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Pedicle screws, FEA, CAD, 3D modeling

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NUMERICAL ANALYSIS OF SPINAL LOADS IN SPONDYLOLISTHESIS TREATMENT USING PEDICLE SCREWS – PRELIMINARY RESEARCH

Abstract

The aim of this experimental study was to analyse the influence of lumbar stabilisation used in the treatment of spondylolisthesis on the biomechanical properties of the human lumbar spine. FEM models were built on the basis of pre-surgical CT scans, routinely used in medical practice. MIMICS software was used to process the results of the neuroimaging study and to create 3D models. Two models were built: with and without a stabiliser. A static load analysis was performed for a normal upper-body load. The simulations allowed to determine the stresses in the individual discs for both models, with and without the transpedicular stabiliser.

1. INTRODUCTION

There is probably no other system in the human body that plays as important a role for human health and a long and active life as does the spine. The spine forms a kind of axis for the human body, and its specific structure protects the brain and the spinal cord from shocks and injuries. A healthy spine promotes the proper

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function of the nervous system, which has a great impact on a person's general health and wholesome life. The spine is built of alternating layers of hard bony structures called vertebrae and elastic elements called intervertebral discs. This system is stabilized and supported by soft tissues: joints, ligaments and muscles. Structured in this way, the vertebral column allows the body to move and simultaneously transfer loads generated by body weight and bodily activities (Zubrzycki & Smidova, 2014; Zubrzycki, Karpiński & Górniak, 2016).

As clinical practice shows, the most common spinal conditions are those that affect the lumbar spine. This region is exposed to the largest loads, compared to the thoracic and cervical spine. Approximately 30% of orthopaedic patients are lumbar cases. Almost all instabilities are due to muscular imbalance, disc degradation, and intervertebral degeneration. They all lead to the deepening of the lumbar lordosis and misalignment of the lumbar spine relative to the sacrum. These problems are most often caused by mechanical loads. With age, the ligaments get stretched, which results in loosening of the joints, and the gradual loss of muscle efficiency additionally contributes to instability in the motor segments (Yugang et al., 2015).

A typical stress fracture of the spinal bony structures and one of the most common lumbar disorders is spondylolisthesis. This disease is a form of chronic instability of the lumbar spine. It involves slippage or displacement of the column of vertebrae over a vertebra below it (or the sacrum). In the lumbar region, spondylolisthesis, in most cases, occurs at the L5-S1 level of the spine and, less frequently, at levels L4-L5 and L3-L4. The most common complaints are related to the irritation of and subsequent damage to the nervous system or to a pathology of facet joints (Gzik, Joszko & Pieniążek, 2012; Ciupik et al., 1997).

2. ANATOMY AND PHYSIOLOGY OF THE SPINE

The spine is the moving axis of the body and the neck, which is located in the midsagittal plane on the dorsal side of the body. It stretches from the base of the skull to the lower end of the trunk, forming, together with the ribs and the breastbone, the axial skeleton of the body. The spine is made up of 33 to 34 unpaired, symmetrical vertebrae stacked on top of each other to form the spinal column. The vertebral column in divided into five regions (Fig. 1):

- cervical (C1–C7) – 7 vertebrae,
- thoracic (Th1–Th7) – 12 vertebrae,
- lumbar (L1–L7) – 5 vertebrae,
- sacral (S1–S7) – 5 vertebrae,
- coccygeal (Co1–Co4 / Co5) – 4–5 vertebrae.

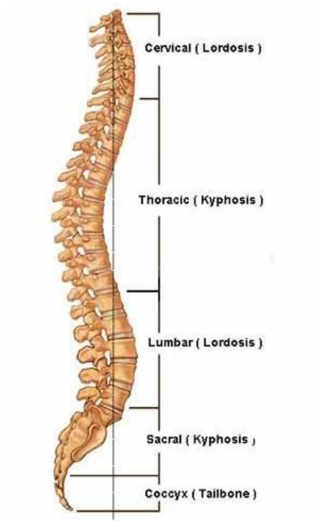


Fig. 1. Lateral view of the spinal column [19]

The spine becomes stronger and more resistant the further down it goes as the vertebrae become larger and more strongly built. This type of structure provides high rigidity and durability, while allowing for flexibility. Owing to this, the spine can fulfil its three basic functions: to protect the spinal cord, to protect the locomotor system and to support the body (Woźniak, 2003; Bochenek, 2010; Karpiński, Jaworski & Zubrzycki, 2016).

