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THE INFLUENCE OF THE "PENETRATION-" AND THE "FILLING-RATIOS" ON THE PULL-OUT STRENGTH OF TRANSPEDICULAR SCREWS

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The stress field developed in the human lumbar vertebral bone due to the presence of transpedicular fixation screws is studied in an attempt to quantify the force required to pull the screw out of the bone. The study is focused on the pure pull-out phenomenon, namely the case in which the external loads act parallel to the longitudinal axis of the screw. The parameters considered are the penetration ratio, i.e. the ratio of the screw length divided by the distance from the posterior pedicle entrance site to the anterior vertebral cortex, and the filling ratio, i.e. the ratio of the major diameter of the screw divided by the transverse diameter of the pedicle. The study is carried out numerically with the aid of the finite element method. The analysis takes into account both the variation of the mechanical properties of the bone in terms of the distance from its surface as well as the geometrical details of a typical transpedicular screw. The results of the analysis are compared with existing experimental data from the literature and the comparison is very satisfactory.

Keywords: Transpedicular screws; pull-out force; penetration ratio; filling ratio; finite element method.

1. Introduction

During the last two decades, the pedicle screws became one of the most commonly used spinal instrumentation tools.[1] The transpedicular screw fixation method is particularly advisable for the treatment of spondylolytic and degenerative spondylolisthesis, trauma, and tumor due to its ability to achieve rigid spinal fixation.

In spite of the constant improvement of both the spinal instrumentation systems and the surgical techniques, there is still no foolproof method for the fixation and

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stabilization of the spine. The most serious problems encountered are the hurtful results to the spinal segments adjacent to the fixated ones and the failure of the spinal instrumentation either due to the fracture of one or more of its structural elements or due to loosening of the fit. Loss of the surgical construction stability as a result of screw loosening is a common complication, particularly in osteoporotic patients.

In order to overcome the above mentioned problems deeper knowledge is required concerning the way that each part of the spinal instrumentation system functions when implanted into the human spine. In this context the present study is focused on the behavior of the pedicle screw and on the factors that influence its mechanical behavior when it is implanted into a human lumbar vertebra and subjected to pure pull-out loads. The parameters studied are the penetration ratio and the filling ratio. The "*penetration ratio*" is the ratio of the screw length divided by the distance from the posterior pedicle entrance site to the anterior vertebral cortex. The term "*filling ratio*" or often called "*percent fill*"[2] denotes the ratio of the major diameter of the screw divided by the transverse diameter of the pedicle (small semi-axis). These two parameters constitute simple and flexible tools for the quantification of the relative dimensions of the vertebra and the pedicle screw. They are taken into serious consideration by orthopedic surgeons in case they have to decide about the most suitable pedicle screw for a specific patient. Textbooks of orthopedic surgery indicate that the penetration length should be greater than 80% while the percent fill should exceed 70%.[2]

It is clear that the pure pull-out loading mode is a rather simplified case compared to the broad spectra of loading conditions experienced by the pedicle screw when it is implanted into the vertebra. Although pure pull-out loads are generated *in vivo* during procedures of spinal correction (remedy of spondylolisthesis), in general the pull-out forces and displacements are often applied in combination with other types of loads on the pedicle screw. However, in spite of its simplicity the study of the pure pull-out case enlightens some controversial points concerning the way two totally different materials (human bone and metal) "cooperate" for the distribution of the stresses developed, the way they react to the separation from each other and the factors that influence their strength.

Numerous studies are found in the literature dealing with the pull-out strength of bone screws in general[3–8] and of pedicle screws in particular.[9–20] These studies have been conducted either experimentally[3–16] or numerically.[17–22] The experimental approach provides indisputable data about the characteristics and the mechanical behavior of the specimens, assuming that the experiments are conducted properly. The limitation of the method emanates from the inevitable variation of the mechanical properties of the specimens (i.e. human vertebras) and the difficulty to achieve reproducibility of the results. In many cases this problem is answered by substituting the biological material by artificial bone. Even though simplified, this experimental approach can lead to very useful results.

On the other hand the numerical approach provides *de facto* reproducibility of the results but it is difficult to ensure that the model designed simulates accurately enough the *in vivo* conditions. Therefore it is absolutely necessary to evaluate the accuracy and reliability of the results by using suitable experimental data, as it is true for any numerical analysis. Especially in case one models numerically human tissues the clinical experience should be taken into serious consideration, also.

2. Materials and Methods

For the purposes of the present analysis a finite element (FE) model of a typical commercial available pedicle screw and a human lumbar vertebra was designed. The analysis is conducted in two steps:

- Initially, the pedicle screw and only the part of the bone, which surrounds the screw, were modeled. This model was studied independently in order to minimize the mesh dependence phenomenon and to optimize the density of the mesh.
- During the second step a more realistic model of the vertebra as a whole was designed, which incorporated the model of the previous stage. Special attention was paid to ensure continuity of the stress and strain fields all over the mass of the vertebra.

