Combination of finite element modeling and optimization for the study of lumbar spine biomechanics considering the 3D thorax–pelvis orientation

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Abstract

A model of the lumbar spine capable of taking into account realistic loads derived from human activity would be of great benefit in studying its normal biomechanical functioning as well as its in vivo behavior in injured and surgically altered states. This paper proposes a method to analyze the mechanical response of the lumbar spine subjected to loads derived from human activity, combining a non-linear finite element model (FEM) and an optimization-based force predicting algorithm. Loads borne by the lumbar spine at the T12–L1 level (joint loads) are first predicted with the optimization algorithm and then applied to the FEM, while a boundary condition prescribing the relative L1–sacrum rotation is imposed onto the FEM to account for three-dimensional physiological thorax–pelvis orientation. The prescribed rotation is achieved through the application of moments on L1. To account for the effect of these moments on lumbar joint loads, an iteration between the optimization technique and the FEM computation has been carried out. This method provides two main benefits over previous studies: first, it allows for the application of any 3D loading condition while considering the real 3D rotation measured between the thorax and the pelvis, and second, it makes it possible to estimate the moments that must be applied on L1 in order to maintain this rotation when predicting joint loads. As an example application of the method, results are presented for the lumbar spine mechanical response at the time of peak T12–L1 joint force during walking.

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Keywords: Lumbar spine; Finite element analysis; Spinal loading; Optimization; Biomechanical modeling

1. Introduction

An analytical model of the lumbar spine, such as a finite element model (FEM), can be a valuable tool in simulating the intact lumbar spine for the purpose of providing data on its normal biomechanical functioning, as well as for studying the in vivo behavior of the spine in its injured and surgically altered states. In both cases, the ability to simulate the mechanical behavior of the lumbar spine under realistic loads derived from dynamic studies of human activities would offer valuable insight which could be applied to a wide range of situations. To accomplish this goal, it is necessary to combine the model with a method to obtain the loads borne by the lumbar spine.

Finite element modeling is a commonly used technique for the analysis of lumbar spine biomechanics [1–6]. Researchers tend to subject their FEM to generic loads such as compression alone [2], pure moments [4] or moments combined with compression [5,6].

The prediction of loads on the lumbar spine has also been extensively researched [7–14]. In order to predict these loads, a strategy must be adopted to partition the reaction moment in the lower back into the different components that share moment generation roles. Two main approaches have been used to solve the partitioning problem: optimization techniques [9,10,13] and EMG driven models [8,12,14]. The EMG driven methods have the advantage of using neural drive measurements from EMG taken from the subject and are thus able to predict individual muscular activation patterns. However, their
experimental requirements cause many researchers to choose more generic methods such as optimization-based ones.

Several optimization methods have been proposed by researchers to predict loads on the lumbar spine. One of these methods involves minimizing the sum of cubed stresses in all muscle fascicles. This method, which maximizes muscular endurance, was first proposed by Crowninshield and Brand [15]. Hughes et al. [16] later found that this algorithm worked better than several other methods at predicting activation patterns of trunk muscles, and since then it has been used by several researchers to predict lumbar loads [13,17,18].

The combination of FEM and load predicting techniques to simulate the mechanical response of the lumbar spine to loads from human activity has yet to be extensively researched. Two different approaches have been adopted thus far: in the first approach, muscle and joint forces are computed using an optimization algorithm and these forces are then applied to the FEM [17,19]; in the second approach muscle forces are considered as a model parameter which is increased step-wise until the FEM response fits the experimental measurements [20].

The loading conditions in these studies were always sagittally symmetrical, including posture loading [19,20] and lifting in the sagittal plane [17]. Only Rohlmann’s group [20] considered the real rotation of the lumbar vertebrae, taking it into account indirectly by using it as a second criterion for predicting muscle forces.

Shirazi-Adl and Parnianpour [21] also considered the real rotation of the vertebrae in an FEM, although in this study a force predicting method was not used. The effects of changes in lordosis on the mechanics of the lumbar spine in lifting were studied by prescribing the changes in sagittal segmental rotation while subjecting the FEM to a theoretical compression load. This approach took into account the moment contribution of the lumbar segment but the effect of this moment on loads on the lumbar spine was not considered.