2.1. Curvature of the spine

In an adult body, when viewed in the sagittal plane, the spine is seen to have two types of curves: lordosis – with anterior convexity (convex forward) and kyphosis – with posterior convexity (concave forward). Starting from the top of the vertebral column, the following four curves can be distinguished: cervical lordosis, thoracic kyphosis, lumbar lordosis and coccygeal kyphosis. Together, they give the spine its characteristic double-S shape. Cervical lordosis transitions smoothly into thoracic kyphosis and there is a similar smooth transition between thoracic kyphosis and lumbar lordosis. By contrast, the transition between the last, fifth lumbar vertebra and the sacrum is more prominent and is known in the literature as the lumbosacral angle. The last intervertebral disc forms an outward curve or a promontory (although the latter term is more often used to refer to the upper edge of the sacral bone) (Woźniak, 2003; Będziński, 1997; Mańko, Zubrzycki & Karpiński, 2015).

The shape of the spine as described above is a specific feature of human beings, a result of verticalisation of the body. The spinal curves bring the centre of gravity into a straight line that projects onto the base of support outlined by the feet.

When the projection of the centre of gravity of the body falls outside the outline of the feet, a person loses balance. The situation is different in animals, in which the centre of gravity of the body projects forward of the feet. This also applies to apes and monkeys, which assume an upright position only occasionally (Woźniak, 2003; Bochenek, 2010).

3. SPONDYLOLISTHESIS

Spondylolisthesis is a disease defined as a displacement of a vertebra (together with all the vertebrae lying above it) with respect to a vertebra inferior to it. Spondylolisthesis most often affects the L5–S1 and L4–L5 levels. Most authors believe that the lumbar spine, as a result of having assumed a vertical position in the course of evolution, is a mechanically weak point of the human locomotor (musculoskeletal) system, which is why spondylolisthesis most commonly occurs just in this region of the spine. Spondylolistheses of other spinal regions are rare. In the general population, the incidence of spondylolisthesis is estimated to be around 5%. The name of the condition comes from the Greek words "spondyl" – the spine and "olisthesis" – slippage. Spondylolisthesis was first described in 1782 by the Belgian obstetrician Herbinaux, who associated the unnatural displacement of the spine forward relative to the bones of the sacrum with problems during delivery (the shifted body of the L5 vertebra reduced the patency of the pelvic birth canal). Since then, spondylolisthesis has been the focus of many orthopaedic, neurological and biomechanical studies (Bartochowski, 2011; Aruna, 2002; Vadapalli, 2004; Ciupik et al., 1997; Awłasewicz, Kędzior & Krzesiński, 1997; Maurel, Lavaste & Skalli, 1997).

3.1. Determining the grade of slip in spondylolisthesis

The severity of a spondylolisthesis is assessed using the Meyerding classification:

- grade I: displacement of neighbouring vertebrae by less than 25% of the vertebral body width,
- grade II: displacement by up to 50% of the vertebral body width,
- grade III: displacement by up to 75% of the vertebral body width,
- grade IV: displacement greater than 75% or total displacement of adjacent vertebral bodies relative to one another (Yugang et al., 2015).