In other words the approach adopted models separately the interface region and the region where the boundary conditions are applied and combines the two models achieving processing time economy and accuracy of the results. The analysis was performed using the ANSYS 9 software.

2.1. *Design of the FE model of the pedicle screw and of its surrounding bone*

The model of the screw takes into account the main characteristics of commercially available pedicle screws. The geometry of the screw as well as the direction of the pull-out loads are shown in Fig. 1. Special attention was paid to the accurate description of the curved parts of the threads as well as of their helicoidal shape. The geometrical quantities describing the thread of a typical pedicle screw are: The major or external radius, r_1, the minor or internal radius, r_2, the pitch p, the thickness of the thread at its peak, e, the radius r_3, and the inclination of the thread, described by the two angles a_1 and a_2. The values of these quantities for the original pedicle screw studied were equal to: $r_1 = 2.75\,\text{mm}$, $r_2 = 1.7\,\text{mm}$, $r_3 = 0.3\,\text{mm}$, $p = 3\,\text{mm}$, $e = 0.1\,\text{mm}$, $a_1 = 5^\circ$, and $a_2 = 25^\circ$. The screw was designed with nine threads yielding an overall length L of 27 mm.

Taking into account the fact that Young's modulus and the yield stress of the materials of the commercial pedicle screws (Ti alloys, stainless steel) are significantly higher compared to those of the vertebral bone it can be assumed that the behavior of the screw is that of an absolutely rigid body. Similar conclusions were

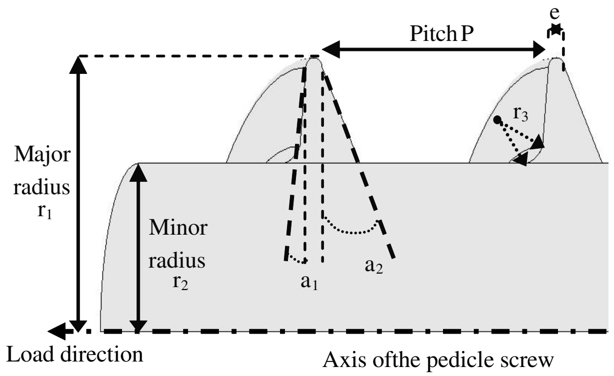

Fig. 1. The geometry and the main characteristics of a typical pedicle screw.

drawn from a recent study of the pull-out phenomenon[19] in which the maximum stress developed in the vertebra body was calculated to about 45% of the respective yield stress while the maximum stress in the screw was 1% of the yield stress of the titanium alloy. The above assumption is supported, also, by many references that report significantly lower deformations of the pedicle screw compared to those of the vertebral bone[17,18,23] and significantly lower stresses compared to the yield stress of its material.

Concerning the FE model of the vertebra, it is easily concluded that the area which surrounds the screw is more likely to experience the highest deformations and strongest stress concentrations, compared to the remaining part of the vertebra. Therefore as a first step only the portion of the vertebra in the immediate vicinity of the screw was designed, in order to optimize the mesh and the boundary conditions on the intersection between screw and vertebral bone and to achieve better accuracy of the numerical results. In this context a "*cylinder*" of vertebral bone surrounding the pedicle screw was designed (Fig. 2). The threaded hole along the axis of the cylinder, where the screw is driven, was assumed to have identical shape and dimensions with the screw itself, ignoring in this way any pre-stress phenomena. For simplicity and processing time economy, symmetry of the system with respect to a plane including the axis of the screw was considered and therefore only one half of the pedicle screw–vertebral bone model was modeled.

Special attention was paid to the optimum simulation of the bone–screw interface, which defines the load transfer mechanism from the screw to the bone.[24] This interface can be designed in two different ways; either by assuming that the bone and the screw are fully bonded or that they are in simple contact to each other. The first one corresponds to the long term conditions, about two years after the insertion of the pedicle screw into the body of the patient, while the second one to the conditions during the first postoperative weeks. For the purposes of the present study the interface conditions adopted were these of simple contact. The decision

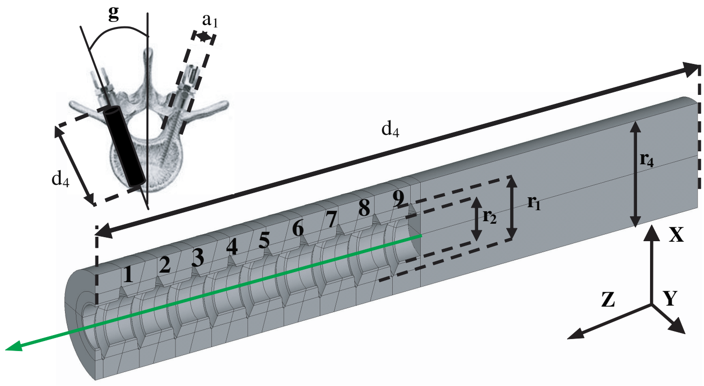

Fig. 2. The geometry of the FE model of the vertebral bone a_1 is the small semi axis of the pedicle, d_4 the distance from the posterior pedicle entrance site to the anterior vertebral cortex along the pedicle axis, $r_4 = r_2 + 0.5(a_1/2 - r_2)$.

was based on the fact that the pedicle screw–vertebral bone system is more likely to fail when the screw and the bone are in simple contact to each other. In addition simple contact interface conditions permit more accurate simulation of the experiments that are performed *in vitro* and also direct comparison of the respective numerical and experimental results.

