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PUSHOUT STRENGTH OF TIBIAL LOCKING SCREWS: DEVELOPMENT OF FINITE ELEMENT MODELS

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ABSTRACT

This study investigated the bone holding power of tibial locking screws by mechanical testing and finite element analysis. In mechanical tests, five types of commercially available tibial locking screws: Howmedica, Osteo AG, Richards type I, Richards type II, and Synthes were inserted in a cylinder of polyurethane foam bone. Axial load was applied to the screw tip to push the screws out of the foam bone by a material testing machine. The pushout strength of each screw was recorded. In finite element analysis, three-dimensional finite element models with nonlinear contact interface between the screws and the bones were created to simulate the mechanical testing. The results showed that the order of the pushout strength of the locking screws from high to low was Osteo AG, Richards type II, Richards type I, Howmedica, and Synthes. The results of mechanical tests were highly correlated to the results of finite element analysis. Finite element models with low elastic modulus of bone and no frictional force between the screws and bones can better simulate the situations of mechanical testing. The finite element models built in this study may help the manufacturers evaluate new designs of locking screws before manufacture and assist surgeons to select suitable devices for their patients.

Key Words: finite element analysis, holding power, tibial screw.

I. INTRODUCTION

Interlocking nailing is widely used for treatment of tibial shaft fractures because of its advantages of minimal soft tissue injury and stable fracture fixation (Alho *et al.*, 1990; Brumback, 1996; Greitbauer *et al.*, 1998). However, loosening and back-out of the locking screws may result in loss of fracture fixation and impairment of fracture healing. This problem may be more serious in patients with osteoporosis. Bone holding power of different screws, such as cortical screws, cancellous screws, cannulated screws, transpedicle screws in spine or even tibial locking screws has been widely studied by mechanical testing with different kinds of bones (Asnis *et al.*, 1996; George *et al.*, 1991; Leggon *et al.*, 1993; Lin *et al.*, 2001). However, mechanical testing for newly developed screws requires pre-manufacture of the screws and facilities or materials for testing. A good analytical model to investigate the mechanical behaviors of the new devices can effectively save the expenditure and the effort involved in mechanical testing. The purpose of this study was to develop finite element models to simulate pushout tests of the tibial locking screws. Meanwhile, mechanical tests of the bone holding power were conducted on five types of commercially available tibial locking screws. The results of the mechanical tests were compared with those of the finite element analysis. The feasibility of the finite element models was studied by changing different loading conditions, material properties and the interface properties.

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II. MATERIALS AND METHODS

1. Structures of the Tested Screws

Five types of commercially available stainless steel tibial locking screws: Howmedica (Howmedica,

Table 1 The dimensions of the locking screws

Geometry of locking screws	Synthes	Howmedica	Richards type I	Richards type II	Osteo AG
Outer diameter (mm)	4.88	4.46	4.5	4.96	4.96
Inner diameter (mm)	4.32	3.80	3.71	3.96	3.52
Proximal root radius (mm)	0.3	0.4	0.1	0.23	0.2
Distal root radius (mm)	0.3	0.4	0.4	0.43	1
Pitch (mm)	2.75	1.48	1.26	1.29	1.74
Proximal half angle (°)	30	22.5	10	10	10
Distal half angle (°)	30	67.5	45	45	35
Thread depth (mm)	0.28	0.33	0.395	0.5	0.72
Thread width (mm)	0.5	0.2	0.3	0.15	0.1

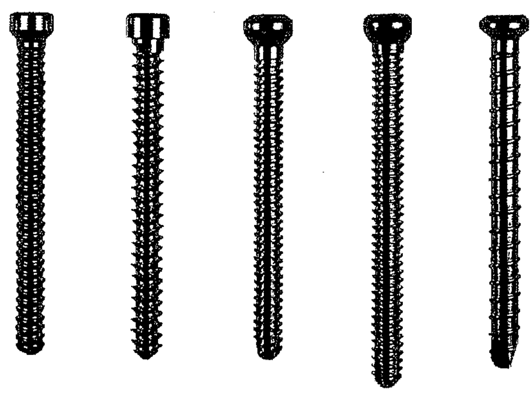

Fig. 1 Five types of commercially available stainless steel tibial locking screws (Howmedica, Osteo AG, Richards type I, Richards type II, Synthes)

