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Pre-clinical testing of hip prosthetic designs: a comparison of finite element calculations and laboratory tests

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To investigate the accuracy of finite element (FE) models for pre-clinical testing of unbonded hip prostheses, relative to aspects of load transfer and micromobility, two previously published laboratory experiments were simulated, using three-dimensional FE models. It was found for the load-transfer analyses that the experiment and the FE study revealed results that were very similar. The trends in the mobility experiments were also reproduced in the FE simulations, although quantitative differences were found. It is concluded that FE analysis can effectively be used for design evaluation of hip prostheses before prototypes are made.

1 INTRODUCTION

Total hip replacement is one of the most successful operations world-wide. However, the number of reoperations and revisions increases each year. Usually, the reason for a revision is aseptic loosening of the prosthesis. Prosthetic design greatly influences average survival rates of total hip replacements (1). Hence, it is of importance to test new prosthetic designs before they are implanted routinely. This can be done in computer simulation models, laboratory bench tests, animal experiments or clinical trials. Of course, not all these test methods address the same aspects. Computer simulation models and laboratory bench tests particularly address the adequacy of a design relative to mechanical factors, such as strength, load transfer, micromobility and adaptive bone remodelling. The accuracy of finite element (FE) models of hip arthroplasties has been investigated by a number of authors (2–5). The general approach was to measure strains at the bone or prosthetic surfaces and compare those to the findings of FE analyses. All investigators found good correlations in a relative sense, that is to say, in the trends of the stress patterns. The purpose of this study was to test the accuracy of finite element models for pre-clinical testing, relative to aspects of load transfer and micromobility of unbonded prosthetic hip stems. For this purpose, laboratory experiments with one particular stem design (6, 7) were simulated with the three-dimensional FE method, and results were compared.

2 MATERIALS AND METHODS

2.1 The load-transfer experiment

Carlsson *et al.* (6) investigated the influence of femoral neck resection on the stress patterns at the proximal/medial interface when a press-fitted stem was used. The stress applied to the medial cortex was recorded by

placing a pressure-sensitive film between the medial edge of the prosthesis and the inner aspect of the femoral neck. Six cadaver specimens were used.

The prosthesis was loaded by means of a lever arm, which was connected to the greater trochanter, the testing machine and the prosthetic head. In this way the loading mode of the one-legged stance in gait was simulated with a 1000 N force on the greater trochanter and 1500 N on the prosthetic head.

2.2 The rotation stability experiment

Using three different types of the same prosthesis, Tanner *et al.* (7) studied the rotational relative motions between prosthesis and bone due to cyclic anterior/posterior directed forces of 400 and 800 N on the femoral head, before and after femoral neck resection. The three prosthetic types investigated were of the same shape, but either cemented, smoothly surfaced press-fitted or laterally ridged press-fitted. Three femurs were tested for each type of prosthesis.

2.3 The finite element simulations

The simulations were carried out with a three-dimensional FE model symmetrical relative to the mid-frontal plane. The model, consisting of 1052 eight-node isoparametric brick elements with 1600 nodal points, is shown in Fig. 1. Four different materials were assumed in the model with each having its own mechanical properties. Young's moduli were 17 GPa for cortical bone, 1 GPa for cancellous bone, 200 GPa for the prosthetic material and 2 GPa for the acrylic cement. All materials were assumed to behave in a linear elastic, homogeneous and isotropic manner. Poisson's ratio was 0.3 in all materials. As shown in Fig. 2, resection of the femoral neck was modelled in such a way that the medial/lateral distance between the proximal endosteal bone and the centre of the prosthetic head was about

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Fig. 1 The three-dimensional finite element model

3.5 times smaller in the retained neck case than in the resected neck case. All prosthetic configurations were of the same shape and dimensions. For the cemented stem, rigidly bonded interfaces were assumed. The press-fitted, non-cemented stem interfaces were assumed unbonded with no friction (gap elements, MARC Analysis Research Corporation, Palo Alto, California). The ridges in the third type were simulated with ties of the lateral/proximal stem nodes to the bone, in the anterior/posterior and medial/lateral directions only.