The current study proposes a method to analyze the biomechanical response of the lumbar spine combining an FEM of the lumbar spine and an optimization-based force predicting technique. The aim of this method is to develop a model of the lumbar spine which permits the analysis of its mechanical response under realistic loads derived from human activity. For this purpose, the lumbar spine has been modeled taking into account that it must bear loads derived from human activity while at the same time allowing for the relative motion needed between thorax and pelvis to carry out the activity. The proposed method offers some significant advantages over the previous studies mentioned above. First, it permits the analysis of the response of the lumbar spine to any 3D load derived from human activity while considering the 3D thorax–pelvis rotation needed for that activity, and second, it makes it possible to estimate the moments that must be applied on L1 in order to maintain this rotation, taking them into account when predicting the loads borne by the lumbar spine. In order to verify the applicability of the method, the response of the lumbar spine at the time of peak T12–L1 joint force during level walking is analyzed in this paper. This case addresses the two main questions of whether or not the imposition of the relative thorax–pelvis rotation is able to stabilize the ligamentous lumbar spine, and how the imposed rotation affects the predicted joint loads and intradiskal pressure.

2. Method

The method proposed herein involves an iteration between a lumbar spine FEM and a lumbar spine load predicting technique. This section therefore describes the FEM developed and the optimization technique used to determine loads, as well as the iterative procedure used to apply the joint loads to the FEM.

2.1. General description

The lumbar spine is a flexible structure which bears loads derived from human activity while allowing for motion between the thorax and the pelvis. The method proposed in this paper simulates this situation by using an FEM of the lumbar spine which is subjected to the T12–L1 joint forces derived from human activity, while the relative rotation between the L1 vertebral body and the sacrum is constrained to maintain the thorax–pelvis orientation needed for that activity.

The human activity under study was optically measured and analyzed to find the relative orientation between the thorax and pelvis, as well as the reactive forces and moments at the T12–L1 level (net reaction). The thorax–pelvis orientation measurements were imposed on the FEM of the lumbar spine as a boundary condition prescribing the relative rotation between L1 and the sacrum (L1–S BC) to match the 3D orientation observed between the pelvis and thorax. This is achieved by applying a 3D moment on L1, which represents the moment transmitted through the T12–L1 ligamentous motion unit (joint moment) that would be needed to bear the joint forces while maintaining the measured thorax–pelvis orientation.

The T12–L1 joint forces were estimated from the net reaction by an optimization algorithm which allows for the inclusion of the T12–L1 joint moment. Since this moment was obtained from an FEM computation, the following iterative strategy was carried out, which involves computing both the FEM and the optimization algorithm (a flow chart of this iterative process is illustrated in Fig. 1):
Fig. 1. Flow chart of the iterative process combining FEM analysis and load predicting algorithm. $\Delta M_{L1} < \delta$ represents the convergence criterion used, where $\Delta M_{L1}$ is the difference between the moment generated by the L1–S BC in the last two iterations, and $\delta$ is the assumed error.

- Initially, the T12–L1 joint moment was assumed to be zero, and the T12–L1 joint forces computed by the optimization algorithm along with the L1–S BC were applied to the FEM.
- The FEM was analyzed under this initial set of conditions to establish the moment on L1 generated by the L1–S BC (ML1).
- Joint forces were computed again by the optimization algorithm using the moment on L1 generated by the L1–S BC as the T12–L1 joint moment.
- The new joint forces were then applied to the FEM, and the FEM was analyzed again, thus obtaining a new moment generated by the L1–S BC on L1.
- This process was repeated until the difference between the moment generated by the L1–S BC in the last two iterations ($\Delta M_{L1}$) was smaller than a predetermined value (convergence criterion). $\Delta M_{L1}$ represents the difference between the joint moment used for the estimation of loads and the one actually borne by L1. As the joint moment is one of the terms added to balance the reactive moments in the load predicting algorithm, we chose this predetermined value as a percentage of the reactive moment.
- Finally, the joint forces obtained in the last step are considered as the definite joint forces borne by the lumbar spine.

This process was applied for the example application described in Section 2.7, the mechanical response of the lumbar spine at the time of peak joint force during walking. In this case, the iteration was stopped when $|\Delta M_{L1}|$ was below 1% of the reactive moment at the T12–L1 level.