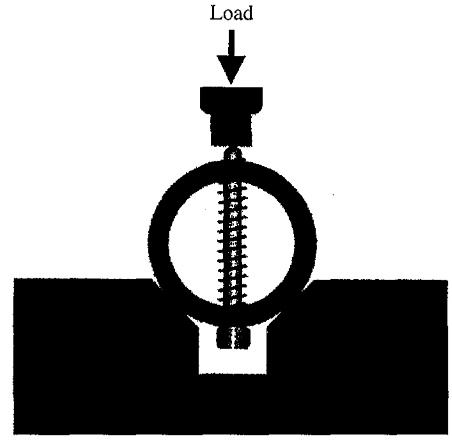

Fig. 2 Setup of pushout tests in this study

Rutherford, NJ), Osteo AG (Osteo, Selzach, Switzerland), Richards type I, Richards type II (Richards, Memphis, TN), and Synthes (Synthes, Paoli, PA) (Fig. 1) were tested in this study. The dimensions of these locking screws were measured with a Surface Roundness RA-100 (Mitutoyo, Kawasaki, Japan) including the outer diameter, inner diameter, root radius, pitch, half angle, thread depth, and thread width (Table 1). All of the locking screws were fully threaded with a self-tapping tip and the lengths were about 50 mm.

2. Mechanical Tests of Bone Holding Power

A polyurethane foam bone (Bayer, Leverkusen, Germany) with a density of 0.5 g/cm^3 was used for mechanical tests of locking screws. The foam bone was cylindrical in shape with an outer diameter of 40 mm and inner diameter of 30 mm. The compressive modulus was measured by applying a compressive load on the cross section of the cylindrical foam bone using a material testing machine (Bionix 858, MTS Corporation, Minneapolis, MN). Five types of locking screws were inserted through the center of the cylindrical foam bone with an interval of 30 mm between two neighboring screws to prevent interference among screws. The pilot hole was 3.5 mm in diameter and the screws were inserted without tapping until the screw cap compressed against the surface of the foam bone. Then the screws were axially loaded at the screw tip by the actuator of the materials testing machine (Fig. 2). Because of easier application of the load, the screws were pushed out from the foam bone instead of being pulled out. The loading mode was displacement control with at a loading rate of 2.5 mm per minute. The screw was pushed for a distance of 5 mm and the maximal load was recorded as the pushout strength.

3. Finite Element Analysis

Three-dimensional finite element models were established with commercial software ANSYS 5.7 (ANSYS, Inc., Canonsburg, PA). The computer used is a Intel Pentium IV-2.53 GHz with 1 Gb RAM (Acer, Taipei, Taiwan). At first, three-dimensional locking screw models were generated by helical sweep with ANSYS Parametric Design Language

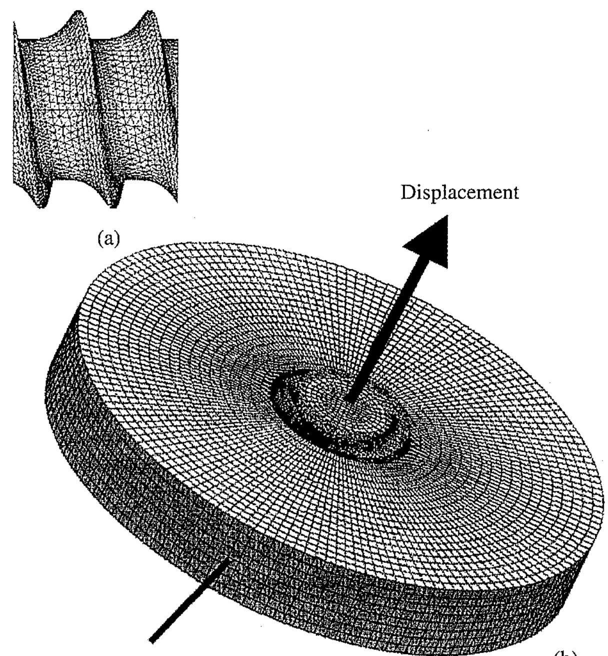

Fig. 3 (a) The configuration of the screws; (b) The boundary condition and loading condition of the finite element models

