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THA loading arising from increased femoral anteversion and offset may lead to critical cement stresses

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Abstract

Aseptic loosening of artificial hip joints is believed to be influenced by the design and orientation of the implant. It is hypothesised that variations in implant anteversion and offset lead to changes in the loading of the proximal femur, causing critical conditions to both the bone and cement. The goal of this study was therefore to analyse the role of these parameters on loading, bone strains and cement stresses in total hip arthroplasty (THA). A validated musculo-skeletal model was used for the analysis of muscle and joint contact forces during walking and stair climbing. Two different anteversion angles (4° vs. 24°) and prostheses offsets (standard vs. long) were analysed. The loads for each case were applied to a cemented THA finite element model. Generally, stair climbing caused higher bone strains and cement stresses (max. +25%) than walking. Variations in anteversion and offset caused changes in the loading environment, bone strain distribution and cement stresses. Compared to the standard THA configuration, cement stresses were raised by increasing anteversion (max. +52%), offset (max. +5%) and their combination (max. +67%). Femoral anteversion, offset and their combination may therefore lead to an increased risk of implant loosening. Analyses of implant survival should consider this as a limiting factor in THA longevity. In clinical practice, implant orientation, especially in regard to pre- and post-operative anteversion, should be considered to be more critical.

Keywords: Anteversion; Joint geometry; Muscle forces; Total hip arthroplasty; Finite element analysis

Introduction

Total hip arthroplasty (THA) is one of the most successful procedures in orthopaedic surgery with respect to immediate pain relief and re-establishment of the joint function [2]. However, long term survival of the re-constructed joint is influenced by factors like the prosthesis design [24,25], the quality of the bone stock [30], the degree of patient activity [38] and surgical aspects such as the orientation of the implant [7]. In cemented hip reconstruction, these factors may contribute to the predominant failure mode, aseptic loosening [23]. Results from radiographic analysis and implant retrieval studies suggest that the first loss of fixation occurs at the prosthesis–cement interface. It is assumed that the applied loads are a major contributor to the initiation of failure [1,18,33]. These loads, caused by muscle and hip contact forces, affect the femoral strain distribution [12,13] and the magnitude and orientation of cement stresses [37].

It is accepted that modifications in joint geometry have an impact on, e.g. joint function [21], primary stability [8] and bone re-modelling [41]. Femoral anteversion and the offset (orthogonal distance between the centre of rotation of the hip and the femoral axis) both contribute towards characterisation of the hip geometry. These factors influence the loading conditions at the joint and consequently the bone straining [12,13].

Femoral anteversion is a parameter that is under the control of the surgeon during THA. It is suggested that femoral anteversion influences the function of the hip joint and plays an important role in the loading of endoprostheses [22], and consequently in the outcome of THA [4,19,31]. It is assumed that proper joint reconstruction by means of implant positioning is capable of preventing dislocation of THA [11]. Differences of up to 22° between the pre- and post-operative femoral anteversion have been measured [35]. Such differences may increase loading, which would be most prominent

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during repetitive daily activities, such as walking and stair climbing [3,28] and therefore influence the longevity of the implant [5]. The role of anteversion on THA loading, its potential influence on the stress transfer between implant and bone and its contribution to the survival of the artificial joint has so far not been evaluated.

A variation in prosthesis offset allows the surgeon to re-construct the femoral joint with respect to individual patient anatomy. Like femoral anteversion, changes in the offset can affect the loading of the joint. An increase in offset could raise the stability of the joint [34] and reduce the hip contact force due to the longer lever arm of the abductor muscles [9,27,34]. This idea has been supported experimentally [9]. However, offset alterations cause ambivalent results: On the one hand, a reduction of the muscle forces is to be seen as a positive effect on the primary and long term stability of the artificial hip joint. On the other hand, this same increase in offset might cause higher bending and torsion loads despite lower joint contact forces of the artificial joint [9,36]. In addition, the prosthesis offset seems to influence the use of the artificial hip joint: in patients with bilateral THA, significantly higher polyethylene wear has been found on the side of decreased post-operative offset compared to the side on which the offset was maintained [34]. The explanation given was that similar femoral offset before and after surgery tended to restore pre-operative hip biomechanics more closely. In light of these results, the femoral offset appears to contribute to the loading conditions at the hip and therefore probably influences the outcome of THA.