2.2. Finite element model

The FEM herein is a detailed, three-dimensional, non-linear model of the entire L1–S ligamentous lumbar spine (Fig. 2), which is comprehensively described in Refs. [22,23].

Its geometry is parametric and can be defined by a set of geometric parameters (up to 55 for each vertebra) which can be measured from radiographs and CT scans. Some of the main parameters used are shown in Fig. 3. The mean values of these parameters, as published by Panjabi et al. [24], have been used here for the application of the method.

All lumbar vertebrae as well as their articulations (intervertebral discs, ligaments and facet joints) were modeled using the COSMOS/M commercial finite element package. Each vertebra has been modeled with eight node isoparametric elements (SOLID) using homogeneous and isotropic material properties. Different modeling techniques were used for the intervertebral disc. The nucleus pulposus was simulated as an incompressible fluid using SOLID elements with a low Young
Fig. 3. Main geometric parameters used for the reconstruction of the FEM geometry. (a) Anterior view of a vertebral body, (b) lateral view of two consecutive vertebral bodies and (c) longitudinal view of a vertebral body and its pedicles showing a transverse cut of the articular facets.

... modulus and a Poisson ratio close to 0.5; the disc annulus was modeled as a composite material using SOLID elements to simulate the elastic ground substance and using tension only (cable) elements with elastic non-linear material properties to represent the annulus fibers. Ligaments were also simulated as cable elements, oriented along the respective ligament fiber alignment, with elastic non-linear material properties (Fig. 4). Finally, the facet articulation was modeled as a three-dimensional contact problem using interface elements (GAP) to simulate the contact between the articulating facets, which were represented by two cylindrical surfaces. A total of 20 GAP elements were used to model each contact surface. Based on CT scan measurements, in the application shown here, a value of 0.5 mm was chosen for the initial gap between facets. The element types and material properties used are listed in Table 1.

The material properties used in the study were derived from the literature [3,6,25,26]. We selected material properties whose behavior in the model response best reflected those of published experimental lumbar response. The global coherence of the FEM was checked by comparing the predicted functional unit and entire L1–S response under different loading conditions with experimental data found in the literature, as well as with results from previously validated models [1,2,6,27–33]. Compared results included: vertebral mobility, mean stiffness, coupled motions, intradiskal pressure, the influence of compression on the functional lumbar unit response under axial torque, and the influence of posterior elements. A detailed description of the validation results can be found in Ref. [22]. As an example of the FEMs behavior, Fig. 5 shows the predicted mean stiffness of a motion segment and compares it to reported experimental results. Also, as in the application shown here, the model was subjected to a combined load whose main components were compression and axial moment, it is worth mentioning that the mean axial torsion stiffness found for the motion segment increased from 5.3 to 6.9 N m/degree when a compression of 400 N was added to the 6 N m axial moment. This result concurs with previously published in vitro studies [28,30] and other finite element studies [6,31].

2.3. Estimation of loads on the lumbar spine

The loads on the lumbar spine due to human activity were estimated by optically measuring and dynamically analyzing the activity under study. Three-dimensional image analysis, using two cameras and the direct linear transformation algorithm [34], was used to reconstruct the link segment model of the upper part of the body, as explained in Section 2.7.

Once the link model was constructed, the net reaction at the T12–L1 level was obtained by solving the inverse dynamics problem. This reaction was subsequently distributed among the supporting tissues (muscle forces, joint forces and joint moments) by an optimization algorithm performed using the MatLab optimization toolbox (The MathWorks Inc.). The criterion used for optimization was the minimization of the sum of the cubed stresses in all muscle fascicles cut by an imaginary transverse plane across the T12–L1 lumbar level [15], thus the following cost function was employed:

\[
\text{Cost} = \sum_{j=1}^{m} \left( \frac{F_j}{\text{pcsa}_j} \right)^3
\]  

where \( F_j \) and \( \text{pcsa}_j \) are the force magnitude and the physiological cross-sectional area of the \( j \)th muscle slip,
Table 1
Element types and material properties used in the FEM. The material properties used were derived from the literature [3,6,25,26]