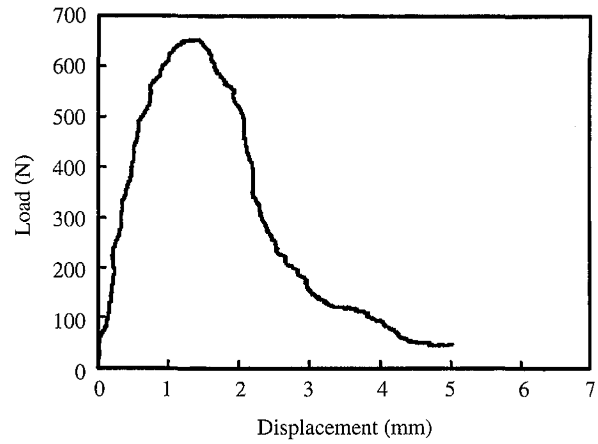

Fig. 4 The load deformation curve of the push-out experiment

(APDL) according to the measured dimensions and geometry. Then the surface models were transformed to solid models by Boolean operations (Fig. 3(a)). Locking screws surrounded by a layer of cortex in round shape with an outer diameter of 20 mm were used to simulate anchorage of the screws in the bones (Fig. 3(b)). The screws were 4 mm in length and the bones were 3 mm in length. The material type of the locking screws was stainless steel with a modulus of elasticity of 230 GPa. The cortical bone as an isotropic material was supposed to be osteoporotic with an elastic modulus of 0.36 GPa. The Poisson's ratio was 0.3 for both locking screws and bones. Locking screws nestled closely on the bone with a nonlinear contact interface without any frictional force. Locking screws and the bones were map-meshed using elements of SOLID 45 except for the outer layers, which were free-meshed using elements of SOLID 95. SOLID 45 was an eight-nodes brick element and SOLID 95 was a high order twenty-nodes brick element with compatible displacement shapes for curved boundary. The element size was 0.4 mm for map-mesh and 0.2 mm for free mesh. Surface-to-surface contact elements were used for the interface between the locking screws and bones. The contact surfaces of locking screws were meshed with CONTA 174 and the contact surfaces of the bone were meshed with TARGE 170. A surface displacement of 0.01 mm was applied to the cross-section surface of one end of the screws and only axial displacement of the screws was allowed. The peripheral margin of the bones was fully constrained (Fig. 3(b)). The solution was done by Precondition Conjugate Gradient (PCG) solver and the convergence tolerance was 10^{-4}. A line search option was on and the maximum number of equilibrium iterations was 30. Time step size, which determined the accuracy of the solution, was 0.01 in this study and the time at the end of step was 1. The maximum time step size was 0.5 and minimum time step size was 0.001. The post-processing analysis included calculation of the total strain energy of the bone (Marks *et al.*, 1993) and total reaction force of the screw. The total strain energy was defined as the summation of the strain energy of the elements of the bones and the total reaction force was defined as the summation of the reaction force of the nodes over the screw surface with pre-applied displacement. Because the results of the finite element analysis might be affected by different loading conditions, material coefficients and the interface property between the screws and the bones, the feasibility of the finite element models was studied based on different values of these factors. The results of mechanical testing and finite element analysis were compared.

III. RESULTS

1. Mechanical Tests of Bone Holding Power

The compressive modulus of the polyurethane foam bones used in this study was 0.36 GPa. In the pushout tests, all failures occurred in the foam bones. The screws were pushed out by shearing a cylinder of foam materials surrounding the threads and the structures of the screws were completely intact. As the screws were loaded, the load increased rapidly and dropped abruptly when the bone was stripped (Fig. 4). The displacement of the screws at the

Table 2 Pushout strength, total strain energy and total reaction force of the locking screws

Locking screws	Synthes	Howmedica	Richards type I	Richards type II	Osteo AG
Pushout strength (N)	494±47	503±65	585±56	644±31	673±63
Total strain energy (10^{-3} J)	19.52	19.96	20.88	22.01	21.62
Total reaction force (N)	3.92	4.00	4.19	4.41	4.34

Table 3 Total strain energy and total reaction force of the locking screws in situations with different pre-applied displacements

Displacement (mm)	Synthes	Howmedica	Richards type I	Richards type II	Osteo AG	Correlation coefficient
Total strain energy (10^{-3} J)						
0.01	19.5	20.0	20.9	22.0	21.6	0.96
0.1	1933	1990	2082	2194	2156	0.96
0.3	17149	17799	18653	19661	19311	0.95
0.5	47010	49173	51615	54432	53447	0.95
Total reaction force (N)						
0.01	3.91	4.00	4.18	4.41	4.34	0.96
0.1	38.8	39.9	41.8	44.0	43.3	0.96
0.3	115	119	125	132	129	0.96
0.5	189	198	207	219	215	0.95

ultimate load was less than 2 mm. The order of the pushout strength of the locking screws from high to low were Osteo AG, Richards type II, Richards type I, Howmedica, and Synthes (Table 2).