The volumes of the vertebral bone were meshed using the eight node 3D solid element SOLID185 while the areas of possible contact on the intersection between bone and screw were meshed with CONTA173. For the screw only the areas of the intersection were meshed using TARGE170 elements. The degrees of freedom of the TARGE170 elements were constraint using a master node in order to create the rigid body behavior of the screw. Special attention was paid in order to create a uniform mesh near the contact areas.

At this stage of the analysis it was assumed that the vertebral bone is homogenous, isotropic, elastic-perfectly plastic material with a constant friction coefficient. The modulus of elasticity was considered to be 100 MPa, Poisson's ratio 0.2, the yield stress 2 MPa, and the friction coefficient between screw and bone 0.2. The previous material properties were taken from the literature.[17,18,24]

The cylinder of vertebral bone was properly constrained and a pure pull-out displacement equal to 0.02 mm was applied to the screw. The specific value of the pull-out displacement was determined by assuming that the equivalent stress developed should not exceed the respective yield stress at any point of the system.

To ensure that the behavior of the model is mesh-independent the final mesh of the FE model was created following an optimization procedure. Preliminary "runs" were performed using different element sizes until convergence of the results was achieved. It was concluded that a total element number higher than 34,000 was sufficient (Fig. 3).

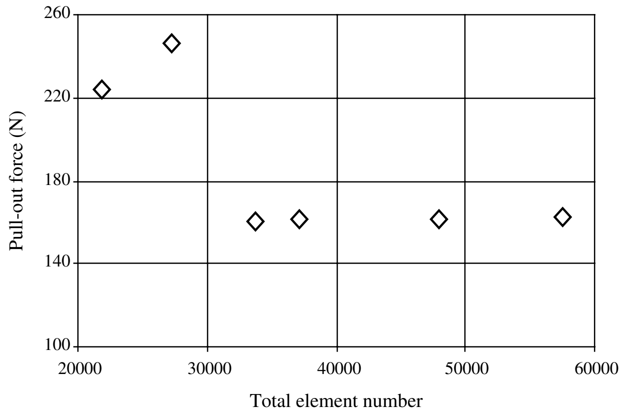

Fig. 3. The pull-out force (that produces pull-out displacement equal to 0.02 mm) for FE models with different total element number. Convergence of the results is achieved for a total element number higher than 34000.

2.2. *Design of the FE model of the vertebra*

The body and the pedicle of the human lumbar vertebra were simulated with elliptical shaped volumes, as it can be seen in Fig. 4. The part of the vertebra that was modeled is defined by the sagittal plane along the axis of the pedicle screw. The

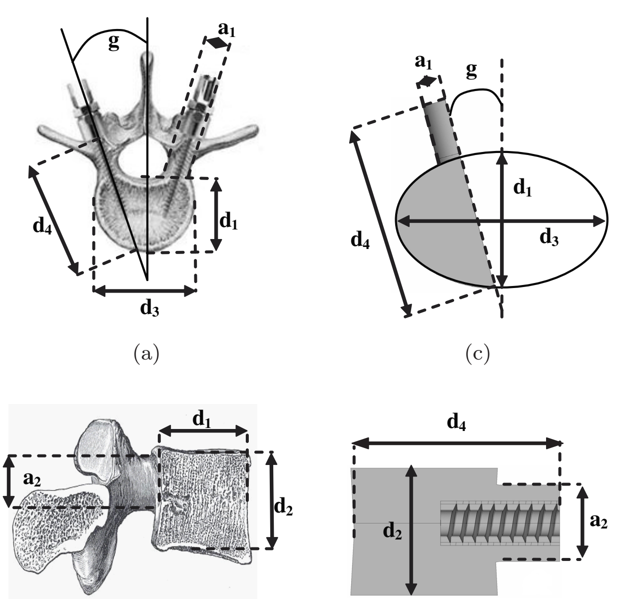

Fig. 4. The basic geometrical characteristics of the human lumbar vertebra and their simulation with the FE model.