Fig. 2 A proximal/medial part of the bone was removed from the model to study the effect of femoral neck resection

The cemented prosthesis was completely surrounded by a cement layer. The distal part of the press-fitted prosthesis had no initial contact with the endosteal bone.

The loading conditions in the two simulations are clarified in Fig. 3. For the load-transfer simulation a hip joint force (F_h) of 1500 N was applied to the prosthetic head with an angle of 13° relative to the femoral axis. The greater trochanter was loaded with 1000 N (F_m), making an angle of 20° with the femoral axis.

For the rotation stability analysis a force (F) of 400 and 800 N respectively was applied to the prosthetic

Fig. 3 Schematic representation of the loading conditions and the regions of interest

head in the anterior/posterior direction. In order to prevent rigid-body motions in the rotation stability analysis, a compressive force of 100 N in the axial direction of the stem was applied.

3 RESULTS

3.1 The load-transfer analysis

The load-transfer experiment (6) was in fact semi-quantitative in the sense that the interface compression patterns were graphically represented by colour intensities. In the experiment the inner surface of the medial neck cortex was not perfectly flat and was made up of reamed trabeculae, which resulted in spotty staining patterns. The surfaces in the FE simulation were flat and more continuous stress patterns were generated as shown in Fig. 4. In spite of this difference, the stress patterns in experiment and simulation were very similar. In the experiment, Carlsson *et al.* (6) used films with a sensitivity of 7–25 MPa. They found stained areas of about 1–15 per cent when the neck was retained and about 5–41 per cent in the resected neck case (Table 1). In the computer simulations, the percentages of areas with stress values higher than 7 MPa were also calculated, by counting the percentage of nodal points with stress values higher than 7 MPa. For the retained neck case 2 out of the 24 nodal points (8 per cent) had stress values higher than 7 MPa, while in the resected neck case 3 out of 15 nodal points (20 per cent) showed values higher than 7 MPa. These values correspond rather well with the findings of Carlsson *et al.* (6).

Significant compressive stresses were not only generated at the medial interface but also at the lateral interface, as shown in Fig. 5. These stresses were calculated

Fig. 4 Compressive interface stresses occurring in the load-transfer analysis for the retained neck case (left) and resected neck case (right)

Table 1 Compression forces on the medial cortex of the femoral neck measured by pressure-sensitive film. Figures are total stained area (mm²) on the film and the percentage of the total area studied. [Reproduced from Carlsson *et al.* (6)]

Specimen	Neck retained		Neck resected	
	mm ²	%	mm ²	%
1	4	1	15	5
2	60	9	95	21
3	33	4	120	31
4	120	15	170	41
5	32	5	64	21
6	85	13	105	39

Fig. 5 Compressive interface stresses occurring at the medial and lateral interface. Stresses were averaged in the anterior/posterior direction. A hip joint force of 1500 N was applied to the prosthetic head with an angle of 13°. The greater trochanter was loaded with 1000 N, making an angle of 20° with the femoral axis

Table 2 Mean displacements (mm). [Reproduced from Tanner *et al.* (7)]

		Cemented			Uncemented					
					Smooth			Ridged		
		1	2	3	1	2	3	1	2	3
Neck	400 N	0.100	0.085	0.070	0.700	9.940	0.35	0.242	0.373	0.177
Intact	800 N	0.220	0.173	0.170	—	—	—	0.992	0.723	0.42
Neck	400 N	0.240	0.080	0.240	*	†	0.83	1.090	‡	0.485
Resected	800 N	0.640	0.242	0.560	—	—	—	§	—	¶

* Fractured at 200 N.

† Fractured at 100 N.

‡ Transducer mounting loosened during neck resection.

§ Fractured at 500 N.

¶ Fractured at 540 N.

by considering not only the nodal points of the interface located in the mid-frontal plane but also the neighbouring anterior and posterior points. Hence, they are average stresses occurring at the medial and lateral interfaces. The compressive stresses increase due to neck resection except in the proximal/lateral region. This phenomenon was already suspected by Carlsson *et al.* (6), but was not actually measured.