2. Finite Element Analysis

Total element number of the models in this study ranged from 70,000 to 190,000. Total node number ranged from 80,000 to 240,000 and the computer solution time ranged from 5 to 35 hours. The bone surrounding the screw threads had higher element strain energy than elsewhere. The reaction force was evenly distributed on the cross-section surface of the screws with pre-applied displacement. Initially when the total strain energy was analyzed, it was found that there were a number of elements with extraordinarily high strain energy sporadically scattered within the regions of thread roots. This resulted in abnormally high total strain energy and total reaction force. These elements had irregular contours and were considered to be caused by inappropriate element size. Consequently, convergence study was conducted by increasing the mesh density in the regions of thread roots. This local element size was decreased from 0.2 mm to 0.03 mm and the solutions were considered converged when the differences between two neighboring data were less than 2%. With increasing meshing density, the abnormal elements were eliminated and the total strain energy and the total reaction force converged. Only the models with convergence of total strain energy or total reaction force were used for subsequent analysis. The study showed the total strain energy of the bones was closely related to total reaction force of the screws with a very high correlation coefficient of 0.99. The results of the finite element analysis were also very closely related to the push-out strength of the screws in the mechanical tests with the correlation coefficient of 0.96 for both total strain energy and total reaction force (Table 3). In the feasibility study of the finite element models, increasing the pre-applied displacement of the screws from 0.01 mm to 0.50 mm might increase the absolute value of total strain energy and total reaction force but it had minimal effects on the relationship between them and the pushout strength obtained in the mechanical tests. The change of the correlation coefficients was from 0.96 to 0.95 and 0.96 to 0.95 respectively. Increasing the elastic modulus of bones from 0.36 GPa to 15 GPa might also increase the total strain energy and the total reaction force but the correlation coefficient dropped from 0.96 to 0.66 and 0.96 to 0.88 respectively. Increasing the frictional coefficient of the interface between the bone and the screw also increased the total strain energy and total reaction force but the correlation coefficient also dropped from 0.96 to 0.81 and 0.96 to 0.88 respectively.

IV. DISCUSSION

With optimal modeling, the present study demonstrated an excellent relationship between the results of finite element analysis and mechanical tests. Bone holding power of the screws has been examined by pullout tests using cadaver bones (Evans *et al.*, 1990; George *et al.*, 1991; Kwok *et al.*, 1996; Okuyama *et al.*, 1993; Zdeblick *et al.*, 1993), animal bones (Halsey *et al.*, 1992), and synthetic bones (Brown *et al.*, 2000; Chapman *et al.*, 1996; DeCoster *et al.*, 1990; Evans *et al.*, 1990; Lin *et al.*, 2001; Liu *et al.*, 1995). The synthetic bones have the advantages of less individual variation, more consistent test results, higher availability and easier specimen handling. The synthetic bones were used to simulate the situations in real bones, but the relative values among different screws were more useful than absolute values (Brown *et al.*, 2000; Thompson *et al.*, 1997). The real pushout strength in vivo is further affected by density, the geometries, the material properties of the bone and the size of the pilot holes. In a previous study, the density of the testing bone and the size of the pilot holes did not affect the relative pushout strength of the screws.

In this study, polyurethane foam bone with a density of 0.5 g/cm^3 represented the worst scenario of the clinical condition. Most patients with loosening or back-out of screws are associated with remarkably osteoporotic bones which cannot hold the screws tightly. The compressive modulus of this polyurethane foam bone was 0.36 GPa, which is much lower than that of normal cortical bone. In the push-out tests, the failure was cylinder shear of the bone surrounding the screw threads and the screw structures were completely intact. This was compatible with the clinical findings in patients with severe osteoporosis. This was also the reason why the elastic modulus of the bone similar to that of the foam bone was used in the following finite element studies.

The finite element method has become a widely used tool in orthopedic biomechanics. It is a computer-simulation method suitable for determining the stresses and strains at any given point of a structure of arbitrary geometrical and material complexity. The model relies on accurate constitutive modeling of the structure and the material coefficients, loading characteristics and boundary and interface conditions. Considering inventions and new products, finite element studies of the mechanical behaviors of a new design can save the expenditure, time and effort of manufacturing and mechanical testing (Brown *et al.*, 2000). Hence, finite element models were used in this study to examine the bone holding power of tibial locking screws and the results of finite element analysis were compared with those of mechanical testing. The finite element models with only one cortex could represent situations with two cortices with completely the same structures and material properties. The round shape of the bone could ensure the identical boundary condition over the entire circumference of the screws, otherwise the analytical results might be affected by the position of the screws in the bone.