<table>
<thead>
<tr>
<th>Material</th>
<th>Element types</th>
<th>Young’s modulus (MPa)</th>
<th>Poisson’s ratio</th>
<th>Cross-sectional area (mm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical bone</td>
<td>Volumic (SOLID)</td>
<td>12000</td>
<td>0.3</td>
<td>–</td>
</tr>
<tr>
<td>Cancellous bone</td>
<td>Volumic (SOLID)</td>
<td>3500</td>
<td>0.25</td>
<td>–</td>
</tr>
<tr>
<td>Bony posterior elements</td>
<td>Volumic (SOLID)</td>
<td>100</td>
<td>0.2</td>
<td>–</td>
</tr>
<tr>
<td>Nucleus pulposus</td>
<td>Volumic (SOLID)</td>
<td>100</td>
<td>0.2</td>
<td>–</td>
</tr>
<tr>
<td>Disc</td>
<td>Volumic (SOLID)</td>
<td>0.13</td>
<td>0.499987</td>
<td>–</td>
</tr>
<tr>
<td>Annulus ground substance</td>
<td>Volumic (SOLID)</td>
<td>175 (&lt;15%)</td>
<td>0.3</td>
<td>3</td>
</tr>
<tr>
<td>Annulus fibers</td>
<td>Cable (TRUSS3D)</td>
<td>450 (&gt;15%)</td>
<td>0.45</td>
<td>–</td>
</tr>
</tbody>
</table>

Facet joints
Contact (GAP)
Ligaments
Cable (TRUSS3D) (Fig. 4)

Anterior longitudinal 53
Posterior longitudinal 16
Supraspinous 23
Ligamentum flavum 67
Interspinous 26
Transverse ligament 17.3
Capsular 60

respectively, and \( m \) is the number of muscle slips considered, this being 18 in our study.

The imposed constraints (Eq. (2)) are: the moment equilibrium, the fact that muscle forces must be equal to or greater than zero, and the muscle stresses must not exceed the maximum permissible stress.

\[
\sum (r_i \times F_i \cdot u_i) + M_{L1} - M = 0
\]

\[
\frac{F_{i_{pcsa}}}{F_{i \geq 0}} \leq S
\]

where \( r_i \) and \( u_i \) are, respectively, the vector from the T12–L1 disc center to the centroid of muscle slip ‘i’ and the unit vector representing the direction of muscle slip ‘i’; the symbol ‘\( \times \)’ is the vector cross product; \( M_{L1} \) is the T12–L1 joint moment; \( M \) is the reactive moment (net reaction) at the T12–L1 level, and \( S \) is the maximum permissible muscle stress. We used different muscle stress limits within the physiologically acceptable range of 30–100 N cm \(^{-2}\) [35], but in the application presented here (Section 2.7), the same results were obtained regardless of the limits used since no muscle ever reached maximum activation.

This optimization algorithm is used in the iteration process described in Section 2.1. The T12–L1 joint moment was neglected in the first step of the iteration and was obtained from the FEM (moment generated by the L1–S BC) in subsequent iterations. Therefore, it was considered as a known input in the optimization formulation. The intra-abdominal pressure contribution was neglected and therefore the T12–L1 joint forces were computed as the sum of the muscle and reactive forces.

2.4. Muscle forces

Two main types of muscle forces can be defined: forces exerted between structures above and below the
lumbar spine (global forces), and those exerted by muscle fascicles attached to the lumbar vertebrae (local forces). Various studies on the influence of muscular forces on the mechanical behavior of the lumbar spine in neutral postures [19], as well as in flexion of the upper body and standing [4,20], have shown that global muscle forces are crucial to the control of the overall spinal posture and have a strong influence on the stresses in the lumbar spine, whereas local muscle forces have little influence on its mechanical behavior. Based on the results of these studies, we have not included local muscle forces in the FEM of the lumbar spine in the application presented here, which involves level walking, and thus the FEM was only subjected to joint loads.

However, in the method described here, the effect of the muscles has been taken into account in two ways: first, muscle forces were included into the load predicting scheme, and second, the rotation between L1 and the sacrum was prescribed, which indirectly accounts for the contribution of all muscle forces to the stability of the lumbar spine.

