Influence of femoral head size on impingement, dislocation and stress distribution in total hip replacement

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Abstract

Dislocation remains a serious complication of total hip replacement. An insufficient range of motion can lead to impingement of the prosthetic neck on the acetabular cup. Together with the initiation of subluxation and dislocation, recurrent impingement can cause material failure in the liner. The objective of this study was to generate a validated finite element (FE) model capable of predicting the dislocation stability of different femoral head sizes with regard to impingement in different implant positions as well as the corresponding stress distribution in the liner. In order to cover posterior and anterior dislocation, two total hip dislocation associated manoeuvres were simulated using a three-dimensional nonlinear finite element model. The dislocation stability of two head sizes was determined numerically and experimentally. After validation, the FE model was used to analyse the dislocation stability of four different head sizes in variable implant positions. Range of motion (ROM) until impingement, the resisting moment that was developed and ROM until dislocation were evaluated. Additionally, stress distribution within the polyethylene liner during impingement and subluxation was determined. For both dislocation modes, a cup position of 45° lateral abduction and 15° up to 30° anteversion resulted in appropriate ROM and dislocation stability. In general, larger head diameters revealed an increase in ROM and higher resisting moments. Stress analysis showed decreased contact pressures at the egress site of the liners with the larger inner diameters during subluxation. The analysis shows that an optimal implant position and a larger head diameter can reduce the risk of dislocation induced by impingement. The finite element model that was developed enables simplification of design variations compared to experimental studies since prototyping and assembling are replaced by prompt numerical simulation.

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1. Introduction

Dislocation of the artificial joint is a serious complication of total hip replacement (THR). The probability of dislocation ranges from 2 to 5% in primary THR [1] and up to 16–20% in case of revision [2] and tumour surgery [3]. Dislocation ranks second as a cause for revision of THR, following aseptic loosening [4]. The most common risk factors include insufficient pseudo-capsular tissue, muscle weakness and malpositioning of the implant components [5–7]. The latter causes about 30–40% of all dislocations [8]. Approximately one third of these patients suffer recurrent dislocations and consequently require revision surgery. The annual costs caused by dislocation are estimated to be at least $75 million in the United States alone [7]. The desired range of motion of THR is set on implantation and depends on position and design. Excessive joint motion can lead to impingement of the femoral neck on the acetabular cup. If impingement occurs, the centre of rotation moves from the head centre to the rim of the cup. Further motion leads to subluxation of the femoral head and lever-out. Moreover, recurrent impingement can cause material failure of implant components, such as excessive wear of polyethylene liners as well as brittle fracture of ceramic
components. These failures occur as a result of high localised contact stresses at the impingement site. Contact stresses also increase during subluxation due to the decrease of contact areas between the femoral head and the liner. High shear stresses at the cup/bone interface are induced by impingement of the neck on the liner and could possibly lead to cup loosening. Besides impingement-related failure, malposition of THR and unfavourable manoeuvres entail the prosthetic head to slip out of the cup without leverage of the neck [6,9].

To evaluate the dislocation stability of different implant designs, multiple studies have been conducted. Scifert et al. [7] found that the development of the resisting moment yields sound information about the dislocation behaviour. They analysed a large series of implant designs with regard to posterior dislocation. A recent study of Steward et al. [10] deals with capsule implementation into a hip dislocation finite element model. Bader et al. [5,6] developed an experimental testing device to analyse the range of motion of THR, the resisting moments during subluxation and the stability against dislocation. Due to the necessity of prototyping and assembling, the experimental testing of a large series of implant designs and positions is a rather laborious way compared to finite element modelling. Implementing capsule representation into the finite element model is undoubtedly an interesting approach, but with regard to impingement-related failure, the geometry of the implant is the crucial factor.

The objective of the present study was to generate a simplified three-dimensional finite element (FE) model capable of predicting dislocation stability of various designs of the prosthetic head, neck and liner. We were also interested in the effects different implant designs and implant positions have on the stress distribution during impingement and subluxation. In contrast to the studies of Scifert et al., we defined the prosthetic head and neck as accurate rigid analytical surfaces instead of approximate modelling with finite elements. Nadzadi et al. [9] emphasised the importance of anterior dislocation, which is why we chose to simulate two dislocation-prone manoeuvres, including anterior and posterior dislocation. Different head sizes were analysed with regard to the stress distributions produced in the liners and their effect on the development of resisting moments due to stem rotation.

2. Material and methods

Four different ultra-high-molecular-weight (UHMW) polyethylene liners in combination with cobalt–chromium heads with head diameters of 28, 32, 36 and 40 mm were analysed. The wall thickness of each liner was 7 mm, head inset of the acetabular component was 2 mm. The prosthetic neck diameter was 14 mm and the lubrication gap between prosthetic head and liner was 24 μm, based on our own measurements. Since this study was intended to investigate the influence of liner design only, bone was not included into the FE-model.

