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Research Article

Numerical Simulation on the Effects of the Bone Quality and the Insertion Type for the Pedicle Screws Pull Out Strength

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Abstract

Post-surgical complications such as screw loosening due to fatigue loading and screw breakage still need investigations. Clinical parameters such as the screw insertion type and depth, the bone density and the patient degree of mobility, greatly affect the mechanisms of the implant's failure/success.

The current finite element study focused on the prediction of the pedicle screw pull out strength under various conditions such as; insertion type , bone quality and loading mode.

As portrayed in this study, the preservation of the pedicle cortex as in the N1 insertion technique and the screw anchoring depth gearing a maximum number of threads promote a better protection against premature breakouts of pedicle screws Gearing a maximum number of threads promoted a better protection against screw breakout.

In agreement with experimental data, the type of insertion in which, the first screw thread is placed immediately after the preserved pedicle cortex, showed the best pullout resistance for both normal and osteoporotic bone.

The study confirmed the significance of bone density in spine fixation procedures. The pull out strength computed for normal bone drastically dropped in osteoporotic bone. That is, low-density bone characterizing elderly is found to have most of its pull out strength (~65%) lost with age. The use of acrylic cement and pedicle screws with special hydroxyl apatite coating may strengthen in that case the anchoring sites and improve the pull out strength of such implants.

Keywords

Finite element, pullout strength, spinal fracture, pedicle screws, spinal fixation,Osteoporosis.

Declaration of Conflicting Interest

The author(s) declared no potential conflicts of interest with respect to the research, authorship, and/or publication of this article.

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1.Introduction

The critical location of the screws near the main nerve roots makes these elements very delicate in spine surgery causing serious and traumatic postoperative complications to the patient when inserted inaccurately [1]. These pullout efforts vary in terms of amplitude and rate of loading depending upon the patient's mobility.

They might be either slow, low in magnitude leading to static loading conditions or fast and high in magnitude as in the case of dynamic or impact loading conditions. The consequence of these loading conditions on the bone at the vicinity of pedicle screw anchorage is devastating, especially in the early months of stabilization. The need to study the bonding mechanisms between the pedicle screws and vertebra requires a close collaboration between orthopedics surgeons and engineers [2].

The stress distribution in bone was predicted and expected to help understand the mechanical behavior of the screw during the tearing process, under the influence of the above-mentioned geometric, physiological and surgical parameters [3], [4], [5]. Waqas A.L and al [6] proposed an method to investigate bone quality

2. Material and Methods

In order to generate the 2D Computer Assisted Design (CAD) model, a portion of the vertebra and the screw were isolated at the screw centreline level in order to keep the axis of symmetry of the screw, through which the pull out efforts are transmitted. This permits to abridge a complex 3D structure into a simpler 2D axisymmetric model that is accurately representative of an adult lumbar vertebra.

2.1 Geometric considerations

To reflect the multi-scale aspect of the bone, a partition into three areas was made during model development, see **Figure 1a**. This consideration was helpful in the implementation of different mechanical properties for the cortical, subcortical and trabecular bone structures, respectively [7]. In all model investigations, the cortical bone thickness was set equal to one (1) millimeter [8]. It is to be noted that the pedicle is made only of cortical and subcortical bone tissues [9].

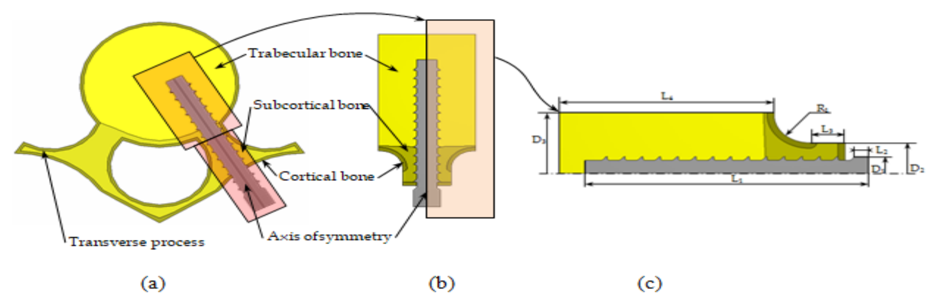

Figure 1a Extraction of a 2D axisymmetric model for the screw and surrounding bone structures: (a) global model of the vertebra and pedicle screw, (b) axisymmetric model boundary delimitation, (c) significant dimensions exploited in the development of the 2D axisymmetric model.