In the finite element models created in this study, the meshing of the bone at the region of thread roots showed uneven element contours which led to abnormally high element strain energy when the element size was not adequate. This meshing problem was more severe in screws with a small root radius, like Richards type I screws. However, increasing the meshing density in these regions could eliminate these abnormal elements (Keyak *et al.*, 1992) and the total strain energy as well as the total reaction force could converge.

This study showed that the analytical results of the finite element models of the tibial locking screws were closely related to the results of mechanical tests with the correlation coefficient of 0.96 for both total strain energy and total reaction force, but it was found that the three-dimensional nonlinear finite element models were additionally affected by the loading condition, the material property of the bone and the frictional coefficient between the screw and the bone. Consequently, the feasibility of the finite element models was studied. First, different displacements were applied to the screws. As the pre-applied displacement increased from 0.01 mm to 0.50 mm, the total strain energy and total reaction force of the screws increased, but the relationship between the finite element results and the testing results was almost the same for both total strain energy and total reaction force (Table 3). This indicated the pre-applied displacement did not affect the predictability of the finite element models, but the pre-applied displacement could not be larger than 0.50 mm, because the contact elements might separate and the solution would diverge. To conserve computational time, pre-applied displacement of 0.01 mm was used in the subsequent studies.

Secondly, the effects of the elastic modulus of the bone were examined. It was found that both total strain energy and total reaction force increased as the elastic modulus of the bone was increased from 0.36 GPa to 15 GPa. However, the correlation coefficient between the finite element results and the testing results substantially decreased for both total strain energy and total reaction force (Table 4). The reason accounting for this decrease of relationship was increased deformation of the screws in situations with

Table 4 Total strain energy and total reaction force of the locking screws in situations with different elastic moduli

Elastic modulus (Gpa)	Synthes	Howmedica	Richards type I	Richards type II	Osteo AG	Correlation coefficient
Total strain energy (10^{-3} J)						
0.36	19.5	20.0	20.9	22.0	21.6	0.96
5	267	273	304	300	289	0.74
10	525	537	598	587	558	0.65
15	776	793	852	864	812	0.66
Total reaction force (N)						
0.36	3.91	4.00	4.18	4.41	4.34	0.96
5	53.9	55.2	57.2	60.7	59.1	0.94
10	107	109	113	120	116	0.91
15	159	163	169	178	171	0.88

Table 5 Total strain energy and total reaction force of the locking screws in situations with different frictional coefficients between the screws and bones

Frictional coefficient	Synthes	Howmedica	Richards type I	Richards type II	Osteo AG	Correlation coefficient
Total strain energy (10^{-3} J)						
0	19.47	19.97	20.92	22.01	21.63	0.96
0.13	19.66	20.06	22.35	22.09	21.71	0.83
0.27	20.11	20.25	22.50	22.23	21.89	0.83
0.4	20.79	20.48	22.70	22.42	22.11	0.81
Total reaction force (N)						
0	3.91	4.00	4.18	4.41	4.34	0.96
0.13	4.05	4.07	4.21	4.48	4.40	0.96
0.27	4.18	4.13	4.26	4.53	4.46	0.94
0.4	4.30	4.18	4.30	4.57	4.51	0.88

high elastic modulus of the bone. These findings supported using bones with low elastic modulus in finite element analysis.

Thirdly, the frictional coefficients between the bone and the screws were increased from 0 to 0.4, the frictional coefficient between stainless steel and normal cortical bone. It was observed that the correlation coefficient between the finite element results and the testing results also dropped for both total strain energy and total reaction force (Table 5). The reason for this finding was similar to that in the situations with increasing elastic modulus of the bone. This implied that the finite element models couldn't be right unless the frictional coefficient between the bone and metal was low or neglected. This was reasonable because the friction force between osteoporotic bone and metal should be much lower than that between normal cortical bones and metal.

The present study had drawbacks. The effects of the size of the pilot holes were not considered in the finite element models. However, this effect was not important in screws with the same inner diameter or in situations with osteoporotic bones. The absolute values of the reaction force were much lower than those in the mechanical tests. However, practically, relative values are more critical than absolute values in both mathematical models and mechanical testing. The computer time used for this finite element study was still too long because of the complexity of the structures. Further studies to decrease the computer time by modification of the models are warranted.

V. CONCLUSIONS

The finite element models developed in this study could reliably predict the pullout strength of locking screws in osteoporotic bones. The finite element models may help manufacturers to evaluate new designs for locking screws before manufacture and assist surgeons to select suitable devices for their patients.

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