Biomechanical evaluation of human lumbar spine in spondylolisthesis

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ABSTRACT

One of the least known conditions of the lumbar spine in terms of biomechanics is spondylolisthesis which causes many serious consequences for the patient. This research aimed to perform a mechanical analysis of the origins of spondylolisthesis and its impact on the biomechanics of the lumbar section of the spine. Within the framework of this study, a physiologically model of the lumbar spine was created in the MADYMO software. In the next stage a slip of vertebra L4 was simulated by means of a controlled forward displacement of the vertebral body of vertebra L4. 10 variants of spondylolisthesis (W1 – W10) of different degrees were subjected to a biomechanical evaluation. In maximum bending of the physiological spine at an angle of 27° the value of the shear force amounted to 1.9 kN, while for the spine affected by spondylolisthesis with slip grade W9 at the maximum bending of 34° the shear force amounted to 5.5 kN. It was observed that the lumbar spine with the simulated spondylolisthesis had greater mobility in comparison with the physiological spine, which was shown by maximum bending angles (physiological 27°, W9 34°).

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Introduction

Research premises

Health ailments connected with the spine are mainly caused by the degeneration of the intervertebral disc, joints surfaces and post-traumatic changes. Despite numerous studies, the diseases of the spine still pose a major problem. Moreover, it is expected that with the further civilization development and ageing society such illnesses will only continue to grow. The research of the lumbar section of the human spine is a frequent subject of scientific deliberations. The interest in this field has increased also in Poland in the past years. Nevertheless, there are still many issues connected with the lumbar spine which remain unsolved. The authors were motivated to undertake the research on this issue by the mass occurrence of pain and ailments in the lumbar spine in patients. This condition is connected with technological development and related lifestyle, prolongation of working hours and lack of active leisure. In spite of many educational programmes aiming to increase the public awareness in this scope, the pain of the back is one of the most common reasons for seeing the doctor. One of the most frequent conditions occurring in the lower section of the spine is spondylolisthesis, which is a form of a chronic instability of the vertebral column. It consists in the displacement of the column of vertebrae in relation to the vertebra located below it (Hanson et al., 2002; Labelle et al., 2004; Natarajan et al., 2003; Sairyo et al., 2001). The etiology of its occurrence in terms of biomechanics has not been fully studied yet. The knowledge of the mechanism of its origin will have a vital importance for the improvement of treatment of such conditions.

Objective of this work

The objective of the research was defined as follows: to determine forces occurring in this segment of the spine for different degrees of progression of the vertebral body slip. In order to implement this objective, a dynamic model of the human lumbar spine was developed using a method of multibody systems. The determination of spinal loads in dynamic conditions enabled a better understanding of the mechanism of spondylolisthesis occurrence. Performed researchers were carried out with the cooperation with neurosurgeon, who indicates the specific patient with isthmic spondylolisthesis on the L4–L5 level. According to the
new classification of the spine deformity (Labelle et al., 2009; Labelle and Mac-Thiong, 2011; Mac-Thiong et al., 2012) considering patient can be classified to type 2 of isthmic spondylolisthesis. This is very rare case, but the preoperative surgical planning is usually used for custom cases. It was also another reason for doing this research because of the lack of reports in the literature.

Material and methods

A numerical model of the lumbar spine was formulated in the MADYMO software using a method of the dynamics of multibody systems. This method consists in modelling of objects by means of a chain of rigid bodies linked by kinematic pairs. Individual rigid bodies are described by mass, inertia moments and the position of the gravity centre. Moreover, in the kinematic pairs the bonds between the bodies are defined as well as their position in relation to a reference coordinate system. Both the bodies and the kinematic bonds are described by means of a series of physical parameters making it possible to solve movement equations of the whole system (1):

\[ M\ddot{x} + C\dot{x} + Kx = P_t \]

where:
- \( M \) – system mass matrix;
- \( K \) – system stiffness matrix;
- \( C \) – system damping matrix. This matrix is most often adopted in a form of the so-called proportional damping (depending on \( K \) and \( M \) matrices);
- \( P_t \) – vector of external loads of the system;
- \( x \) – vector of linear displacement;
- \( \dot{x} \) – vector of linear velocity;
- \( \ddot{x} \) – vector of linear acceleration.