Table 1. The geometrical quantities which define the geometry of the body and the pedicle of the human lumbar vertebra.[25,26]

Geometrical quantity	Size
Body length (d_1)	35 mm
Body depth (d_2)	30 mm
Body width (d_3)	52 mm
Distance from the posterior pedicle entrance site to the anterior vertebral cortex (d_4)	50 mm
Transverse diameter of the pedicle (small semiaxis) (a_1)	12 mm
Sagittal diameter of the pedicle (big semiaxis) (a_2)	18 mm
Transverse pedicle angle (g)	16.7°

values of the geometrical quantities used were taken from the literature[25,26] and are recapitulated in Table 1. As a next step the volume that simulates the vertebra was combined with the model of the pedicle screw and of its surrounding material, which was previously described, as it is shown in Fig. 5. Therefore it can be safely concluded that the contact analysis that will take place on the contact interface between bone and screw will be mesh independent and accurate.

Concerning its mechanical properties the vertebra was assumed to consist of cortical, subcortical, and cancelous bone. The partition of the vertebra into these three regions (Fig. 6) was based on measurements of the bone mineral density (BMD) made by Hirano *et al.*,[12] who measured the BMD at certain bisections of the body and the pedicle and gave the percentage of the total area that is occupied by each material. The values of the BMD were then used for the estimation of Young's modulus for each of the three materials. In order to achieve this, the corresponding expression given by Kopperdahl *et al.*[27] was applied. The material properties that were finally obtained are presented in Table 2, while the distribution of the different materials into the final model is shown in Fig. 7.

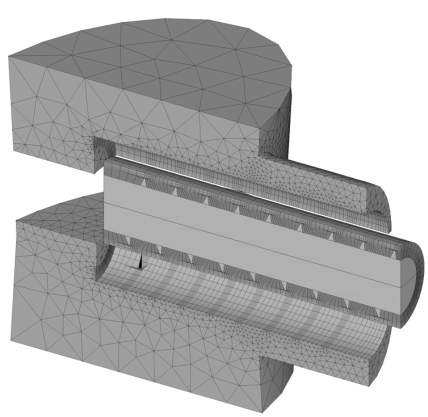

Fig. 5. The insertion of the model of the screw and its surrounding material in the model of the vertebra.

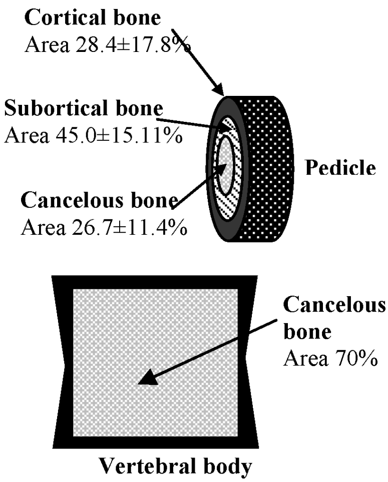

Fig. 6. The partition of the body and pedicle of the human lumbar spine into three regions of different bone mineral density and consequently of different material properties.[12]

Table 2. The BMD and the respective mechanical properties of the cortical, subcortical, and cancelous bone.

	Cortical	Subcortical	Cancelous	Reference
Bone mineral density (mg/cm^3)	856	423	175	12
Modulus of elasticity (MPa)	2770	1370	56	26
Poisson's ratio	0.3	0.3	0.2	17, 18

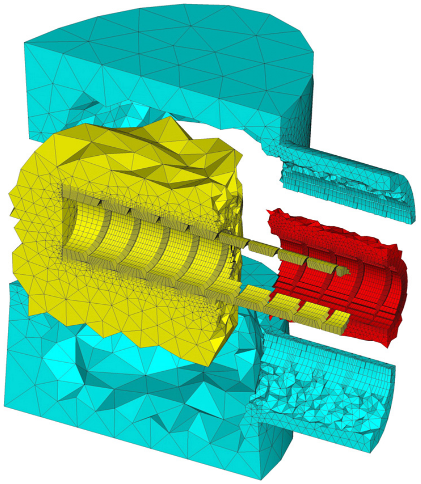

Fig. 7. The areas of different material properties into the body and the pedicle of the FE model of the vertebra.

The upper and lower areas of the FE model of the body were fully constrained while a pull-out displacement was induced to the screw. The value of this displacement was again 0.02 mm.

The above described FE model was used to study the influence of the relative dimensions of the pedicle screw on the behavior of the vertebra–pedicle screw system. In order to study the influence of the penetration ratio of the screw, various FE models were designed with lengths (L) varying from 6% to 87% of the total distance from the posterior pedicle entrance site to the anterior vertebral cortex (d_4).

In addition the significance of the percent fill was investigated. For this purpose the major radius of the pedicle screw was changed yielding values of percent fill varying from 46% to 83%. The length of the pedicle screw was kept constant.