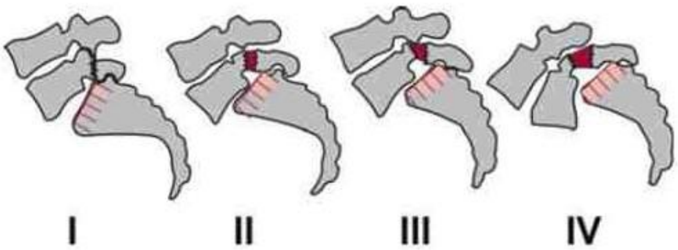


Fig. 2. The spondylolisthesis grading scale

To calculate the percentage of vertebral slip, the following formula is used:

$$P = A / B * 100\% \quad (1)$$

P – percentage of displacement;
A – length of the overhanging part of the superior vertebral body;
B – total width (anteroposterior length) of the vertebral body;

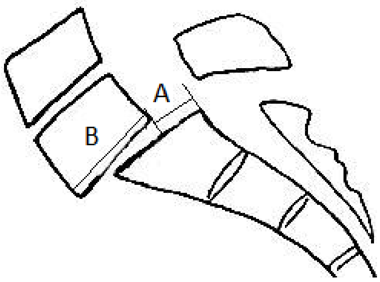


Fig. 3. Determination of grade of slip

3.2. Treatment methods

Treatment of spondylolisthesis is based on conservative and operational procedures. The doctor decides what treatment to use based on observation of the patient, radiographic assessment and the progression of the disease. The purpose of treatment is to achieve stability of the spine and prevent neurological disorders (Bartochowski, 2011). One of the main methods of treatment is to insert specially designed stabilisers called pedicle screws (Fig. 4).

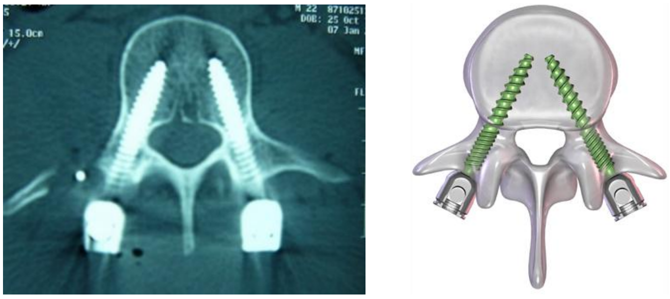


Fig. 4. Examples of use of transpedicular screws

4. DEVELOPMENT OF NUMERICAL MODELS OF VERTEBRAE

Numerical models and numerical strength tests were performed based on the results of CT scans obtained in studies done in cooperation with the Independent Public Teaching Hospital No 4 in Lublin. The tests were anonymous. The geometry of the vertebrae and their anatomical alignment were extracted using Mimics and SolidEdge software. CT images were taken in accordance with the DICOM standard. Data consisted of 226 tomograms made at a resolution of 1.25 mm. The subject was a 21-year-old man with a second grade L5-S1 spondylolisthesis.

Due to the localization of the slip, further investigations were focused on the region of the spine comprising L3, L4 and L5 vertebrae and the sacral bone (Fig. 5).

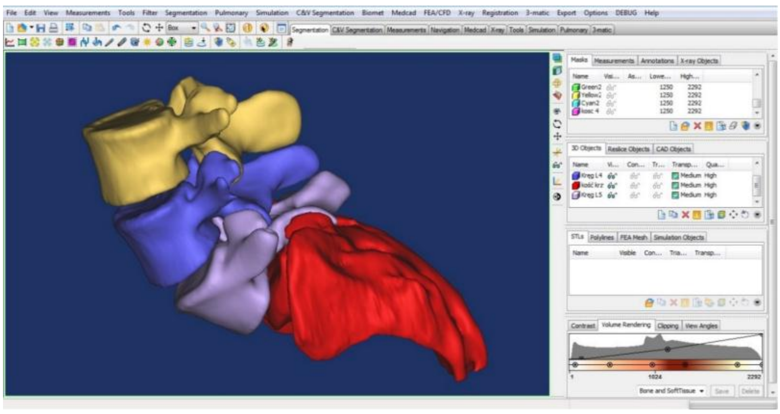


Fig. 5. Models of L3, L4, L5 vertebrae and the sacral bone (Mimics)

4.1. Simplification of numerical models of intervertebral discs and the spine before surgery

Because the CT images did not visualise intervertebral disc structures, it was necessary to create simplified models of the discs (Mańko et al., 2015; Zubrzycki & Braniewska 2017). To capture the shape of the discs and the relations between discs and vertebrae, it was necessary to simplify the earlier obtained vertebral models by levelling their upper and lower surfaces (removing the anatomical unevenness of these surfaces). After completing these steps, the Assembly module was used to develop a final model of the investigated region of the spine (Fig. 6) (Zubrzycki & Smidova, 2014).