The authors’ previous experience in the scope of modelling in biomechanics (Wolanski et al., 2013a) as well as awareness of the issues connected with IT methods used for numerical computation constituted the basis of the applied simplifications. The process of modelling was preceded by literature studies in the scope of the anatomy of the human lumbar spine, the properties of its individual elements as well as the methods of their modelling (Chosa et al., 2004; Konz et al., 2001; Natarajan et al., 2003; Sairyo et al., 2006a,b).

The initial phase of the modelling process omitted the anatomical elements whose influence on the behaviour of the lumbar vertebrae was of little importance in this case and whose modelling would only complicate the model. Therefore soft tissues surrounding the lumbar section were omitted, including: elements of the digestive, cardiovascular and nervous systems. The model consists of the following anatomical parts of the human lumbar spine: five lumbar vertebrae L1–L5, joints connections, intervertebral discs, ligamentous apparatus and sacral bone.

The geometry of individual vertebrae which build up the model was obtained on the grounds of CT images of a patient whose vertebral column in the lumbar section was physiologically correct. CT imaging made it possible to segment a very complex and elaborate geometry of vertebrae. This enabled the recreation of the system of load transfer by three main elements: vertebral bodies as well as superior and inferior articular processes, i.e. a natural triad of support. The geometry of individual vertebrae (L1–L5) along with the sacral bone (S1) served the purpose of the development of a dynamic model of the human lumbar spine in the MADYMO software. The modelling process adopted the following assumptions:

- structure of models of lumbar vertebrae and the sacral bone reflects their irregular shape which is symmetrical to the sagittal plane;
- vertebrae are treated as rigid bodies having six degrees of freedom (except for the sacral bone constituting an immovable basis) whose position and movement depend on the bonds and inertia forces;
- intervertebral discs connecting the vertebrae are treated as massless elastic and damping elements having a nonlinear characteristic of stiffness and damping;
- intervertebral joints are connected by means of elastic and damping elements with a nonlinear characteristic taking into account the features of both cartilages and joint capsules;
- ligaments are treated as massless elements of a stiffness characteristic which is active only during tension;
- model takes into consideration physiological loads caused by the natural position of the spine – compression and a forward bend – bending.

A numerical model for the calculation of a general motion Eq. (1) used a modified Euler’s method with one stage and constant time step \( t_s \). The position of a local coordinate system of individual rigid bodies in relation to the global coordinate system is defined by the following Eq. (2):

\[ \dot{X}_i = \ddot{T} \dot{t}_i + A_i \dot{x}_i \]

where:
- \( X_i \) – matrix of the leading vector coordinates;
- \( \ddot{T} \dot{t}_i \) – matrix of the coordinates of the vector connecting the beginnings of both coordinate systems;
- \( A_i \) – matrix of direction cosines;
- \( \ddot{x}_i \) – matrix of the coordinates of the displacement vector of the local coordinate system.

Numerical model formulation

In the analysis of dynamic interactions in the lumbar spine in a physiological system and in the case of pathological changes the researchers adopted models of lumbar vertebrae in a form of rigid bodies having six degrees of freedom whose position depended on the ligaments, intervertebral discs and joints surfaces. The geometry of individual vertebrae was developed on the basis of the data selected from computer tomography (CT) which had been performed as routine checks in the scope of diagnostic tests. The segmentation of the vertebrae was made in specialist Mimics software.