2.5. Estimation of the relative thorax–pelvis orientation

A set of markers on the thorax and pelvis of the subject under study (Fig. 6) were put in place to measure the 3D relative orientation between the thorax and pelvis from the digitalized images. The relative orientation between the anatomical coordinate systems of both body segments (anterior, longitudinal and lateral axes) was calculated using Cardan angles in the following sequence:

1st rotation: around the transverse axis of the thorax. Flexion–extension.
2nd rotation: around the anterior axis of the thorax. Lateral flexion.
3rd rotation: around the longitudinal axis of the thorax. Axial rotation.

2.6. FEM boundary conditions

To reproduce the relative rotation between the thorax and the pelvis in the FEM, the method considered the sacrum as a fixed structure and the rotation of the L1 vertebral body was constrained to the measured thorax–pelvis Cardan angles, from which 20% was subtracted to take into account the relative rotation between the rib cage and L1. This value was chosen considering the fact that there are three vertebral segments between the rib cage and L1. If the rotation between the rib cage and pelvis were equally distributed among the five lumbar and three lower thoracic segments, the lower three would account for 3/8 of the total thorax–pelvis rotation, or 37.5%. However, the range of motion of the lower thoracic segments is slightly smaller than that of the lumbar segments [25], so it was assumed that their rotation would also be proportionally smaller, and they thus account for approximately 20% of the total thorax–pelvis rotation.

The L1 vertebral body rotation was imposed using a boundary condition consisting of a set of equations coupling the displacements of different L1 vertebral body nodes:

\[ \mathbf{u}_{p1} - \mathbf{u}_{p2} = \mathbf{R}_{ZYX}(\mathbf{x}_{p1} - \mathbf{x}_{p2}) - (\mathbf{x}_{p1} - \mathbf{x}_{p2}) \]  \hspace{1cm} (3)

where \( \mathbf{u}_{p1} \) and \( \mathbf{u}_{p2} \) are the displacements of two arbitrary L1 vertebral body nodes, \( \mathbf{R}_{ZYX} \) is the rotation matrix defining the L1 vertebral body orientation, which in turn depends on the Cardan angles, and where \( \mathbf{x}_{p1} \) and \( \mathbf{x}_{p2} \) are the coordinates of the P1 and P2 nodes.

Eq. (3) is deduced when the L1 vertebral body is considered as a rigid body. In this case, if the motion of the vertebral body is defined by a displacement \( \mathbf{X} \) and a rotation matrix \( \mathbf{R}_{ZYX} \), then a point \( P \) of the vertebral body with coordinates \( \mathbf{x}_p \) after the motion yields \( \mathbf{x}_{p'} = \mathbf{X} + \mathbf{R}_{ZYX}\mathbf{x}_p \), and its displacement is \( \mathbf{u}_p = \mathbf{x}_{p'} - \mathbf{x}_p \). Therefore, the difference between the displacements of any pair of L1 vertebral body nodes must satisfy the relation stated in Eq. (3). The imposition of this relationship, which couples the displacements of a series of L1 vertebral body nodes in the model, results in attaining the desired

![Fig. 6. Markers used to define the thorax and pelvis coordinate systems: thorax; (1) suprasternale; (2) and (3) R & L thelion; (4) substernale; pelvis; (5) and (6) top of iliac crest; (7) and (8) R & L anterior superior iliac spine; (9) pubic crest.](image)
3. Results

Fig. 7 shows the time histories of the three predicted components of joint force at T12–L1 during walking when the joint moment at T12–L1 is assumed to be zero. Peak joint force during the analyzed movement occurs approximately at the time of toe off with a value of 976.3 N. At this time the following values for the reactive moment, joint force and pelvis–thorax relative rotation were obtained:

**Joint Force:**
- Longitudinal force: 974.2 N (compression)
- Lateral shear: 24.1 N (right shear)
- Anterior–posterior shear: 59.5 N (forward shear)

**Relative rotation of the thorax with respect to the pelvis:**
- Flexion–extension: 1.5° (flexion)
- Lateral bend: 2° (right lateral bend)
- Axial twist: 4° (right twist)

The FEM was first subjected to the predicted maximum joint forces, and the L1 vertebral body rotation was prescribed to 1.2° flexion, 3.2° axial twist and 1.6° lateral bending (the result of reducing the pelvis–thorax relative rotations by 20%). Convergence was achieved after three iterations, finally yielding the results listed in Table 3. The residual moments are also listed in this table. These values indicate the differences between the magnitudes found in the last two steps of the iteration for the components of the moment generated by the L1–S BC.