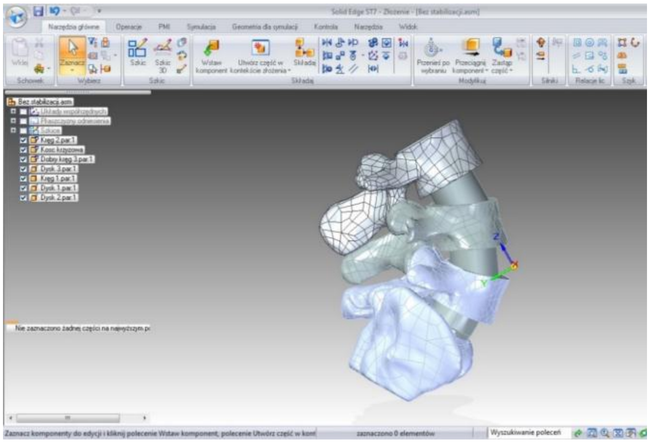


Fig. 6. Numerical model of the investigated spine segment

Numerical stress distribution tests were performed for this model.

The next step of the experiments was to develop a numerical model of a pedicle screw stabiliser. For this purpose, fixation of the screws and the entire structure of the stabiliser were designed. This article presents a simplified model of a stabiliser with straight rods. Due to the preliminary nature of the study, the curvature that should normally be given to the rods was omitted. The model of the investigated fragment of the spine with the stabiliser is shown in Fig. 7.

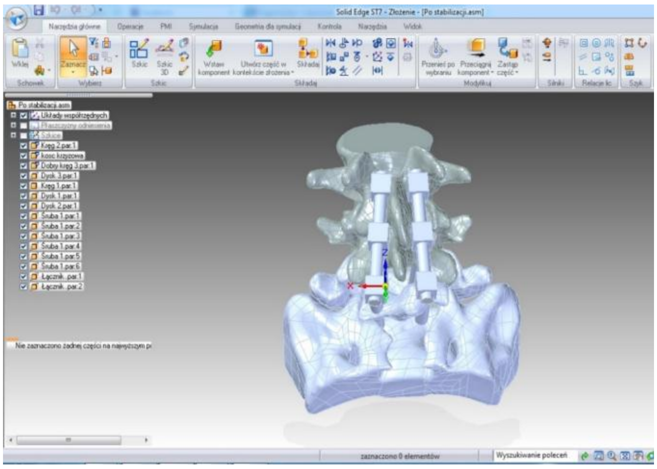


Fig. 7. Numerical model of the spine with a stabiliser

The spine was stabilised using the following elements: 6 transpedicular screws and 2 connectors. The screws had a diameter of 8 mm and a length of 55 mm and were made of Ti 6Al 4V titanium. Previous research had shown that 8 mm diameter screws could be safely used to fix the L3 vertebra. The connector had a diameter of 6.5 mm and a length of 70 mm and was made of Ti6Al 4V titanium. The screws were inserted into the vertebra at an angle of 130° relative to the xz plane, at an angle of 5° relative to the yz-axis on the left side and -5° relative to the yz-axis on the right side.

5. NUMERICAL STRENGTH TESTS

Numerical strength tests were performed using Finite Elements Analysis (FEA). The tests were performed for two cases:

1. An anatomical spine with a diagnosed spondylolisthesis,
2. A “postoperative” spine with an implanted transpedicular stabiliser.

5.1. Test assumptions

Strength analysis was performed for an upper-body load of 500 [N] applied to the top part of the L3 vertebra, which allowed us to determine the state of stress in the individual vertebrae. The model fixation was applied to the sacral bone. The material of the vertebrae and discs was assumed to be isotropic. The properties of the vertebrae and discs were determined on the basis of an analysis of available research reports (Awłasewicz et al., 1997; Maurel et al., 1997; Yugang et al. 2015; Gzik et al., 2012). The parameters are given in the table below.