The geometry of the lumbar section vertebrae was subject to further processing in the LS-PREPOST software in order to build a dynamic model of the spine. After the transformation of the models of vertebrae and the sacral bone, the models were imported to the MADYMO software (Fig. 1). Individual vertebrae models were connected by means of elastic and damping elements which reflected different intervertebral structures. In such a way a
model of the lumbar section of the human spine was created. This model faithfully recreates the structure of the vertebral column.

Identification of model parameters

For the simulation of a mathematical model one needs information describing its properties and inertia features. The required data include mass. Mass inertia moments of individual vertebrae were determined for the bone density selected on the basis of the data found in literature (Gzik, 2007) (Table 1).

The position of individual vertebrae was defined by means of the determination of the position of their mass centres in an absolute system. The distance between the vertebrae was determined by the models of intervertebral discs. The discs interact with the cores of the connected vertebrae both during compression and tension. That is why in this model of the lumbar spine the researchers decided to adopt discs as weightless elastic and damping elements connecting the adjacent vertebrae.

The model includes intervertebral joints which are represented by elastic and damping elements acting between the joint surfaces. The stiffness characteristic of these elements also takes into consideration the stiffness of cartilages and joint capsules. That is the reason why intervertebral joints are represented by a set of elastic and damping Kelvin-model elements which are located perpendicularly in relation to each other and constantly parallel to the Cartesian coordinate system x, y, z. Joint capsules are represented by four separate elastic and damping elements placed on the circumference of each intervertebral joint (Fig. 2).

Elastic and damping elements of a Kelvin type were adopted as the representation of discs. The elements were located centrally on the surfaces of the vertebral bodies. Individual elastic and damping elements were positioned perpendicularly in relation to each other and constantly parallel to axes x, y, z of the Cartesian coordinate system. Application of the above-mentioned solution enabled the definition of different mechanical characteristics in relation to the axes of the Cartesian coordinate system (Fig. 3).

The stiffness characteristic of the elements representing intervertebral joints was the same as for the intervertebral disc on axes x and y. This simplification was confirmed by experimental tests conducted on the samples of both structures.

Ligaments included in the model are treated as massless elements exerting forces on the characteristic attachment points of the connected rigid bodies (vertebrae). The exerted forces depend on the displacement of the connected elements and the preset stiffness which act on the line connecting the attachment points. Nonlinear stiffness of individual ligaments was selected on the basis of literature data (Rohlmann et al., 2001). Fig. 4 shows models of the above-described elements.

Table 1

<table>
<thead>
<tr>
<th>Model</th>
<th>Mass [g]</th>
<th>Moment of inertia [g mm²]</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Ix</td>
</tr>
<tr>
<td>L1</td>
<td>89.7</td>
<td>21642.0</td>
</tr>
<tr>
<td>L2</td>
<td>86.4</td>
<td>20587.5</td>
</tr>
<tr>
<td>L3</td>
<td>95.9</td>
<td>24291.0</td>
</tr>
<tr>
<td>L4</td>
<td>96.8</td>
<td>24697.5</td>
</tr>
<tr>
<td>L5</td>
<td>84.7</td>
<td>21019.5</td>
</tr>
<tr>
<td>S6</td>
<td>250.1</td>
<td>299895.0</td>
</tr>
</tbody>
</table>

Fig. 2. (a) Representation of intervertebral joints; (b) representation of joint capsules; (c) representation of an intervertebral disc.

The angle of lordosis of the physiological model of the lumbar spine developed in the MADYMO software equalled 51°. The set boundary conditions of the model in the MADYMO programme were constant during the analysis. The model was burdened with a bending moment in the sagittal plane equalling 7.5 Nm. This value resulted from the influence of the upper part of the human body. The support of the model was set on the sacral bone (Fig. 4).

During the simulation of the model a time interval of 1.5 s was taken into consideration. The same time interval was adopted for all analyzed grades of spondylolisthesis. In order to make the differences in the model’s behaviour more visible, the results were presented on a time axis. This way of presenting the results enabled the comparison of dynamic interactions between the spinal structures at certain moments of time. In order to assess the spine biomechanics in the case of spondylolisthesis, the motion of the whole system was analyzed in the function of angular displacement.