In the case studied, all three joint force components decreased when the joint moment applied on L1 was considered, with a reduction of 21% in compression, 38% in anterior shear and 25% in lateral shear. This reduction is due to the change in muscle forces when the T12–L1 joint moment is taken into account. Table 4 lists the muscle forces obtained in the first iteration, when the T12–L1 joint moment was neglected, and the final muscle forces, when this moment was taken into consideration.

Fig. 8 shows sagittal and frontal profiles of the FEM before and after load application. All vertebrae moved anteriorly in the sagittal plane and to the right in the frontal plane, with the greatest displacements being in the case of the L1 vertebral body: 4.2 mm in the anterior direction and 4.9 mm in the lateral direction. The vertebral rotation and the corresponding lumbar motion segment rotation were also calculated using the Cardan angles (Fig. 9). The results for flexion–extension show differences in the direction of rotation of the different lumbar motion segments: segments L5–S and L1–L2 flexed while the others extended, with the lumbosacral joint showing the maximum rotation. As for lateral flexion, all motion segments rotated in the direction of the pelvis–thorax rotation, that is, to the right, with the upper two segments showing very little lateral flexion.
Table 2
Physiological cross-sectional area, direction cosines and centroid coordinates at the T12–L1 level for the muscles considered [37,38]. The parameters referred to the local axes system located at the center of the T12–L1 disc (t: left lateral direction, l: longitudinal direction and ap: anterior direction)

<table>
<thead>
<tr>
<th>Area (cm²)</th>
<th>Direction cosines</th>
<th>Centroid coord.</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>t</td>
<td>l</td>
</tr>
<tr>
<td><strong>Left side muscles</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Rect. abdominis</td>
<td>5.67</td>
<td>−0.04</td>
</tr>
<tr>
<td>External oblique Slip 1</td>
<td>1.96</td>
<td>−0.47</td>
</tr>
<tr>
<td>Slip 2</td>
<td>2.32</td>
<td>−0.29</td>
</tr>
<tr>
<td>Slip 3</td>
<td>2.43</td>
<td>−0.19</td>
</tr>
<tr>
<td>Slip 4</td>
<td>2.34</td>
<td>−0.15</td>
</tr>
<tr>
<td>Quadratus</td>
<td>1.90</td>
<td>0.11</td>
</tr>
<tr>
<td>Psoas</td>
<td>2.17</td>
<td>−0.02</td>
</tr>
<tr>
<td>Erect. spinae equiv</td>
<td>24.24</td>
<td>−0.01</td>
</tr>
<tr>
<td>Latissimus dorsi</td>
<td>3.37</td>
<td>0.003</td>
</tr>
<tr>
<td><strong>Right side muscles</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Rect. abdominis</td>
<td>5.67</td>
<td>0.04</td>
</tr>
<tr>
<td>External oblique Slip 1</td>
<td>1.96</td>
<td>0.47</td>
</tr>
<tr>
<td>Slip 2</td>
<td>2.32</td>
<td>0.29</td>
</tr>
<tr>
<td>Slip 3</td>
<td>2.43</td>
<td>0.19</td>
</tr>
<tr>
<td>Slip 4</td>
<td>2.34</td>
<td>0.15</td>
</tr>
<tr>
<td>Quadratus</td>
<td>1.74</td>
<td>−0.02</td>
</tr>
<tr>
<td>Psoas</td>
<td>2.17</td>
<td>0.05</td>
</tr>
<tr>
<td>Erect. spinae equiv</td>
<td>24.17</td>
<td>0.02</td>
</tr>
<tr>
<td>Latissimus dorsi</td>
<td>3.60</td>
<td>−0.13</td>
</tr>
</tbody>
</table>

Fig. 7. Joint forces at T12–L1 during walking when joint moment is considered zero. Fl, Ft and Fap: compression, right lateral shear and anterior shear, respectively.

The axial rotation was more homogeneously distributed among the different motion segments. All motion segments exhibited the same direction of axial rotation as the pelvis–thorax, with the outermost segments L1–L2 and L5–S showing a somewhat greater rotation.