Tab. 1. Properties of materials used during FEM analysis

Material	Young's Modulus [MPa]	Poisson's ratio
Vertebra	10000	0.3
Intervertebral disc	100	0.4
Screws and connectors (Ti 6Al 4V Titanium)	104800	0.31

The first test was an analysis of the physiological lumbar region using the input data as given in 5.1. The obtained stress distribution data are presented in the figures below. FEM analysis was performed according to the Huber–von Mises hypothesis.

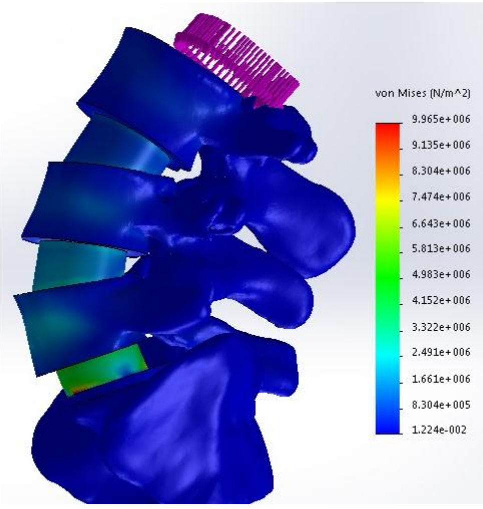


Fig. 8. Stress distribution in the model without a stabiliser – general view

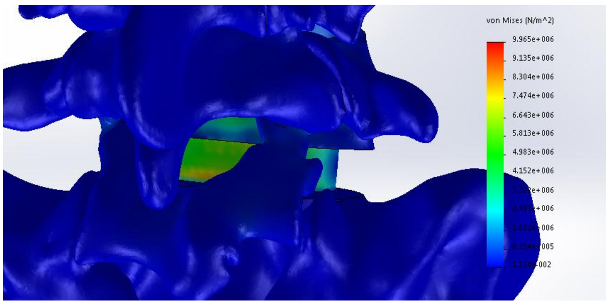
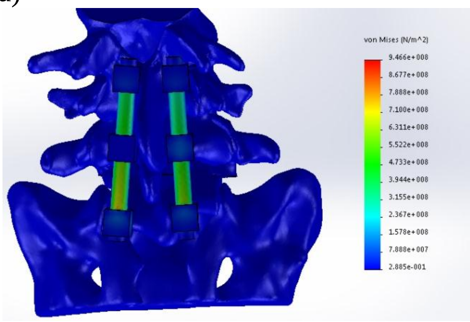


Fig. 9. Stress distribution in the model without a stabiliser – details

It is easily seen in the figures above that the stresses concentrate in the posterior and anterior parts of the intervertebral disc between the L5 vertebra and the sacral bone (9.135 MPa). In the vertebrae themselves, the stresses are not high and are evenly distributed over the entire surface of the bones. The observed stress concentrations are very dangerous to the patient because they can lead to severe damage to the disc and consequently to discopathy and further aggravation of spondylolisthesis and increased pressure on the spinal cord. In extreme cases, a patient may suffer loss of sensation (numbness) in the lower limbs and loss of walking ability.

Further analysis concerned the model of the lumbar region of the spine with the transpedicular stabilizer. An analysis of the obtained stress distribution maps showed that, in the examined element, the use of the stabiliser as a support structure substantially reduced the stresses acting on the vertebrae and the intervertebral discs (Figs. 10, 11). In the patient investigated in this study, insertion of the stabiliser to the lumbar spine transferred nearly all of the load to the stabiliser rods (Fig. 10a). In this way, the deformed parts of the spine were decompressed and could be subjected to corrective and rehabilitation care (Fig. 10b). The results show that the highest values of Huber–von Mises reduced stresses were observed in stabiliser rods (a maximum value of 946 MPa). This considerable increase in stresses in the system with the stabiliser compared to the physiological model was mainly due to the change in spinal biomechanics and was also caused by the fact that the cross sections of spinal vertebrae and intervertebral discs were much larger than those of the stabiliser rods (d = 6.5 mm, l = 70 mm) and screws (D = 8mm).