2.1. Load cases

Two total hip dislocation related manoeuvres described by Kummer et al. [11] were analysed. The first mode inducing anterior dislocation was external leg rotation in combination with 10° extension and 15° adduction. The second mode inducing posterior dislocation was internal leg rotation with 90° flexion and 0° abduction. Fig. 1 shows the corresponding position of the implants and the direction of rotation.

2.2. Finite element modelling

The computational calculations were performed using ABAQUS V 6.4 (ABAQUS Inc., Providence, RI) in combination with Patran 2004 (MSC Software, Santa Ana, CA). Each finite element analysis was geometrically and physically nonlinear due to finite sliding contact, large deformations and

![Fig. 1. Illustration of the two dislocation modes: mode 1; external leg rotation in combination with 10° extension and 15° adduction; mode 2; internal leg rotation in 90° flexion and 0° abduction. Sites of posterior and anterior impingement are marked.](image-url)
nonlinear material definition. The liners were discretised geometrically using approximately 10,800 hexahedral 8-node elements as well as 6-node wedge-elements at the axis of rotation, each with linear interpolation (Fig. 2). Convergence of the FE model was assessed to justify the number of elements. A special feature in this study was to model the femoral head and neck as a rigid analytical surface instead of approximate modelling using finite elements [7]. Considering the huge difference of the structural stiffness between the metal head and the polyethylene liner, the deformation of head and neck could be neglected. The acetabular cup which is proximally fixed in the ilium and supports the liner was assumed to be rigid compared to the relatively small stiffness of the polyethylene. For reasons of computational economy, the acetabular cup was replaced by blocking the degrees of freedom of the proximally located nodes of the liner. The leg rotation was applied by a reference node in the head centre. Consequently, modelling the stem became redundant.

2.3. Material model

An elastic-plastic material model with isotropic hardening was used to approach the visco-elastic-plastic behaviour of the ultra-high-molecular-weight polyethylene (UHMW-PE). The equation developed by Fregley et al. [12], supplemented by the most recent values of Young’s modulus (945 MPa), Poisson’s ratio (0.45) and Yield strength (23.56 MPa) [13], was used to describe the material model. To satisfy the ABAQUS input requirements the equation developed by Fregley et al. was converted into a function of true stress and logarithmic plastic strain and then discretised into eight segments. Between the articulating surfaces, Coulomb friction was assumed, with a friction coefficient of $\mu = 0.01$, as discovered by Saikko [14].

2.4. Boundary conditions

The hip joint loads as determined by Bergmann [15] using an instrumented hip endoprosthesis were applied to the models being tested in the present study. Their published data was analytically converted to the two modes being analysed in the present study. The resulting hip joint force in 90° flexion was $F_x = 506$ N, consisting of the components $F_x = -15$ N, $F_y = 270$ N, $F_z = 427.5$ N. The resulting hip joint force in 10° extension and 15° adduction was $F_x = 1288$ N, consisting of the components $F_x = 40$ N, $F_y = 1244$ N, $F_z = 340$ N. The Cartesian coordinate system used in this study originates in the head centre of the right leg, with the $x$-axis pointing medially, the $y$-axis pointing cranially and the $z$-axis pointing anteriorly.

In the first calculation step, the hip joint force was applied to the reference node in the head centre, while the rotational degrees of freedom of the proximally located nodes of the liner. The leg rotation was applied by a reference node in the head centre. Consequently, modelling the stem became redundant.

2.5. Variations

Both load cases were analysed with four different head sizes (28, 32, 36, 40 mm). Variations in implant position were performed considering all combinations of lateral cup abduction (inclination) (30°, 45°, 60°) and cup anteversion (0°, 15°, 30°). The position of the prosthetic neck was modelled with defined stem anteversion of 0°. At both dislocation modes the resisting moment emerging from the applied rotation was calculated. ROM until impingement, the maximum resisting moment, ROM until dislocation as well as the stress distribution in the liner were evaluated.

2.6. Mechanical testing device

An experimental test setup previously developed to evaluate factors for dislocation stability [5,6] was used to vali-
date the resisting moments from the FE-model (Fig. 3). The acetabular component was placed in a retainer in a definite position and fixed with epoxy resin. The retainer was mounted movable in three perpendicular translational degrees of freedom. Translation of the retainer in the vertical and in the horizontal plane was realised by linear guides. The retainer, attached to the linear guides, was mounted in a universal testing machine. The vertical component of the hip joint load ($F_y$) was applied by the actuator of the testing machine. The horizontal components of the hip joint load ($F_x$, $F_z$) were applied by weights, wires and guide rollers. Rotational degrees of freedom were blocked in order to keep the appropriate cup position. The femoral component was embedded in the measuring fixture with epoxy resin. The measuring fixture included a torque transducer and a rotary transducer and was mounted on the base plate of the testing machine. A pivot bearing was used to allow stem rotation. The stem was rotated by an electromechanical drive with an angular velocity of $2^{\circ}$ s$^{-1}$, while the torque transducer and the rotary transducer recorded the resisting moment and the corresponding rotational angle respectively. Detailed information about the test setup has been reported previously [5,6].