The hypothesis of this study was that variations in implant anteversion and offset can lead to changes in the loading of the proximal femur that may cause conditions critical to bone straining and increase the risk of cement failure. A better understanding of these parameters, which may limit the life-span of THA, may draw attention to possible improvements to current surgical procedures. The goal of the present study was therefore to quantify the influence of both femoral anteversion and prosthesis offset on bone and cement loading.

Methods

Musculo-skeletal analysis

A numerical model was used to determine the musculo-skeletal loading conditions for different angles of femoral anteversion and prosthesis offsets. The method has been presented recently [20] and is therefore only briefly summarised here: An instrumented femoral prosthesis was used to measure the in vivo hip contact forces in four patients (mean 61 years). Clinical gait analysis was conducted for six trials of both walking and stair climbing and time dependent kinematics and kinetics data were gathered. The in vivo hip contact forces were measured during all activities. An optical system (Vicon, Oxford Metrics, UK) consisting of six infrared cameras and 24 reflective markers attached to the patients' skin was used to determine movement of the lower limbs. A computer model of the human lower extremity (CT-Data Visible Human, NLM, USA) consisting of bony structures and muscles, was then scaled to all patient anatomies. Muscle wrapping was included where appropriate. The forces in each muscle during both activities were then computed using a numerical optimisation algorithm. The calculated forces were validated with the experimental in vivo data [3,20].

In this study, the hip joint anatomy of the musculo-skeletal model was then altered to analyse the influence of femoral anteversion and offset. The gait patterns and ground reaction forces were assumed to be identical, despite the modifications to the hip joint geometry. A complete re-analysis of the muscle and joint contact forces was then performed for each of the modifications for all four patients using optimisation that minimised the square of the muscle stresses. One representative patient (male, bodyweight of 878 N, 4° angle of femoral anteversion) was selected for the geometry for the following finite element analysis.

Finite element model

Based on CT-scans from the Visible Human (NLM, Bethesda, USA), both endosteal and periosteal contours of the femur were determined by means of thresholding methods. The geometry of the bone was then scaled to the anatomy of the representative patient femur [20]. Finite element (FE) models of both the intact and THA situations (Fig. 1) were then generated using TrueGrid software (XYZ Scientific Applications, Livermore). A normal neck and a long neck version (+4.8 mm medio-lateral; Fig. 1) of a tapered, collarless and polished stem prosthesis (MS-30, Sulzer Orthopedics Ltd., Switzerland) were meshed. The implants were inserted in two different positions: 4° anteversion (based on the specific patient THA data) and 24° anteversion, a position of maximum rotation, limited by anatomy (Fig. 1). The gap between prosthesis and bone, which had a mean thickness of 3 mm, was filled with cement elements. A cement void was present distal to the tip of the stem, as would be created by a centraliser. The boundary between the prosthesis stem and cement was modelled as a fully debonded interface using coulomb friction, with a friction coefficient of 0.25 [26].

A Young's moduli of 17 GPa was assigned to the cortical bone together with a Poisson's ratio of 0.33 [32]. The properties of the trabecular bone were graded from proximal to distal in four steps (2.0, 1.0, 0.5 and 0.25 GPa), also with a Poisson's ratio of 0.33. The cement properties were taken from the prosthesis manufacturer data ($E = 2.6$ GPa), whilst the stainless steel (Protasul S30, Sulzer Orthopedics Ltd., Switzerland) stem was assigned a Young's modulus of 200 GPa. A Poisson's ratio of 0.3 was used for all artificial materials. All materials were assumed isotropic and linear elastic in behaviour. The complete THA model with the standard prosthesis offset contained 12546 eight-node brick elements; the model with increased prosthesis offset contained 12768 elements. The models were analysed using the MARC/Mentat software (MSC, Palo Alto, USA).