While **Figure 1b** shows the boundary of the bone domain commonly called area of influence where the bone is affected by the implant's tearing and loosening, **Figure 1c** describes and sets all parameters pertaining to the development of the bone and screw axisymmetric model. Those parameters are fully described and listed in **Table 1**. In order to make the most of the axisymmetric geometry at the vertebral level, we proceed to the virtual removal of the vertebral transverse process while retaining its effect in the form of coupling equations implemented during the computational works, this issue is discussed in more details later on in the manuscript.

Parameter	Description	Value (mm)
L1	Screw length	45
L2	Screw head length	5
L3	Pedicle length	5
L4	Length of the vertebral body specimen	34
D1	Screw diameter	6
D2	Pedicle diameter	11
D3	Diameter of the vertebral body specimen	22
R1	Pedicle to vertebral body connector	5

Table 1 Description and value for the parameters used in the 2D CAD model generation.

The screws used in orthopaedic surgery have very intricate shapes and dimensions [10]. The major dimensions for the pedicle screw shown in figure 2 were set according to the British standards [14] and listed in **Table 2**.

d1	Q	a1	a2	r1	r2	e	P
6	0.7	3°	35°	0.8	0.3	0.2	2.5

Table 2 Geometric parameters for the screw design in millimeters.

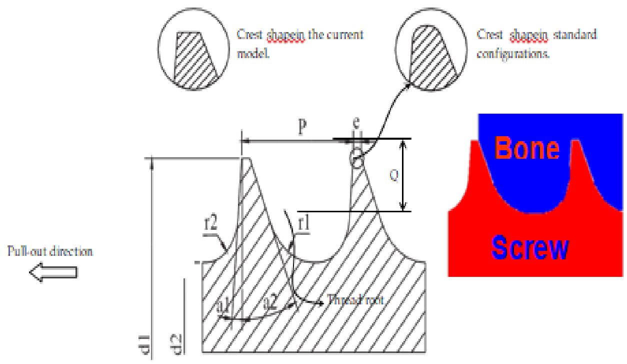

Figure 2 Detailed screw parameters associated with the 2D axisymmetric CAD model.

2.2 Stress-strain relationship for vertebral bone structures

The mechanical behaviour of the human vertebral cortical, subcortical and trabecular bone structures is described in several studies , [11]. To simulate high loading conditions leading bone failure we opted for an elastic-perfectly plastic behaviour that describe the stress-strain relationships in the elastic zone and perfect plastic zone to end up to abrupt failure [15].. All the parameters associated with the the stress-strain relationship for the cortical, subcortical and trabecular bone structures, in their normal and osteoporotic conditions are grouped in **Table 3**. The young's modulus (E), the plastic yield stress (σ_p) and strain (ε_p) were deduced from density values based on empirical relations .Also listed in this table are, the mechanical properties of the titanium alloy used for the pedicle screws.

		ρ (g/cm3)	E (MPa)	ν	σ_p (MPa)	ε_p %
Normal Bone	Cortical	1.6	12000	0.3	100	3
	Subcortical	0.5	360	0.3	14.5	20
	Trabecular	0.2	100	0.2	3	40
Ostoporotic Bone	Cortical	1	2900	0.3	40	3
	Subcortical	0.3	78	0.3	6	20
	Trabecular	0.13	75	0.2	1.5	40
	Screws (Titanium)	-	124000	0.3	-	-

Table 3 Mechanical properties adopted for normal and osteoporotic bone structures as well as titanium used for pedicle screws.

2.3 Mesh generation and boundary conditions

The discretization process of the model passed through critical steps prior to mesh generation. A preliminary investigation was performed to find the optimal mesh parameters for each of the partitions shown in **Figure 1**. That is, in this phase, a minimal number of elements and nodes was sought to achieve appropriate convergence of the computed results. The grid density was then increased to account for higher stress gradient occurring at the screw-bone interface, an intricate area responsible for the load transfer (see **Figure 3a**).