2.7. Validation

Two reference THR systems with a cobalt–chromium head (28 and 32 mm diameter) were used for validation. Both THR systems were tested using a metal taper of 14 mm diameter and a matching ultra-high-molecular-weight polyethylene (UHMW-PE) liner with 2 mm head inset. The cup positions for validation were $45^{\circ}$ lateral abduction in combination with $30^{\circ}$ anteversion and $60^{\circ}$ lateral abduction in combination with $30^{\circ}$ anteversion. Stem anteversion was $0^{\circ}$. The implant components were placed in the testing device in the appropriate positions. The measuring fixture holding the stem was placed in $90^{\circ}$ flexion, and internal rotation was applied, according to the second load case from the FE model. Rotation of the stem was performed until total dislocation occurred, and the resisting moment and the corresponding rotational angle were recorded. Each experiment was repeated two times ($n = 3$).

The same reference THR systems were modelled as described above using the finite element method. For validation, the same implant positions were analysed and the progression of the resisting moment was calculated. The results of the experiments were compared to the analytical data.

3. Results

3.1. Validation

The development of the resisting moment, emerging as a reaction of the tangential stress and friction in the joint due to leg rotation and hip joint force, consisted of three phases. Initially, rotation was only impeded by the friction of the articulating surfaces. As soon as impingement between the neck and the liner occurred, the resisting moment increased abruptly upward due to shifting of the head rotation centre to the rim of the liner. With further rotational movement, the resisting moment progressively decreased due to subluxation associated with a decrease of the articulating contact area. Fig. 4 shows the resisting moments developed by the reference model with 28 mm head size under load case 2 with the acetabular cup at $60^{\circ}$ lateral abduction and $30^{\circ}$ anteversion. The experimental and numerical (FEM) data are in good agreement, including the three characteristic phases of the moment progression. The sole exception in congruency is the second peak in the experimental data set. This peak is due to impingement of the test setup and thus is clearly a testing artefact which may be ignored. The comparison of the experimental and numerical data of both reference THA systems with 28 and 32 mm head size in all implant positions used for validation is shown in Fig. 5. The mean error in impingement-free ROM was 12.0%, the mean error in maximum resisting moment was 6.5%.

![Fig. 4. Course of the resisting moment produced by internal rotation in 90° flexion and 0° abduction (mode 2). Liner position is 60° abduction and 15° anteversion. Comparison of three experimental trials and the results of one finite element analysis. The experimental data are not smoothed nor filtered.](image-url)
3.2. Implant position

With regard to implant position, the analysis revealed a counteracting effect: there was early impingement in extension and adduction with a highly anteverted and abducted cup. In contrary, 90° flexion caused early impingement in a less anteverted and less abducted cup position. A compromise for a maximum impingement-free ROM for both modes was a cup position of 45° abduction and 15° up to 30° anteversion (Fig. 6). The analysis showed an increase in resisting moments with more anteverted cups. In 45° lateral abduction at 90° flexion, the maximum resisting moment rose from 2.595 Nm (0° anteversion) to 4.082 Nm (30° anteversion). In 60° lateral abduction at 90° flexion, the maximum resisting moment rose from 1.591 Nm (0° anteversion) to 3.137 Nm (30° anteversion). The decrease of the resisting moment with higher lateral abduction indicates that there is a risk of dislocation without impingement at very steep abduction angles. The smaller the developed resisting moment, the higher is the risk of dislocation.

3.3. Head size

Increasing the femoral head size led to decreased contact stresses at the liner during subluxation. Using a head diameter of 28 mm, the resulting hip joint force in mode 1 produced a maximum contact pressure of 3.25 MPa, which was reduced to 1.25 MPa by using a head diameter of 40 mm. Accordingly, the development of the resisting moment (Fig. 7) showed an increased impingement-free ROM as well as higher maximum resisting moments with larger femoral head diameters. In mode 1, i.e. external rotation in combination with 10° extension and 15° adduction, the impingement-free ROM rose from 59° (28 mm head) to 86° (40 mm head). In mode 2, i.e. internal rotation in 90° flexion and 0° abduction, the impingement-free ROM rose from 2° to 18°. Additionally, in this mode the maximum resisting moment increased from 3.46 Nm to 4.98 Nm.