Model verification

The verification of the dynamic numerical model was conducted on the basis of the data found in literature (Rohlmann et al., 2001). The results obtained from experimental tests and shown in Fig. 5 were compared with the results of numerical computation of a physiological model of the lumbar spine. Both results were obtained in the same boundary conditions. Due to

Fig. 3. Stiffness characteristic of intervertebral discs: (a) in the direction of axis z; (b) axes x and y.
different duration of the experiment from the time of simulation, the verification of the model was performed on the basis of the comparison of a maximum bending angle at the applied load of 7.5 Nm. In the case of such enforcement the response obtained from the modelled system was close to the behaviour of the spine sample during the experimental tests. In the analyzed load case, satisfactory compliance of the model results with the experimental results was obtained, which is shown in Fig. 5. The difference between the results did not exceed 15%. Qualitative and quantitative compliance of the characteristics obtained from the simulation of the model and from the experimental tests constitutes the basis for an assumption that the lumbar spine model was formulated in a correct way and may serve as a tool for biomechanical evaluation of the lumbar section of the human spine in the case of spondylolisthesis occurrence.

A positively verified model of the human lumbar spine became the basis for further calculations which considered the mechanism of spondylolisthesis occurrence. Simulations of the human lumbar spine model in the case of spondylolisthesis were conducted for the same loading conditions as in the case of the physiological model of the lumbar section. In this way the same loading conditions were preserved and the obtained results (values of angular displacement and shear forces on the surfaces of the vertebral bodies) were used to define the influence of spondylolisthesis on the biomechanical properties of the lumbar section of the human spine.

Numerical simulations

Having performed the verification of the physiological model, the researchers conducted the analysis of dynamic interactions in the lumbar spine in the case of occurrence of pathological changes typical of different grades of spondylolisthesis. At the level of vertebrae L4–L5 the slip of vertebra L4 was simulated by means of cutting the lamina of the vertebral arch. Next, an elastic and damping element of a preset stiffness characteristic was inserted into this place (Fig. 6). Such a solution enables the analysis of different grades of spondylolisthesis, which may be obtained through the adjustment of the forward displacement (forward slip) of vertebral body L4. The displacement of vertebral body L4 towards the abdominal cavity was performed at a 1 mm interval. In such a way 10 variants of spondylolisthesis of different degrees of progression were obtained W1–W10.

The simulation of the lumbar spine model with the prepared slip of vertebra L4 was conducted in the same boundary conditions as for the physiological spine. The results obtained make it possible
to observe the mechanism of spondylolisthesis occurrence. Moreover, the analysis of dynamic interactions in the spinal structures in the case of pathological changes provides a chance of verification of the guidelines for the application of stabilization in the treatment of such cases and development of new criteria. In order to achieve this, the forces present in the intervertebral discs and joints were subjected to analysis.

Simulation results

Diagram in Fig. 7 presents the comparison of the courses of shear forces recorded in the intervertebral discs L4–L5 for the spine affected by spondylolisthesis and for the physiological spine in consecutive positions W1–W10.

According to the criteria of the application of stabilizers in the treatment of the vertebra slip, the incidence of the second-grade spondylolisthesis is an indication that surgical treatment should be applied. That is why in the next stage of this work variant W3 was selected for further numerical analysis. In variant W3 the slip of vertebra L4 amounts to 4 mm and corresponds to the second-grade spondylolisthesis. For the analyzed case of spondylolisthesis the values of shear forces occurring above and below the level of spondylolisthesis site were determined, i.e. at the level of intervertebral discs L3–L4 and L5–S1 (Fig. 8).

The performed dynamic analysis enabled the comparison of kinematics of the lumbar spine with spondylolisthesis with the physiological spine (Fig. 9). A slight increase of the spine mobility was observed for the analyzed grade of spondylolisthesis in this segment, which was shown as angular displacement. For grade W9 of spondylolisthesis the mobility of the analyzed segment increased from 27° to 34°.