Convergence tests were performed on the THA reference FE mesh, with 4° anteversion and the standard prosthesis offset (Table 1). The elements were refined in the proximal region of the model in two steps: The number of elements in the cortical bone, cement and the prosthesis in contact with the cement was doubled through refinement in the medio-lateral direction. In a second refinement, the number of elements was doubled in the anterior–posterior direction.

Loads

The muscle attachment sites and forces from the musculo-skeletal model were transferred to discrete nodes on the surface of the finite element bone model. The hip joint force was applied to the centre of the femoral head or the centre of the prosthetic head, in the intact and THA models respectively. The knee contact force was distributed equally between the condyles of the femur. The equilibrium of the moments and forces, determined by the musculo-skeletal analysis, was slightly disturbed when the forces were placed onto the finite element model due to differences in the positions of the muscle attachment sites. To re-establish the equilibrium of forces and moments for the finite element model, up to three small forces (each <5% of maximum applied loads) were added. Rigid body motion was constrained at three points on the femoral condyles.

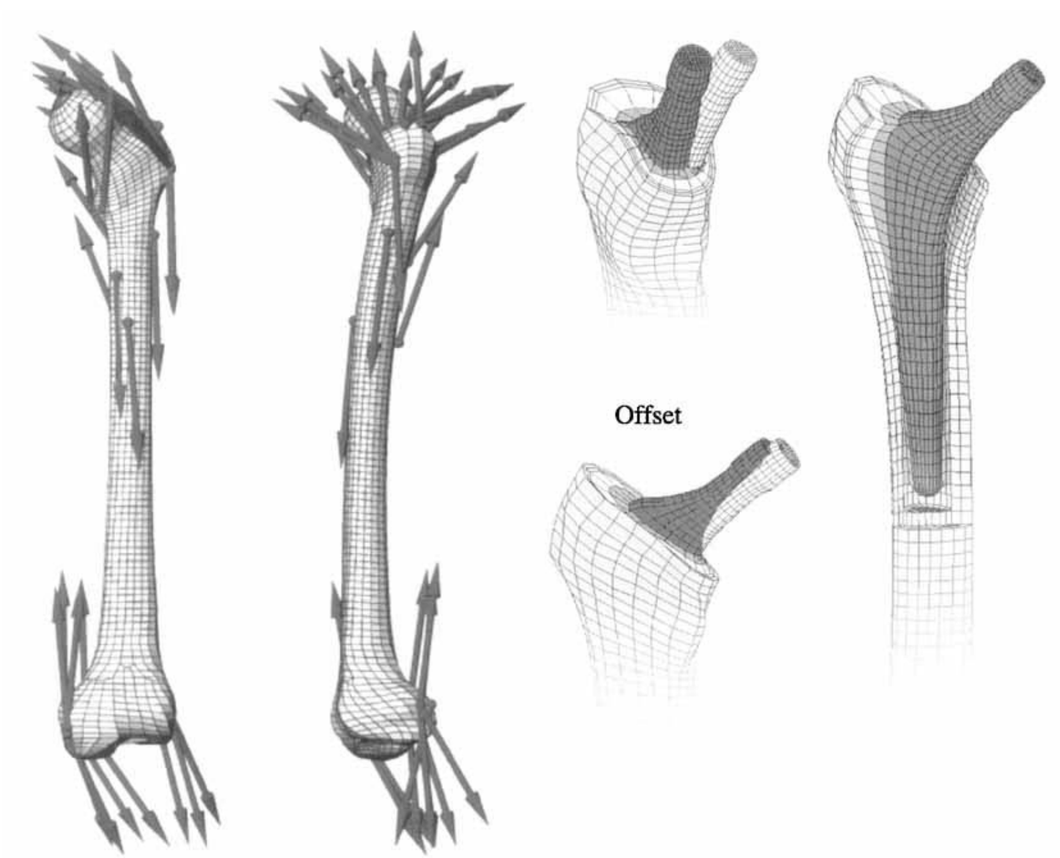

Fig. 1. Finite element mesh including all muscle forces and THA re-construction with a cemented polished tapered stem (MS-30, Sulzer Orthopedics Ltd., Switzerland). Vectors show the orientation of the applied muscle forces. Views of the anteverted (4° and 24°) and increased offset (+4.8 mm medio-lateral) configurations are detailed, together with an open section of the proximal bone.

Table 1
Combinations of geometrical and gait cycle parameters used within the different analyses: Intact, THA reference, anteverted, increased offset and their combination

Analysis	Gait cycle (%)	Anteversion angle (°)	Prosthesis offset
Intact femur	15, 45	4	–