2.3. *Evaluation of the FE model*

At this stage and before carrying out any numerical analysis it was judged necessary to assess the reliability of the approach followed. Obviously the best way to do that is to compare the numerical results with experimental data. Unfortunately it is very difficult to use experimental data from mechanical tests on biological materials, because of the significant uncertainty about their mechanical properties. A more realistic approach is to use data from experiments where the biological material is substituted by an artificial one. In this direction the experimental work of Conrad *et al.*,[3] who carried out a series of pull-out tests using different types of screws, was chosen. The screws were placed into blocks of rigid polyurethane foam and the pull-out force was measured. This foam (Last-a-Foam FR 3710) had a density of about 0.160 g/cm^3, modulus of elasticity equal to 57 MPa, and yield stress equal to 2.2 MPa.

The FE model of the pedicle screw and of its surrounding material previously described was adapted to the geometry of the pedicle screws (Table 3) and to the material properties of the foam used by Conrad *et al.*[3] The screw-surrounding material system was embedded into a model that simulates accurately the foam block and the boundary conditions of the specific pull-out tests, as it is shown in Fig. 8. A pure pull-out displacement was then induced to the screw until the yield of the foam and in this way the pull-out force was estimated numerically.

For the two cases of pedicle screws described in Table 3, the maximum pull-out forces were estimated and their mean value was compared to the respective one given by Conrad *et al.*[3] The results of the comparison are shown in Fig. 9. It is

Table 3. The screws modeled in order to compare the numerical results with the experimental ones by Conrad *et al.*[3]

Manufacturer description	Screw design (mm)	Screw length (mm)	Thread length (mm)	Major radius (mm)	Minus radius (mm)	Pitch (mm)
4 × 12-can	ST	12	8.35	2	1.6	1.85
4.5 × 12-cor				2.25		1.30

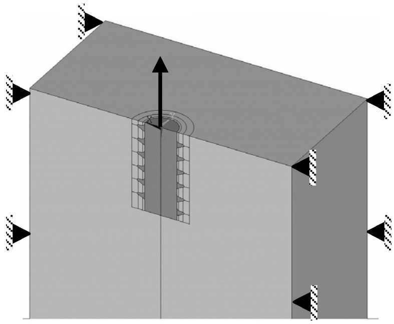

Fig. 8. The model simulating the tests by Conrad *et al.*[3] that was used to assess the reliability of the present numerical approach.

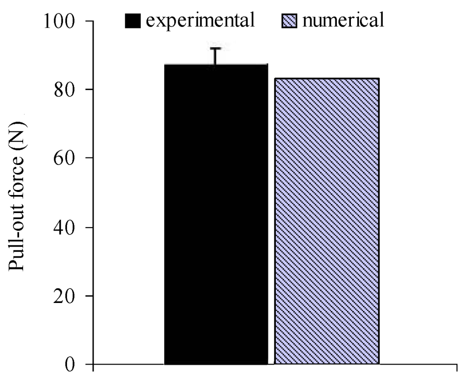

Fig. 9. Experimentally and numerically estimated mean pull-out force for screws with $L = 12\,\text{mm}$ (ST 4×12-can, ST 4.5×12-cor).

seen from this figure that the numerical results are in very good agreement with the experimental ones and within the tolerance of the experimental error.

3. Results of the Numerical Analysis — Parametric Study

The distribution of the Von Mises equivalent stress into the vertebra is plotted in Fig. 10. The stress concentrations around the edges of the threads are clearly visible, indicating the critical role of the shape of the thread edge in the overall performance of the transpedicular screws. However, what is perhaps more important is the lack of uniformity of the equivalent stress as one proceeds towards the innermost threads. This nonuniformity is much clearer in Fig. 11 in which the contact pressure between the vertebral bone and the pedicle screw is plotted. It is clearly seen from this figure that the contact pressure is maximized somewhere in the intermediate threads of the screws.

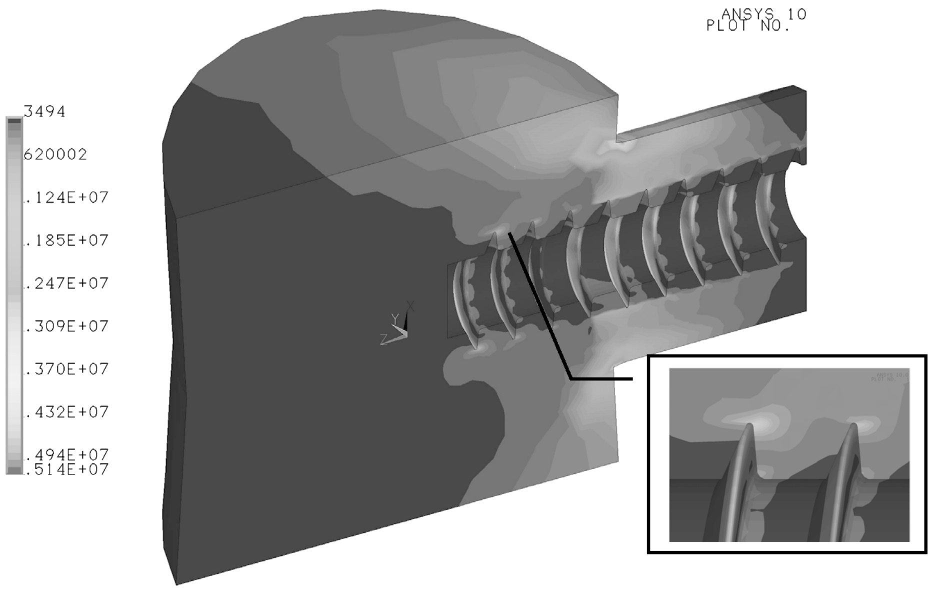

Fig. 10. The distribution of the Von Mises equivalent stress into the FE model of the human lumbar vertebra.