The linear axisymmetric 4-nodes quadrilateral elements were preferred to their linear triangular counterparts due to their ability in producing regular meshes on both sides of the interacting surfaces. Such a regular and identical mesh at the screw-bone interface for the various models permits a better comparison platform of the computed results. A mesh refinement characterized by a minimal element size of 0.1 mm was implemented in such areas, leading to a total number of nodes and elements for all models around 28000 and 27000, respectively.

By their morphological aspect, highly porous vertebrae raise a problem when it comes to compute their contact stresses and load transfer mechanisms while being in contact with an extremely stiffer material, namely, the titanium. Special attention was focused on modeling the contact at the screw-bone interface. To simulate a tensile loading condition, a prescribed displacement was imposed on the screw head via a Multi Point Constraint (MPC) coupling condition.
The coupling condition consists of selecting the nodes on the flat area of the screw head to be coupled to

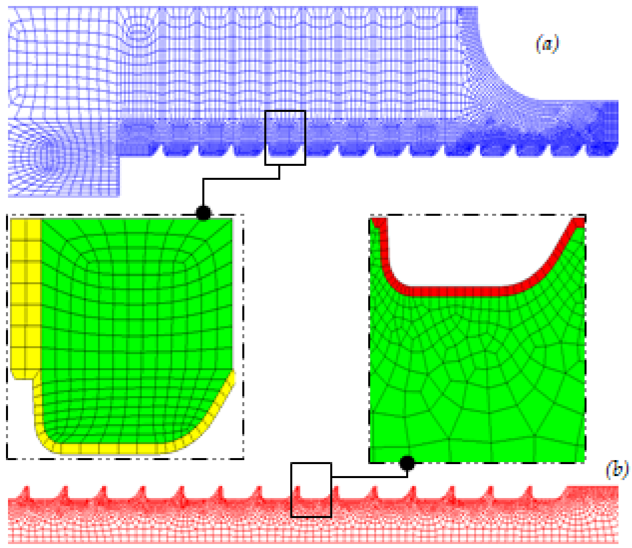

Figure 3 A closer view to the mesh refinement applied at the interface between bone (a) and screw (b).

a reference point (RP) located on the screw's centreline to which a movement is prescribed to reproduce the tensile test. To get closer to the ultimate conditions that supposedly lead to implant failure, large deformation formulation was made active in ABAQUS explicit solver and trial loadings were carried out to establish the amount of load that induces large deformations on the bone structure. A 500μm shift applied to the RP at a rate of 2mm/min was found to trigger such behavior and was adopted for the future large deformation analyses. On the opposite, using the default ABAQUS standard solver, a 10 μm shift on the RP was selected for small deformation situations to simulate static, low magnitude loading conditions. An MPC equation was set to account for the material continuity of the vertebral transverse process and ensure the transmission of the cohesive forces. Finally, a fixed boundary condition was set on the outermost boundary while symmetry boundary was administrated to nodes on the model's axis of symmetry (see **Figure 3b**).

2.4 Screw insertion techniques

The failure mechanisms are known to be most active at the bone-screw interface. Several scenarios can schematically predict the mechanisms of bone failure leading to implant pull out. To help elucidate the failure mechanism in bony structures, the Von-Mises and shear stresses developed due to small deformation and large screw excursions were thoroughly investigated.

The most common screw insertion techniques adopted by surgeons in spinal fixation surgeries were modelled in the current study to analyze their respective effects on the stability of pedicle screws when implanted in the vertebral body. **Figure 4** defines and graphically illustrates each of the selected models.

3. Results And Discussions

The ultimate pullout effort is defined as the load for which the pedicle screw leaves its anchorage in normal and osteoporotic bone. Unlike the loosening mechanism, tearing is often accompanied with unpredictable failure at the bone-implant interface. One of the original inputs implemented in the current study was the incorporation of the subcortical bone while computing the resistive force as a result to the prescribed screw excursion. To our best knowledge, none of the preceding research works had tackled this issue considering only the cortical and trabecular bones.