THA reference	15	4	Standard
Increased offset	15	4	Long
Anteverted	15	24	Standard
Combined	15	24	Long

Straining of the bone at the peaks of the gait cycle loading for both walking and stair climbing was analysed to establish the worst case loading scenarios. These cases were then used for the comparative analyses.

In order to analyse the influence of different femoral anteversion and prosthesis offsets, five different numerical analyses were run (Table 1). Cortical bone strains were analysed by tracing the values at the surface nodes from proximal to distal. Minimum principle (tensile) stresses throughout the complete cement layer were analysed by calculating the arithmetic mean stress of each element at all eight gauss points. Additionally, cement stresses in specific regions of clinical interest (e.g., calcar and tip regions) were evaluated as these regions are considered important for the longevity of artificial joints [42]. The magnitudes of the stresses in the cement mantle were examined for peak tensile stresses over the assumed cement fatigue strength of 8 MPa [7]. The stress range of 3–10 MPa was examined in particular, as this is assumed to be responsible for cement crack initiation and damage accumulation under cyclic loading [17].

Results

The mesh convergence under stair climbing loading conditions (at 15% gait cycle) resulted in a mean decrease in cortical bone surface strains of 2.6% after the first element refinement and 2.7% from the second with respect to the reference results. These differences in strains were considered small enough to allow the initial element size to be adequate for the finite element study.

Intact femur loading

The examination of gait data showed two peak loads at 15% and 45% of the gait cycle during walking and one peak load at 15% of the gait cycle during stair climbing. Although the muscle and hip contact forces were higher at the 45% instant, the maximum bone strains were found to be higher at 15% of the cycle, in both activities