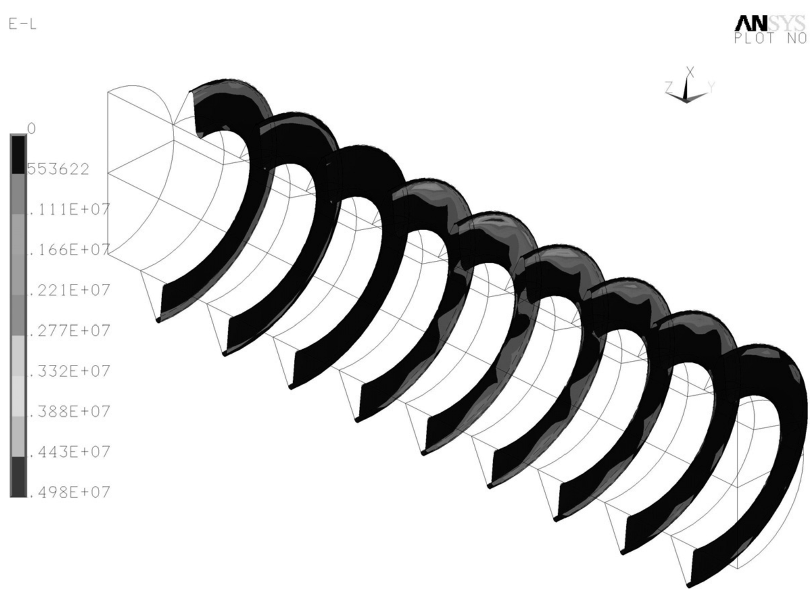

Fig. 11. The contact pressure between vertebral bone and pedicle screw.

3.1. *Investigation of the influence of the penetration ratio on the pull-out force*

During the first part of the parametric study the pull-out force for a displacement equal to 0.02 mm was estimated for pedicle screws with different penetration ratios. According to the results the pull-out force increases in a nonlinear way with increasing penetration ratio. As it is shown in Fig. 12 the pull-out force increases significantly until the screw length approaches the corresponding length of the pedicle. Significant increase appears also when the length of the screw is big enough to approach the anterior cortex of the body. On the contrary, an intermediate region appears, between these two areas, where the pull-out force remains practically constant.

These results reveal the important role of the pedicle and of the anterior cortex for the stability of the pedicle screw–vertebra system. However, in order to obtain a more complete view it is useful to study the way the pull-out force is distributed along the pedicle screw. In this context the force sustained by each thread is plotted in Fig. 13 for three screws of different penetration ratios:

- The shortest one which corresponds to a percent penetration ratio equal to 17% consists of only three threads and all of them are placed into the pedicle. The force is undertaken mainly by the innermost thread.
- The second pedicle screw, which corresponds to a percent penetration ratio equal to 52% has nine threads, from which the three innermost ones (7–9th) are anchored into the cancelous bone of the vertebral body. The decrease of the contribution of these threads compared to those which are placed into the pedicle is obvious.

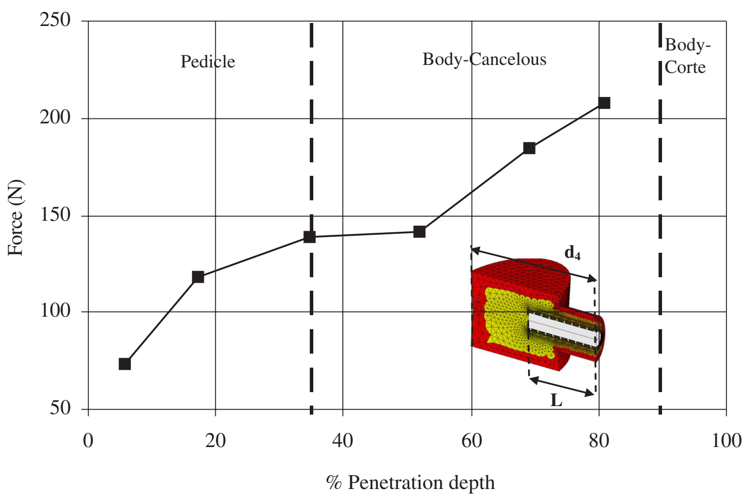

Fig. 12. The pull-out force for a displacement equal to 0.02 mm for pedicle screws of different penetration depths (% L/d_4).