In all cases during subluxation, the yield stress of UHMW-PE (23.56 MPa) was greatly exceeded at the impingement site. The plots in Fig. 8 show the stress distribution on the inner surface of a small diameter liner (28 mm head) and a large diameter liner (40 mm head). The stress states at 90° flexion and 30° internal rotation are shown. Cup position is 45° abduction and 15° anteversion. Since the diameter of the
Fig. 7. The variation of the resisting moment of four head sizes from 28 to 40 mm diameter with angle of internal rotation. Cup position is 45° abduction and 15° anteversion. The analytical results from internal rotation in 90° flexion and 0° abduction (mode 2) are shown. Note the later impingement and higher resisting moments of larger heads.

femoral neck is held constant, stresses at the impingement site are high for both models. However, with larger head diameters, the stresses decrease at the articulating surfaces and at the egress site. The larger and shallower indent found with large heads results in markedly lower stresses being developed at the egress site. The yield stress is exceeded at a markedly smaller area than with smaller heads.

4. Discussion

After total hip replacement the femoral head size and the neck diameter are considered important factors influencing the postoperative range of motion and the risk of dislocation [3,7,16–19]. In experimental and numerical studies, the range of motion and stability against dislocation of THR was evaluated with regard to the femoral head design [8,16,17]. In the present finite element study the dislocation stability and the stress distribution in the acetabular liner as a function of different head sizes was analysed at various implant positions. The implementation of analytical surfaces in the FE-model revealed new possibilities in the investigation of total hip dislocation compared to previous numerical analyses. In previous analyses the rigid surfaces were discretised with finite elements [7,20]. In our study, the time period for finite element calculation could be reduced to one quarter of our prior investigations through the use of analytical surfaces instead of finite elements. Using analytical rigid surfaces the stress plots showed evenly distributed force transmission in the joint. The different dislocation modes [11] were obtained by rotating the femoral component in the fixed cup. Good agreement between the numerically calculated range of motion and resisting moments compared to the experimental data was obtained. The progression of the resisting moment was found to be a valid parameter for describing dislocation stability.

The relationship between resisting moment and rotational angle allows determination of the maximum moment, the range of motion until impingement and the range of motion until total dislocation. This data provides predictions for more optimum orientations and design features of total hip implants.

For both calculated dislocation modes, a position of 45° lateral abduction and 15° up to 30° anteversion of the acetabular cup resulted in appropriate ROM and dislocation stability.

Since recurrent impingement can cause damage to the liner and induces dislocation [17], a maximum range of motion until impingement should be a goal for new implant designs being developed. Different attempts to avoid dislocation by deeper head inset or partial enhancement of the liner rim result in higher resisting moments, but a reduced range of motion [5]. The present analysis showed that stresses well above the yield strength of UHMW-PE occur at the impingement site. Such stresses lead to plastic deformation and spalling of the polyethylene liner. The simulation of larger heads showed reduced contact stresses at the egress site of the liner in case of subluxation as well as later occurrence of lever-out. These calculations indicate that larger heads may potentially cause less damage to the liner.

However, Wang et al. [21] found that larger femoral heads cause higher amounts of polyethylene wear than do smaller heads, when coupled with conventional UHMW-PE liners.
Nevertheless, the low wear rate of the newly developed, cross-linked polyethylene liners in combination with larger heads [22] is promising. In low wear bearing couples such as metal-on-metal and ceramic-on-ceramic, the use of larger heads is also preferred due to lower contact pressures resulting in less risk of mechanical failure, e.g. breakage of ceramic heads [23]. Under unfavourable circumstances, the prosthetic head can slip out of the cup without impingement [6,9]. In the present study, this effect did not occur, but as soon as the head is about to slip out of the cup without impingement, numerical singularities in the finite element formulation will cause the calculation to abort. Hence, an abortion of the calculation is a hint for occurrence of dislocation without impingement. A wide variety of implant designs can be analysed using the newly developed finite element model that uses analytical surfaces. This methodology allows evaluation of new designs much more rapidly than the analyses using rigid finite elements or even experimental techniques. This enables simplification of design variations since prototyping and assembling are replaced by prompt numerical simulation. Optimal positioning and appropriate design of the implants should have as their goal the avoidance of impingement-related failure, while at the same time maximizing the range of motion and in addition providing maximum resistance to dislocation. Particularly with regard to the two dislocation prone manoeuvres including anterior and posterior dislocation in this study, we emphasize the requirement of asymmetric cups. The finite element model can help to identify the best compromise between impingement-free ROM and high resisting moments.

**Conflict of interest statement**

There are no conflicts of interest.

**References**