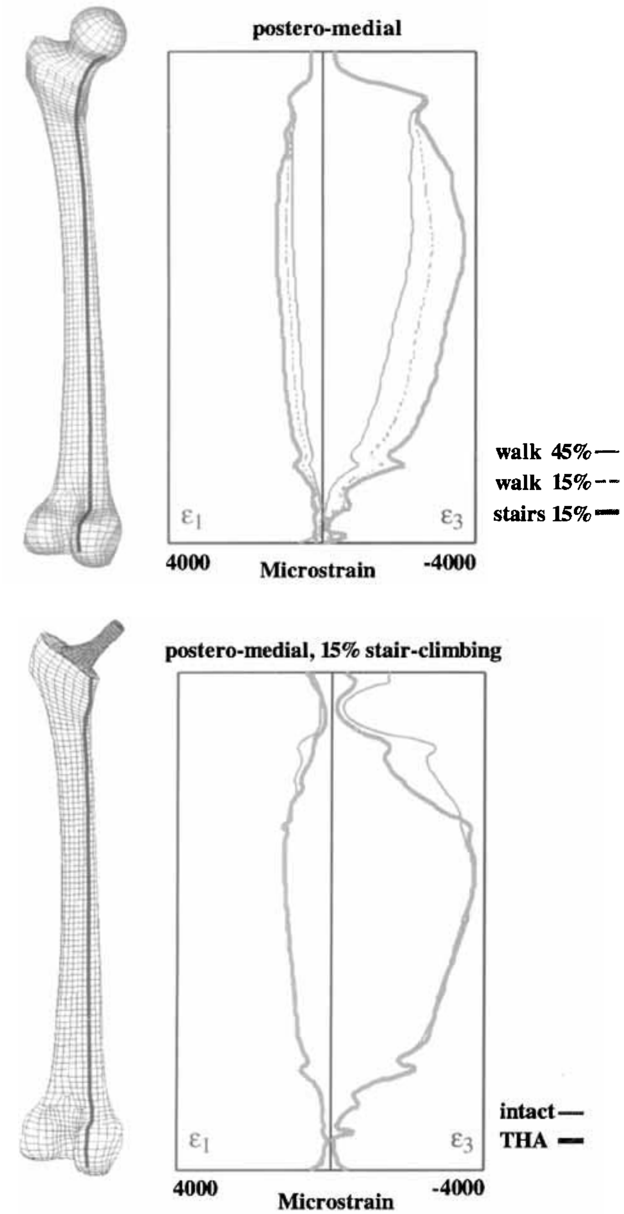

Fig. 2. Principal strains ε_1 (tensile) and ε_3 (compressive) in microstrain of the postero-medial aspect of the human femur at 15% and 45% of the gait cycle during walking and at 15% of the gait cycle during stair climbing (top). Tensile and compressive strains of the implanted femur at 15% of the gait cycle demonstrate unloading of the proximal bone (bottom).

(Fig. 2, top). To simulate a worst-case scenario at the bone-implant interface, loads were applied at the 15% instant of the gait cycle in all models.

Intact femur vs. THA reference

In comparison to the intact femur, the introduction of an implant at 4° anteversion and standard prosthesis offset reduced the principal surface bone strains of the proximal femur (Fig. 2, bottom). Maximum surface bone strains of up to 3800 microstrain ($\mu\varepsilon$) were found in the postero-medial region in both walking and stair climbing exercises. The smallest strains were observed in the anterior region. By dividing the cement elements into discrete stress ranges for the THA reference case, it was shown that more than 80% were found in the 0–3 MPa range (Fig. 3). Almost 18% of the elements were found in the range 3–10 MPa. Only a small percentage (approximately 2%) were above 10 MPa.

Anteversion, offset and their combination

Increasing the prosthesis anteversion from 4° to 24° caused higher muscle and joint contact forces (Table 2), resulting in an increase in bone strains of up to 16%. The maximum strains in the proximal bone shifted from postero-medial to medial. At the same time, the average cement stresses were increased by about 52% during walking and 35% during stair climbing (Fig. 3).

Despite lower muscle and joint contact forces (Table 2) the FE models with an increased prosthesis offset showed a minor increase in strains at the bone surface (up to +5%). Only small changes were found in cement stresses (up to +9%) and their patterns (Fig. 3).

Combining increased femoral anteversion and larger prosthesis offset during walking had a similar effect as increased anteversion alone: The stress magnitudes in the cement mantle were almost doubled compared to the THA reference case and the case with increased offset alone (Fig. 3). During stair climbing, however, the increased loads caused substantial rises in cement stresses (up to +67% mean cement stress) and a minor increase of bone strains (up to +19%). A long prosthesis offset together with increased anteversion raised the percentage of elements with cement stresses in the range responsible for damage accumulation (3–10 MPa) from 19% to 51%.