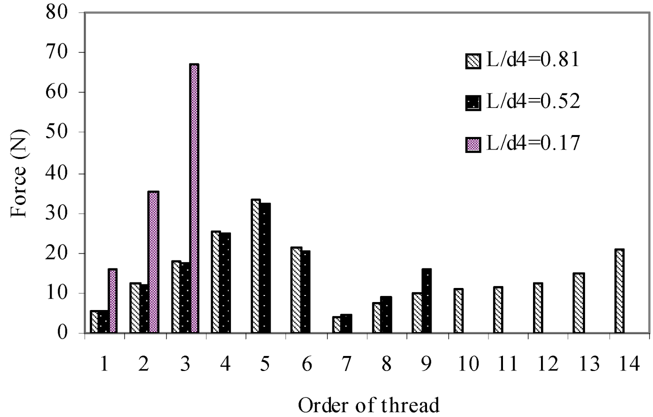

Fig. 13. The distribution of the pull-out force on each thread for three pedicle screws of different lengths.

- Finally the third and longest pedicle screw studied had 14 threads and percent penetration ratio equal to 81%. The first six threads were placed into the pedicle (1–6th) while the remaining eight ones into the cancelous bone of the vertebral body. The innermost one of the threads touches the anterior vertebral cortex. One can notice that the force on the threads, which are into the body, tends to stabilize before it increases again at the innermost thread.

The distribution of the pull-out force confirms among others the important role of the pedicle. For the three cases of pedicle screws, which had length bigger than the respective of the pedicle, the percentage of the pull-out load carried by the pedicle varies between 79% ($L/d_4 = 0.52$) and 56% ($L/d_4 = 0.81$) of the total pull-out load.

The results previously presented indicate that the length of the pedicle screw is a very important factor for the stability of the whole pedicle screw–vertebra system. The ideal value of the length seems to be the one that permits the maximum possible exploitation of the strong materials of the pedicle and the cortex of the body, with the minimum possible risk for the patient.

During the second part of the parametric analysis simulations were performed for different values of the percent fill. The estimated pull-out force for the applied displacement (0.02 mm) is plotted in Fig. 14 versus the corresponding percent fill. The sigmoidal nature of the variation is clearly shown: The pull-out force increases rapidly when the radius of the pedicle screw becomes big enough to reach stronger materials. Especially for percent fill higher than 75% one can notice the significant increase of the pull-out force even though the threads of the screw are not anchored into the cortical bone. Therefore it can be concluded that the major radius of the pedicle screw should be big enough in order to take advantage of the strong subcortical and cortical bone.

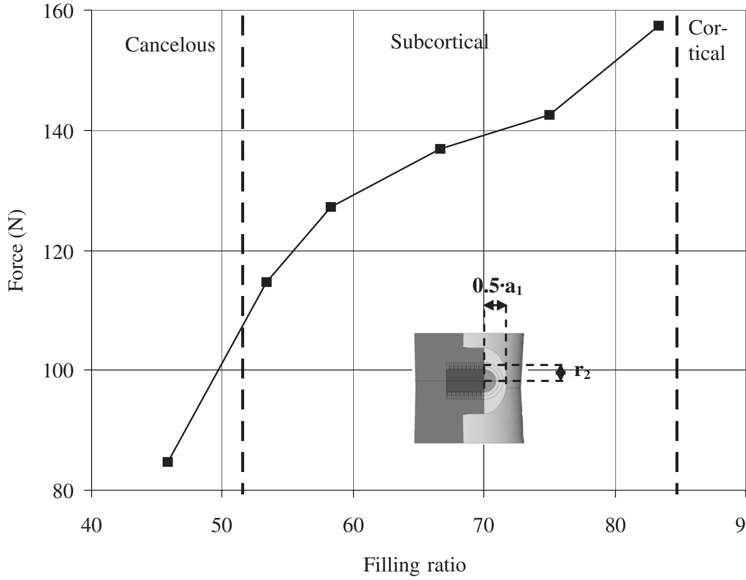

Fig. 14. The pull-out force at displacement equal to 0.02 mm for different percent fills (% $2r_2/a_1$).

In quantitative terms the current FE analysis indicates, that for the particular simplified material properties and geometry of the vertebra, the percent fill should be higher than 75%. Concerning the upper limit of the percent fill it is not possible to accurately determine using a pure pull-out load case. This happens because extremely high values of the percent fill (exceeding 85%) even though could lead to higher pull-out strength could, also, increase the possibility of fracture during different load cases (i.e. flexion-extension).