Table 5 lists the intradiskal pressures of the five lumbar disks and the total facet contact forces at all lumbar levels. The predicted intradiskal pressures of the model vary between 0.58 (L3–L4) and 0.69 MPa (L1–L2). As for the facets, contact forces were higher on the left side than on the right due to the direction of the axial rotation. The greatest forces were found in the L5–S segment, while the L2–L3 and L3–L4 facets did not transmit any force at all.

4. Discussion

Direct validation of the method proposed here is virtually impossible, since it is impractical to measure the majority of the parameters predicted by the model in vivo. However, the FEM has been validated by comparing its output with in vitro experimental results, while the load predicting model has been tested by several researchers. Also, the iteration which couples both models has been performed with the inclusion in the optimization algorithm of a moment generated in the FEM by a boundary condition simulating the relative orientation between pelvis and thorax, which has been measured experimentally.

The method described here was applied to the study of the mechanical response of the lumbar spine at the moment of highest spinal loading during walking. On analyzing the results, it is interesting to note that the vertebral displacements obtained were relatively small, although the only limitation imposed was the prescription of the L1 rotation. The highest values were about 4 mm in the anterior direction and 5 mm in the lateral direction. These results confirm how the L1–S BC used
Table 3
T12–L1 joint forces and joint moments obtained after three iterations. The error indicates the percentage of the residual moments with respect to the corresponding reactive moment.

<table>
<thead>
<tr>
<th>Direction</th>
<th>Longitudinal</th>
<th>Transversal</th>
<th>Anterior</th>
</tr>
</thead>
<tbody>
<tr>
<td>Joint forces (N)</td>
<td>772 (compression)</td>
<td>18 (right shear)</td>
<td>36.7 (forward shear)</td>
</tr>
<tr>
<td>Required joint moment (N m)</td>
<td>4.3 (flexion)</td>
<td>0.33 (left bending)</td>
<td>3.5 (right twist)</td>
</tr>
<tr>
<td>Residual joint moment (N m)/error</td>
<td>0.04/0.1%</td>
<td>0.03/0.7%</td>
<td>0.01/0.2%</td>
</tr>
</tbody>
</table>

Table 4
Muscle forces at the initial and final iteration step, computed using the optimization algorithm

<table>
<thead>
<tr>
<th>Muscular forces (N)</th>
<th>Initial values (no moment at L1)</th>
<th>Final results (considering moments at L1)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Left</td>
<td>Right</td>
</tr>
<tr>
<td>Rect. abdominis</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>External oblique 1</td>
<td>15</td>
<td>0</td>
</tr>
<tr>
<td>External oblique 2</td>
<td>0</td>
<td>17.4</td>
</tr>
<tr>
<td>External oblique 3</td>
<td>0</td>
<td>29.8</td>
</tr>
<tr>
<td>External oblique 4</td>
<td>0</td>
<td>31.2</td>
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<tr>
<td>Quadratus</td>
<td>9.7</td>
<td>1.6</td>
</tr>
<tr>
<td>Psoas Major</td>
<td>10.1</td>
<td>0</td>
</tr>
<tr>
<td>Erect. spinae</td>
<td>401.1</td>
<td>203.1</td>
</tr>
<tr>
<td>Latissimus dorsi</td>
<td>9.7</td>
<td>0</td>
</tr>
</tbody>
</table>

Note: no moment contribution of the T12-L1 motion segment is considered in the initial step.

Fig. 8. Sagittal and frontal profile of the lumbar spine vertebral body centers, before and after load application.
Fig. 9. Individual lumbar segment rotations. A positive sign indicates flexion, left lateral flexion and right axial twist of the thorax with respect to the pelvis.

is able to stabilize the ligamentous lumbar spine, which has been proven to exhibit instability under relatively small compression forces [2,39].

The boundary condition used to prescribe L1–sacrum rotation involves the application of moments on L1. These moments are included in the force predicting algorithm, therefore the transmission of moment through the T12–L1 motion segment (joint moment) was taken into account when evaluating the loads on the lumbar spine. It is important to point out that the application of axial moments is the only way to produce considerable axial rotations in the lumbar spine. Therefore, when considerable axial rotation takes place, the joint moment cannot be neglected and it thus becomes necessary to adopt a method such as the one proposed here.