Having examined the distribution of cement stresses throughout the mantle, it was observed that three main regions of high stresses existed. Therefore, in addition to the clinical regions of interest (around the tip of the stem and in the calcar), values were also calculated at the distal–medial aspect of the stem (Fig. 3). Examination of these regions showed mean stresses of almost 50% greater than in the complete cement mantle: When analysing the combined effects of large anteversion and increased offset, nearly 80% of the elements in these regions were found to be within the 3–10 MPa range. The number of elements with stresses greater than 10 MPa was below 3% for all analyses. The peak cement mantle stresses were observed at the distal tip of the stem and in the calcar region. Cement stresses in the calcar and regions medial and lateral of the implant tip locally exceeded the assumed cement fatigue strength of 8 MPa under the reference THA conditions, but did not change after modifications in anteversion and offset were made.

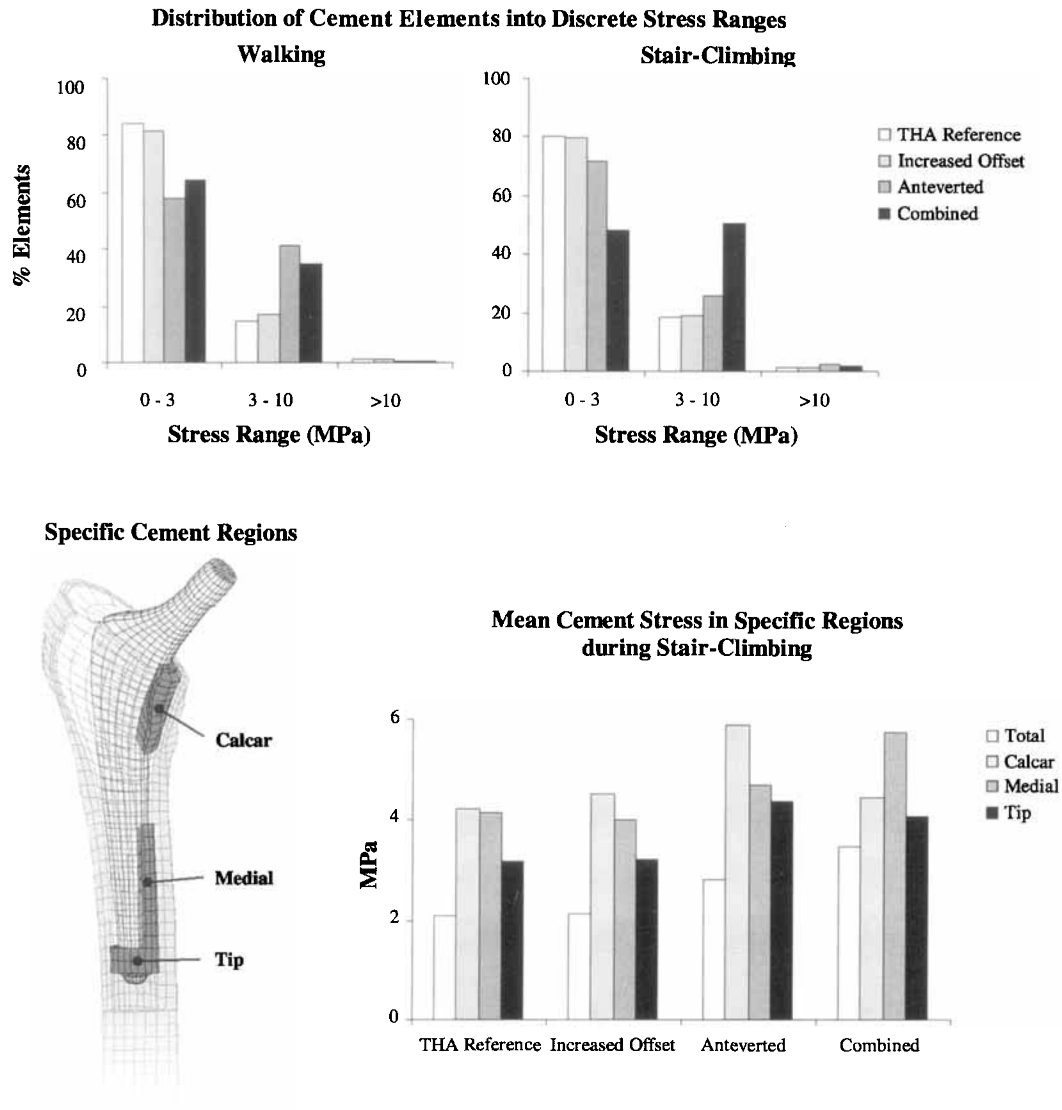

Fig. 3. Distribution of cement elements into discrete stress (minimum principle, i.e. tensile) ranges for different loading scenarios during both walking and stair climbing (top). The 3–10 MPa range is considered critical for cement damage accumulation. The stresses in the specific regions of interest within the cement mantle are shown for the different loading configurations (bottom).

Table 2
Hip joint contact forces for the different loading scenarios

Loading case	Gait cycle (%)	Hip contact force in bodyweight (BW)					
		Walking			Stair climbing		
		Medial–lateral	Posterior–anterior	Superior–inferior	Medial–lateral	Posterior–anterior	Superior–inferior
Intact femur	45	0.96	−0.16	3.02	–	–	–
Intact femur and THA reference	15	1.08	0.25	2.89	1.14	0.32	2.90
Increased offset	15	1.05	0.25	2.79	1.00	0.30	2.75
Anteverted	15	1.40	0.59	3.38	1.87	1.12	4.46
Combined	15	1.37	0.60	3.20	1.87	1.08	4.01

Joint contact forces are altered as a direct result of the changing muscle forces derived from the musculo-skeletal analyses.

The non-linear behaviour of the interface caused slip between the implant and cement (Reference THA walking: 44 μm), which increased by alterations in anteversion and offset. Maximum relative displacements were observed in the analysis with increased anteversion during walking (127 μm) and with the combined increased offset and anteversion during stair climbing (157 μm). The displacement at the tip was always greater than the calcar region (mean +25%).