a)



b)

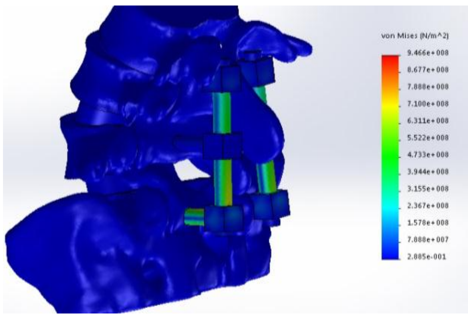


Fig. 10. Stresses acting on the vertebrae and the inter-vertebral discs

It is also important to note that high stress values were observed in stabiliser screws. This applied to screws fixed in the L4 vertebra and the sacral bone (Fig. 11). In contrast, the stresses in screws located in the L5 vertebra were negligible.

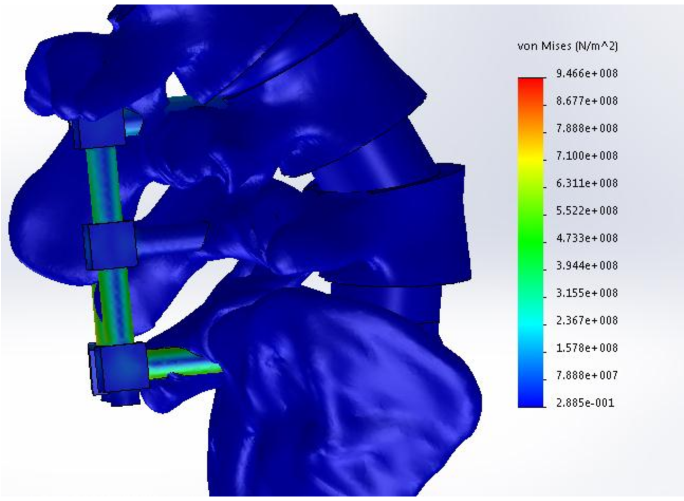


Fig. 11. Analysis of the obtained stress distribution maps

Based on the results of the simulation, it can be said that a transpedicular stabiliser can be used to decompress an injured part of the spine and to help improve the patient's quality of life.

6. CONCLUSIONS

In this study an attempt was made to analyse the complex mechanical system of a spine segment stabilised to promote healing in second-grade L5-S1 spondylolisthesis in a 21-year-old man. The effects of transpedicular stabilisation on the biomechanics of the human lumbar spine were evaluated using two different models of the spine: a physiological model and a model with a transpedicular stabiliser. The models were developed on the basis of CT images which were processed using MIMICS software. The models were then imported into the CAD environment: they consisted of vertebrae and intervertebral discs with and without a transpedicular stabiliser. They were used to perform a static analysis for a normal upper-body load.

The results of the static load simulations of the physiological model showed that, in this case of spondylolisthesis, the areas that were exposed to the largest loads were the posterior and anterior part of the intervertebral disc located between the L5 vertebra and the sacral bone. The purpose of stabilisation with transpedicular screws was to stabilise and decompress these structures. The analysis performed on the transpedicular stabilisation model showed that insertion of transpedicular screws stabilised the system and made it rigid. The L5-S1 disc was decompressed, and the stresses were taken over by screw connectors and the screws fixed in the L4 vertebra and the sacral bone.

The Finite Element Method, which is widely applied in modern engineering, was used to better understand the lumbar spine condition of spondylolisthesis and to evaluate treatment of this condition by transpedicular stabilisation. The application of this method allowed a better insight into the anatomy, physiology and kinematics of a spine with second grade L5-S1 spondylolisthesis as well as the assessment of the effectiveness of its treatment with transpedicular screws.

Despite some simplifications in mapping the reality, FEM can be used extensively for medical purposes. It allows to build patient-customized implants and prostheses, as well as providing detailed information on the biomechanics of systems in particular disease entities. It significantly accelerates the design process and ultimately reduces the price of the products offered, making them more accessible to the general public.

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