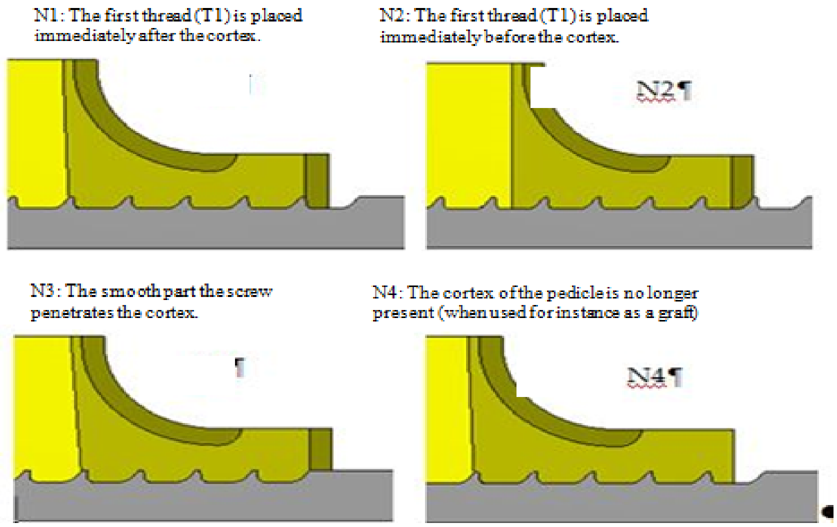

Figure 4 CAD models of four screw-insertion types considered in the present study.

In the upcoming analyses, the pull out effort was predicted for the different screw insertion techniques, bone quality types (normal or osteoporotic) and type of loading imposed on the pedicle screw (small and large deformation formulations).

Under small deformations formulation, a maximal resistive effort of about 155 N was computed for the N2 insertion case, as shown in **Figure 5**.

The reason for that is that the first screw thread is more deeply engaged in the vertebral subcortical bone, leading to a larger amount of bone being located between the first thread andthe cortical bone layer, which substantially enhanced the bone's resistance to the screw pulling out.

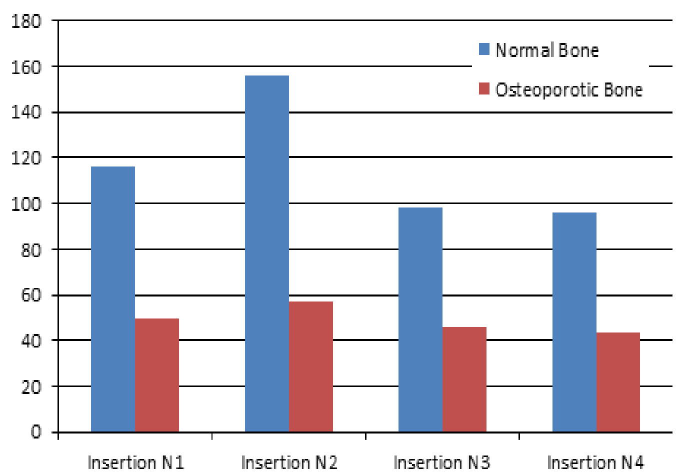

Figure 5 Maximal resistive effort in normal and osteoporotic bone models as a result to a 10 μm screw excursion for various screw insertion types.

The load displacement curves computed under large deformations for the four types of screw insertion techniques, for normal and osteoporotic bones, showed that N1 insertion technique provide d the highest pullout effort (~2300 N) in contrast to the N3 and N4 techniques, supporting each about 1870 N (see figures 8a and 8b). Similarly, the osteoporotic bone similar ultimate pull out strength were recorded for the N1 and N2 insertion techniques (~830N) while the N4 technique provided the least resistance (~650N).

As a general observation, the computed results the pull out effort was highly sensitive to bone density irrespective to the insertion technique with the most drastic loss in the pullout strength of about 65% occurring for the N4 insertion type, characterized by a missing cortical wall.

Again, irrespective to the bone density, the N1 technique come out to be the best screw insertion technique since it it provided the highest ultimate strength at the farthest screw displacement along the screw path, as clearly shown in figures 6a and 6b. These values were close to those established experimentally [12].