4. Discussion

The target of the present analysis is the study of the behavior of the pedicle screw–lumbar vertebra system, when it is subjected to pure pull-out loads and the investigation of certain factors that influence its mechanical behavior. For this purpose an accurate 3D FE model of a standard, commercially available, pedicle screw and a simplified one of the human lumbar vertebra was designed. Special attention was paid to the optimum simulation of the contact conditions on the interface between bone and screw. The contact conditions were chosen (instead of the much simpler fully bonded ones) because they produce a more realistic simulation of the worst scenario and they simulate better existing *in vitro* experiments. The most serious disadvantage of the adoption of contact conditions is that an additional nonlinearity is introduced into the problem. In order to address this nonlinearity a contact analysis is necessary for the accuracy of which it is very important to design a fine mesh as uniform as possible especially in the vicinity of the contact area. To fulfill this demand a novel two-step approach was adopted during modeling as it is described in Sec. 2.

The main limitation of the present study lies in the use of BMD measurements of the pedicle and the body of the human lumbar spine in order to represent the inhomogeneity of the mechanical properties of the vertebra. Clearly this simulation of the material properties and the internal architecture of the vertebra is not suitable for accurate quantitative estimations. However, it can be used for the qualitative and comparative investigation of the pullout phenomenon.

Another limitation of the present study stems from the imposition of symmetry conditions across a plane containing the axis of the screw. However, for the case where a screw is subjected to pure pullout loads the three-dimensional helicoidal shape of the threads seems to play a minor role concerning the screw's mechanical behavior. This assumption is supported by many researchers in the literature.[17,18,21,22] Especially Tafreshi *et al.*[21] have studied the mechanical behavior of "drillstring" threaded connections using the FE method. The authors considered three different loading cases, namely pure pullout, bending and torsion. For each case they designed different FE models including axisymmetry, symmetry along one plane and no symmetry at all, respectively. They concluded that the helicoidal shape of the threads could be ignored for the pullout case and suggested that a three-dimensional model is necessary only for the bending and torsion loading cases. In the same context Zhang *et al.*[17,18] considered symmetry across two planes containing the axis of the screw and finally modeled only one fourth of the screw.

The validity of the assumption that the helicoidal shape of the threads can be ignored during the numerical simulation of pure pullout phenomena is also supported by the very good agreement between the numerical results of the present study and experimental results from the literature.[3] In addition in a resent work,[20] the pullout problem was studied both experimentally and numerically and even though the helicoidal shape of the threads was totally ignored, by imposing axisymmetrical boundary conditions, the results were remarkably close to the respective experimental ones.

Coming to an end and concerning the symmetry assumption it can be said that it is acceptable for relatively simple loading cases, as it is pure pullout, however, it restricts the application of the present FE model for more complex loading cases approaching those applied to the pedicle screw *in vivo*.

Finally a last limitation is related to the simplified geometry of the vertebra adopted. However, this simplification was necessary since it permitted better control of the parameters of the study. In any case since attention is focused in the immediate vicinity of the screw–bone interface it is expected that this simplification does not seriously mask the phenomena studied.

The parametric investigation of the relative length of the screw showed that the ideal value of the penetration ratio seems to be the one that permits the maximum possible exploitation of the strong materials of the pedicle and the cortex of the body, with the minimum possible risk for the patient. If one attempts a quantitative estimation, the penetration ratio should be bigger than 40% so that the pedicle screw is anchored through the entire pedicle. In the literature one comes across controversial experimental and numerical results about the influence of the penetration ratio on the pull-out strength. Zindrick *et al.*[16] tested pedicle screws implanted into cadaveric lumbosacral vertebras subjected to pure pull-out. The authors indicate that there is no significant difference in the pull-out force for penetration ratios bigger than 50%. On the contrary Heller *et al.*[11] indicate that bicortical purchase can significantly increase the pull-out strength. In addition, the numerical analyses of Zhang *et al.*[17,18] showed a linear increase of the pull-out force with increasing screw length. However, the results of these numerical studies seem to be influenced by the assumption than the bone is homogeneous and isotropic.

The only indisputable fact is that the pedicle plays a crucial role for the pull-out resistance of the pedicle screw–vertebra system. After a series of pull-out experiments it was concluded by Hirano *et al.*[12] that the 60% of the pull-out strength of the pedicle screw depends on the pedicle rather on the body of the vertebra. The estimation of the current numerical analysis was that the pedicle sustains about 55–80% of the total pull-out force (excluding very short screws fully mounted in the pedicle) which is in very good agreement with the experimental findings of Hirano *et al.*[12]

The parametric investigation of the percent fill showed that the major radius of the pedicle screw should be big enough in order to reach the areas of subcortical and cortical bone, which are significantly stronger materials compared to the cancelous bone. In the literature it is reported that the percent fill should be higher than 70–80%.[2] These values are consistent with the results of the present parametric study, which indicated a significant increase of the pull-out load for percent fill higher than 75%.

At this point it is important to note that the parametric study of the penetration ratio was performed for constant values of the percent fill (75%) and the parametric study of the percent fill was performed for constant penetration ratio (52%). As a next step the combined study of the mutual interaction of these two quantities appears to be of great importance. Indeed preliminary results available indicate that this interaction should not be ignored.

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