Discussion

This study has examined the hypothesis that changes in implant anteversion and offset may lead to critical loading conditions in the proximal femur. By applying physiological-like musculo-skeletal loading to THA finite element models, it has been possible to investigate the influence of different femoral anteversion angles and prosthesis offsets on bone straining and cement stresses in THA.

It has been suggested that femoral anteversion plays an important role in load transfer between prosthesis and bone [15]. Changes in bone loading can lead to bone re-modelling [41], possibly causing degeneration [31], resulting in a higher risk of implant loosening. The influence of femoral anteversion on bone loading has recently been described: Increasing the angle of anteversion causes an increase in bending moments and hip contact forces [19]. The present study has confirmed these conclusions by showing that muscle and hip contact forces increase with anteversion, but has further demonstrated a remarkable increase of cement stresses under these conditions, especially during walking. The transfer of these larger stresses between implant and bone seems likely to raise the risk of implant failure. In contradiction to the hypothesis of this study, however, only small changes in bone strains were observed between the different implant configurations.

A variation of the prosthesis offset (standard vs. long) resulted in only minor changes of bone loading and cement stressing. It was observed that increasing prosthesis offset reduced the muscle forces and consequently the joint contact forces, findings that were in agreement with a simplified experimental study [9]. From a surgical aspect, it is assumed that an increase in implant offset results in tightening of the soft tissues due to stretching, thus raising the stability of the joint [34]. These changes, however, may influence the amount of polyethylene wear [34] and the increased lever arm of the hip contact force can result in pronounced bending of the implant [36]. Despite a decrease in muscle forces with increased prosthesis offset, the present work did show a small increase in cortical bone strains and cement stresses. Nevertheless, it should be noted that any increase of stresses might raise the risk of implant loosening [6]. It should also be noted that the difference in offset between the two cases examined in this study was small and other clinical offset situations may produce larger changes in the stresses and strains.