3.2 The rotation stability analysis

Results of the rotation experiment by Tanner *et al.* (7) showed that cemented prostheses were more stable than non-cemented ones. Ridged non-cemented prostheses showed better rotational stability than smooth non-cemented prostheses. Resection of the neck affected the rotational stability in a negative sense. The smooth press-fitted prosthesis was most susceptible for neck resection (often bone fracture occurred), while the cemented types were the least affected by femoral neck resection. These experimental results are summarized in Table 2.

Although the differences were smaller, the ranking order of rotation stability of the three types of prosthesis was the same in the FE analyses as in the experiments. The rotational motions in the three cases, expressed by the posterior displacements of the prosthetic head relative to the bone, are shown in Fig. 6. The smallest displacements were found with a cemented prosthesis, and were entirely due to elastic deformation, hence linearly related to the applied force magnitudes. The smooth press-fitted type underwent the highest rotational displacements. The lateral ridges caused a reduction of the motions.

Resection of the neck resulted in higher posterior displacements in all configurations, especially in the smooth stem case, where the displacement increased from 0.674 to 0.842 mm (25 per cent) at a load of 800 N. The presence of longitudinal ridges on the proximal/lateral aspect of the prosthesis reduced the influence of resection of the neck. In this case the increase of the displacement was about 5 per cent. The cemented configuration was almost insensitive to femoral neck resection. Obviously, prosthetic type and femoral neck resection not only affect the relative displacements but

Fig. 6 Displacements of the prosthetic head relative to the bone for the three types of prostheses, obtained by FE calculations for anterior/posterior loads of 400 and 800 N respectively

Fig. 7 Von Mises stress patterns in a proximal transverse section, just below the resection level due to an anterior/posterior load of 800 N. High grey level indicates high stresses; low grey level indicates low stresses

also the stress distribution. As an example, Fig. 7 illustrates the von Mises stress distribution around the prosthesis.

4 DISCUSSION

In the load-transfer analysis, the experiments and the FE study showed corresponding results. The stress range was the same and the effect of femoral neck resection was very similar. Due to resection of the neck, the percentage of interface area with compressive stresses higher than 7 MPa increased from about 10 per cent to about 25 per cent. This was found in the experiments as well as in the FE simulations.

The actual displacements and the effect on the displacements of femoral neck resection, as found in the rotation stability simulations, were smaller than those of the experiments. However, the ranking order of the micromobility of the various prosthetic types matches that of the experiments. The FE study also showed larger relative displacements in the resected neck case for all prosthetic types, as was found in the experiment.

Differences in an absolute, quantitative sense are obvious and unavoidable, due to the schematic character of FE models. For instance, the FE model assumed perfect prosthetic fit, which is obviously never realized, either experimentally or clinically.

The interface conditions used in the simulations are approximations of reality. In the cemented case a perfect bond was assumed, while in the press-fitted case an unbonded interface with no friction was modelled. For the ridged press-fitted prosthesis, ridges were not actually modelled but the interface nodal points were tied in the anterior/posterior and the medial/lateral directions. This method will cause errors in the vicinity of the ridges. However, the rotation stability is focused on global displacements and therefore the model is

expected to produce acceptable results in a qualitative sense.

Other causes for deviations were the mechanical properties of the various materials assumed. In this investigation, they were assumed to be linear elastic, homogeneous and isotropic, whereas in reality bone is inhomogeneous and anisotropic. An elastic modulus of 17 GPa was used for cortical bone. This is appropriate for the longitudinal direction. However, in the transverse and radial directions the stiffnesses are about half this value (8). Undoubtedly, these assumptions will cause deviations when FE results are compared with laboratory experiments in a quantitative sense. In addition, the experiments are subject to variations in test conditions. In the experiments given in references (6) and (7), the results vary from specimen to specimen, due to technical imprecision and variability.

FE simulations as a pre-clinical test procedure have the capacity to perform several tests with the same model (as shown here) and produce any relevant stress, strain or relative motion variable desired. In most cases it is practically impossible to assess these variables experimentally. The present study indicates that computer simulation studies with FE models can produce results that are similar to those of laboratory tests. Even more importantly, the conclusions drawn from the experiments are the same as those drawn from the FE analyses, meaning that this information could have been available even before prototypes of the designs were made.

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