Results

On the basis of the performed numerical analyses for the dynamic model of the lumbar section of the human spine with simulated spondylolisthesis it was observed that in the scope of angular displacement from 0° to 19°, the shear force in intervertebral disc L4–L5 was at a similar level to a physiologically correct spine. However, after exceeding the angle of 19° in the analyzed motion segment the increase of the shear force was observed with the increase of spondylolisthesis grade. In maximum bending of the physiological spine at an angle of 27° the value of the shear force amounted to 1.9 kN, while for the spine affected by spondylolisthesis with slip grade W9 at the maximum bending of 34° the shear force amounted to 5.5 kN. In the places above and below the site of spondylolisthesis occurrence no considerable increase of shear forces was noted. Moreover, it was observed that the lumbar spine with the simulated spondylolisthesis had greater mobility in comparison with the physiological spine, which was shown by maximum bending angles (physiological 27°; spondylolisthesis W9 34°). The simulations conducted enabled also the determination of the values of forces on the joints surfaces at the level of vertebrae L3–L4. In this site, it was also observed that the force on the joints surfaces in the analyzed mobility segment increased along with the increase of the displacement of the vertebral body in the sagittal plane in the direction of the abdominal cavity.

Discussion

Spondylolisthesis is strictly connected with the occurrence of pars fracture, which is a unilateral or bilateral defect of a vertebra occurring in 5–6% of the population. Other factors, such as discopathy, degenerations or injuries, also play an important role in the occurrence and progression of spondylolisthesis. It is estimated that around 80% of the patients with pars fracture at the level of vertebra L5 have spondylolisthesis, and 20% of the patients from this group show the slip exceeding 25% (Saraste, 1993). In children, increased stress in the structures surrounding epiphysial cartilages of vertebral bodies may cause degeneration of the intervertebral disc and as a result lead to spondylolisthesis (Farfan et al., 1976; Mac-Thiong and Labelle, 2006; Reitman et al., 2002; Sairyo et al., 2004, 2006a,b). The research on spondylolisthesis described in the literature focuses mainly on the radiological determination of various parameters of the spine which may have an influence on the increase of the risk of spondylolisthesis progression. An important factor contributing to the risk assessment of spondylolisthesis progression is the parameters related to PT and SS angles of the pelvic inclination and sacral inclination (Hresko et al., 2007). On the basis of these parameters, authors

Fig. 6. Stiffness characteristic of the element connecting vertebral body with vertebral arch.

Fig. 7. Lumbar spine model with minimum W1 and maximum W10 displacement of vertebral body.
(Mac-Thiong and Labelle, 2006) propose a clinical method of the classification of slip grades. The same authors also point out to the biomechanical aspects of the occurrence and progression of spondylolisthesis. He suggests that dysplastic changes may have an impact on the direction and load values and thus increase the risk of the slip progression.

Numerical simulations are often applied to evaluate the degree of pathological conditions in anatomical structures or to perform

![Diagram of shear forces](image)

**Fig. 8.** Diagram of shear forces: (a) in intervertebral disc L3–L4 for the second grade of spondylolisthesis; (b) in intervertebral disc L5–S1 for the second grade of spondylolisthesis.

![Angular displacement of lumbar spine](image)