In the application studied, the highest computed components of the T12–L1 joint moment were the flexion moment (4.3 N m) and the axial moment (3.5 N m), both of which are within the physiological range, with the lateral bending moment being considerably smaller at 0.33 N m. The direction of transversal (flexion) and axial joint moments was the same as that of the corresponding reactive moments at the T12–L1 level, so the motion segment contributes to withstand the external reactive moments, and thus diminishes the muscle activation required. The predicted T12–L1 axial joint moment was somewhat lower than the corresponding reactive moment. This means that the reactive axial moment, which is mainly produced by the inertia derived from movement, can provide the axial rotation needed at the moment of peak joint force during walking, with no need for additional muscular activity.

The muscle forces predicted by the optimization algorithm were, obviously, different when the T12–L1 joint moment was considered than when it was assumed to be zero (Table 4). These differences caused a reduction in the T12–L1 joint forces from 21% (compression) to 38% (anterior shear) when the T12–L1 joint moment was considered. The differences in joint forces when the joint moment is included in the force predicting algorithm have not been taken into account in previous lumbar spine FEM studies that have considered the prescription of vertebral rotation [2] or which combined optimization and FEM [17]. The main reason for the reduction of the joint force is the decrease in the muscle-generated axial moment due to the moment contribution of the T12–L1 motion segment. This results in a lower prediction of the right external oblique activity, and thus a lowering of the shear joint force due to the orientation of this muscle, which is shown in Table 2, as well as a decrease in the opposition to the reactive lateral moment. Alongside the flexion contribution of the motion segment, this decrease results in a lowering of the left erector spinae force, and thus a lowering of the joint compression force.

Intradiskal pressure was also reduced as a direct consequence of the decrease in joint forces. The predicted L4–L5 intradiskal pressure was reduced from 0.76 to 0.62 MPa when the T12–L1 joint moment was considered. The latter value is within the range found by Wilke et al. [40] for walking: 0.53–0.65, while the former value is higher than the highest value measured by Wilke et al.

Finally, one must take into account that the iterative nature of the method implies successive non-linear finite element calculations of the detailed model of the lumbar spine, which may involve a high computational cost. However, it is of interest to point out that in the application shown here, convergence was reached in only three steps, with negligible residual moments.

5. Conclusions

A method for the analysis of the biomechanical response of the lumbar spine subjected to realistic loads

<table>
<thead>
<tr>
<th>Lumbar motion segment</th>
<th>L1–L2</th>
<th>L2–L3</th>
<th>L3–L4</th>
<th>L4–L5</th>
<th>L5–S</th>
</tr>
</thead>
<tbody>
<tr>
<td>Intradiskal pressure (MPa)</td>
<td>0.69</td>
<td>0.64</td>
<td>0.58</td>
<td>0.62</td>
<td>0.63</td>
</tr>
<tr>
<td>Total contact forces (N)</td>
<td>Left</td>
<td>13.7</td>
<td>0</td>
<td>39.7</td>
<td>128.6</td>
</tr>
<tr>
<td></td>
<td>Right</td>
<td>0</td>
<td>0</td>
<td>66.7</td>
<td></td>
</tr>
</tbody>
</table>
derived from human activity has been proposed. The method involves the combined use, in an iterative manner, of an optimization-based force predicting algorithm and a detailed FEM of the lumbar spine.

The method can be applied to study the normal biomechanical functioning of the lumbar spine as well as the in vivo behavior of the spine in its injured and surgically altered states. Regarding the latter point, the methodology proposed could be of great benefit to develop a model to study different surgical techniques used for the stabilization of the injured lumbar spine. This model could serve to analyze the mechanical behavior of the fixation devices used and different lumbar components, and would thus provide valuable assistance when designing the surgical technique to be used.

The method was applied to study the mechanical response of the lumbar spine at the time of peak T12–L1 joint force during walking. The vertebral displacements obtained were relatively small, showing how the boundary conditions used are able to stabilize the ligamentous lumbar spine. The T12–L1 joint moments generated by the L1–S BC, which simulates the measured thorax–pelvis rotation, contribute to withstand the reactive moments, thereby reducing the muscle activation and the joint loads predicted by the optimization algorithm in comparison with the values obtained should these moments be neglected. As a consequence of this reduction, the predicted intradiskal pressure also decreased when the motion unit moment contribution is reduced, the predicted intradiskal pressure also

References