The load displacement curves for an osteoporotic bone generally presented dips that are more pronounced what have been depicted in normal bone. This may be due to the relatively effortless radial deformation of the bone material freeing, sequentially, the implant threads. Similar irregularities were reported in an in-vitro study [13] aimed to estimate the pull out strength of conical and cylindrical screws of 6.5 to 7 mm diameters implanted in cadaveric vertebral pedicles.

The authors claim that such irregularities were due not only to the shape of the threads at the bone-implant interface but also to the degree of osseointegration and bonding that have taken place at the interface.

4. Conclusion

The current study focuses on numerical evaluation of the strength of spinal implants inserted

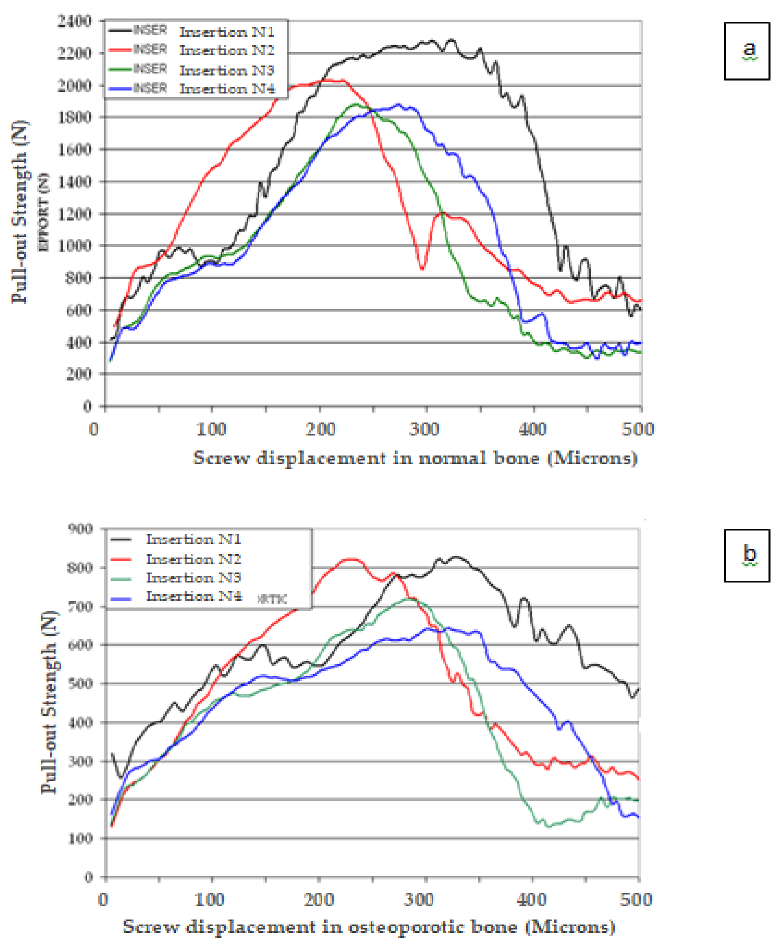

Figure 6 The load-displacement curves under large deformations formulation, plotted for the four types of screw insertion in (a) normal bone and (b) osteoporotic bone

in the pedicle vertebra. Since studies that combine both in-vivo and numerical simulation through patient-specific models remain very limited. The current numerical simulation intended to reproduce, as close as possible, some of the physiological solicitations encountered in normal spine during day-to-day activities. Through this study, the subcortical bone component is implemented in the analysis to better explore the multi scale aspect of the bone. The chosen elasto-plastic stress-strain relationships are characteristic of the vertebral bone material and surely help in the prediction of the pedicle screw tearing mechanisms.

The accuracy and toughness of the surgical procedure during implant placement can lead to more stable implants. Clinical parameters such as the screw insertion type and depth, the bone density and the patient degree of mobility, greatly affect the mechanisms of the implant's failure/success. As portrayed in this study, the preservation of the pedicle cortex as in the N1 insertion technique and the screw anchoring depth gearing a maximum number of threads promote a better protection against premature breakouts of pedicle screws.

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