**Fig. 9.** Angular displacement of lumbar spine.
preoperative planning of treatment (Gzik-Zroska et al., 2013; Wolaniski et al., 2013b, 2015). The literature describes models of the lumbar section of the human spine (Chosa et al., 2004; Konz et al., 2001; Natarajan et al., 2003; Sairyo et al., 2006a,b) enabling the simulation of spondylolisthesis which were formed on the basis of the finite elements method. Authors of such models try to familiarize other researchers with the biomechanical aspects of spondylolisthesis occurrence. Natarajan et al. (2003) developed a static model of the lumbar section of the human spine in a physiological system for different grades of spondylolisthesis at the level of vertebra L5. The developed model included vertebrae L4–L5 and sacral bone S1. On the basis of the formed model the authors defined the stiffness of the analyzed fragment of the spine for different grades of spondylolisthesis. Sairyo et al. (2006a) formed an MES model consisting of vertebrae L3–L5 as well as elements of a nonlinear character which modelled a border plate and anulus fibrosus. At the level of vertebra L4, pars fracture was simulated by removing one layer of finite elements – 1.0 mm thick in each vertebral pedicle. As a result of the simulations conducted, it was stated that the above-mentioned pars fracture may cause the damage of the border plate and consequently lead to spondylolisthesis. Chosa et al. (2004) developed an MES model consisting of vertebrae L4–L5 in a physiological system which was later used for the biomechanical analysis of this section of the spine taking into consideration different compressing and bending loads. The simulations conducted made it possible to determine which of the applied loads led to the occurrence of pars fracture.

Developing of dynamic models of spine is relatively novel approach in biomechanical analysis (Joszko et al., 2016). However, it is difficult to find any dynamic model tests of the lumbar section of the human spine which could serve the purpose of determination of the forces occurring during spondylolisthesis progression. In this work author’s models of the lumbar sections of the human spine in a regular physiological system and in the case of spondylolisthesis were developed. The models were formulated with the purpose of the simulation of internal phenomena which appear in the spinal structures for external load of a dynamic character. A dynamic model of the lumbar spine was treated as a sandwich construction with undeformable bodies (vertebrae) including considerably deformable elements (intervertebral discs) and a series of flexible connectors of a nonlinear characteristic (ligaments). A verified model of the physiological lumbar section was used for the development of spinal models in pathological conditions caused by different grades of spondylolisthesis. A group of the formulated dynamic models in all simulation variants was provided with the same boundary conditions corresponding to the verification of the model of a physiologically correct spine. The adoption of identical computational conditions enabled the comparison of the very data obtained as well as their comparison with the results obtained from the spine model in a natural physiological system.

The analysis conducted in the scope of this work encompassed dynamic interactions in the lumbar spine in the case of spondylolisthesis occurrence at the level of vertebra L4–L5, which was simulated by cutting the lamina of the vertebral arch of vertebra L4. In the next step, an elastic and damping element of a certain stiffness characteristic was inserted into this site. On the basis of the performed simulations of the dynamic model of the lumbar section with spondylolisthesis it can be stated that the value of shear force in the intervertebral disc at the level of spondylolisthesis occurrence grows with the increase of the displacement of the vertebral body in the sagittal plane. The above-mentioned force reaches its critical value and causes shearing of the disc. As a result, the vertebral body slips into the abdominal cavity. The research results presented in this work show that an optimum moment to apply surgical treatment of spondylolisthesis is the time of the vertebral arch fracture as it causes the loss of a natural ‘lock’ protecting it against translation in the sagittal plane. The application of the stabilization at such an early stage of spondylolisthesis development could limit the compression on the nervous structures in this area and protect the vertebral body from further translation. Standard surgery is usually performed at an advanced stage of the disease but analyzing our simulation results we could propose that the better time to surgery is even in the moment of diagnosis. Authors are planning to confirm this hypothesis in other studies.

On the basis of our researches concerning with spondylolisthesis of other levels it was noticed, that there is an analogy between level L4–L5 and L5–S1 if it is considering the optimal moment of surgery. Detailed results are planned for publication in further articles.

**Conclusion**

Spondylolisthesis is a long-term process in which the displacement of the vertebral body is accompanied by the adaptation process of the nervous structures in this area. On these grounds it can be concluded that in an advanced stage of spondylolisthesis the application of the vertebral body repositioning to the level of physiological lordosis may not be possible and it is the physiological lordosis which is adapted to bearing the biggest loads.

**Conflict of interests**

The authors have no conflict of interests to disclose.

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