It is well known that the initiation of cement failure correlates with the applied loads and the cement interface and integrity [1,33]. Combining increased femoral anteversion and larger prosthesis offset resulted in a substantial increase in loading and especially in cement stresses during stair climbing: the number of cement elements with stresses in the range of critical failure was almost doubled (Fig. 3). The larger loading of the cement mantle under these conditions, therefore, is likely to prove detrimental to the artificial joint.

The magnitude of peak stresses rose with the larger offset but was greatly increased when additional anteversion was added. The role of peak cement stresses is assumed to be minor, however, due to the supposed stress reduction caused by cement creep [39]. Nevertheless, it seems that the increase of stresses into the range of 3–10 MPa by modifying anteversion alone and in combination with offset is important for damage accumulation under cyclic loading and consequently increases the risk of implant failure [40].

This analysis has demonstrated that changes in implant orientation and design are capable of causing substantial rises in cement stresses, most importantly in the critical regions (e.g., the calcar). Results from radiographic analyses and implant retrieval studies suggest that the first loss of fixation occurs at the prosthesis–cement interface by crack initiation in the calcar region [14]. The present findings have emphasised that the calcar cement region and the cement around the tip were sensitive to the investigated parameters. In contrast to other studies [16], however, no noteworthy regional differences in cement stress patterns were observed between walking and stair climbing.

The influence of the patient's exercise on bone loading has recently been presented [3]; stair climbing caused greater muscle and joint contact forces than walking and consequently greater loads. The results of the present study underlined this assumption: in most cases, stair climbing produced higher loads, higher bone strains and higher cement stresses than during walking. Previous studies have used simplified loading scenarios with the hip contact force and one or two muscle forces or did not include musculo-skeletal loading during stair climbing [9,10]. Consequently, analyses of THA performance could have more impact by reporting results during stair climbing. For pre-clinical investigations it should considered that loads during stair climbing caused the greatest effects in comparison to other routine activities [3].

Although this study has included a considerable number of muscle forces, it should be noted that modelling their distribution is only an approximation to the physiological situation. The knee contact force was split equally to maintain equilibrium of moments in the finite element model. A distribution of the condyle loads similar to the physiological situation would deviate from the joint kinematics of the musculo-skeletal model. Material properties were also simplified: Isotropic, homogeneous and linear elastic material behaviour was assumed, although the future inclusion of anisotropy is being pursued. In addition, the calculation of muscle forces throughout all variations of hip geometry was based upon the patient gait patterns recorded for their specific hip geometry. These gait patterns may have been slightly different with the variations of hip geometries examined in this study, resulting in altered muscle forces. Differences in gait patterns after these variations, however, are expected to fall well within the limits of intra-individual variation between repetitions of gait cycles. Modelling techniques, however, may only ever approximate the physiological situation, and these limitations should be taken into account when interpreting the results. Nevertheless, as a comparative study, the impact of modelling simplifications has been reduced as the parameters were consistent between models.

The present study is limited to a single set of interface conditions, a single prosthesis design and two variations of anteversion and offset. Thus, the results of the study suppose that other prostheses with similar configurations behave in the same manner, which is likely, but not necessarily true. The debonded interface condition was responsible for the slip of the implant within the cement mantle, which was three times higher in the worst-case loading scenario (24° anteversion and large offset during stair climbing) compared to the THA reference. However, the observed relative displacements were similar to those previously reported [29].

Clinically, the orientation of femoral stems seems to be essential for long-term performance in vivo. The results of this study indicate that anteversion plays a more important role in determining cement mantle loading than prosthesis offset. Femoral anteversion may therefore be considered a more influential parameter than offset in the long-term clinical outcome of THA, but their combination, especially during stair climbing activities, can produce critical cement stresses. In the clinical situation, these undesirable effects should be considered, and when an implant with a large offset is to be used, the surgeon should be careful to avoid large angles of femoral anteversion.

Acknowledgements

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