provide electromyographic (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 tries to overcome some shortcomings of the EMG-driven models and evaluates muscle forces in individual muscle fascicles based on kinematic and optimization criteria, with a full interaction between the passive spine and muscles (Fig. 1). The sequence of equilibrium states of the spine (a multi-joint elastic structure with muscles) for different T1 positions is controlled by kinematic constraints modelled as virtual springs [14]. The virtual springs simulate the overall effect of the neural system on the control of spinal synergy. As such, the method allows for a realistic modelling of muscle activities, depending on the passive stiffness properties and geometric configuration of the spine. The results of the present study are qualitatively corroborated by the EMG findings [32], and give a further insight into spinal mechanics in neutral postures.

Action of the global muscles only, spanning from the rib cage to pelvis, in conjunction with passive spinal resistance, is found to be sufficient for maintaining spinal equilibrium and stability during small sagittal displacements at the T1 level. The activations in the global system depend on the T1 sagittal position, muscle forces increasing linearly [34] with the distance from the initial T1 position. The recruitment pattern of the global muscles shows coactivation [19] (Fig. 5): when in anterior and posterior T1 positioning respectively IC and IO, and EO and RA are active on both the left and right sides. The magnitude of muscle forces in the global muscle system depends also on the degree of horizontal constraint at T12 (Table 2). The postures in which T12 is not allowed to move horizontally with sufficient freedom (Fig. 3) may result in a significant increase of flexion moments at the L1 and L2 levels (Fig. 4). Activation of the local muscles decreases the forces in the global system attached onto the rib cage and provides additional stiffness, thus increasing overall spinal stability [11] (Fig. 9). The majority of activities at the lumbar levels is observed in the multifidi muscles; some activity, 14%, is also observed in the left iliocostalis lumborum pars lumborum (Icl) at the L5 level and 3.5% in the left quadratus lumborum [3, 27].

The vertical load from the upper body weight and the muscles attached to the rib cage cause variations of sagittal moments in the lumbar region due to the lordotic curvature. Without efficient flexor muscles in the lumbar region, since the iliopsoas in neutral posture cannot exert the required flexion moments [6] (Table 3), some moment in the lordotic arch appears to be inherently carried by the passive spine (Fig. 7). The muscular activations in equilibrium positions in neutral posture under the postural load exhibit small variations in the recruitment pattern with varying degrees of activation in global muscles (different values of horizontal constraint at T12) and local muscles (different levels of m_y^{loc}). The position of the thoracolumbar junction, controlled by a horizontal constraint at the T12, has a marked effect on the distribution of the intersegmental rotations of vertebrae in the lumbar spine (T12–S1) (Table 1, Fig. 3) and thereby also on the magnitude of the sagittal moments carried by the passive spine (Fig. 4).

The predictions of the recruitment pattern and magnitudes of muscle forces in this study are dependent upon the cost function, requiring that the compressive load from muscles be minimal at each lumbar level. However, choice of different criteria for resolution of a redundant muscle problem should not lead to a significant augmentation of the reported muscle activities, since they provide a sufficient degree of stability [11]. The modular structure of the presented approach – the passive and the active module – offers a good potential for future development. The passive module, utilizing a commercial structural analysis software, Abaqus, allows for accommodation of a more detailed nonlinear behavior of the vertebral segments, including a contact phenomenon in the facets.

Similarly, the active module, an assembly of the Fortran routines, allows for the incorporation of a more complex model of the muscular behavior. These improvements would extend the presented approach into more general postures, with complex algorithms of muscle activation.

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