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Dynamic Virtual Simulation of the Occurrence and Severity of Edge Loading in Hip Replacements associated with variation in the rotational and translational surgical position

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Abstract

Variation in the surgical positioning of total hip replacement can result in edge loading of the femoral head on the rim of the acetabular cup. Previous work has reported the effect of edge loading on the wear of hip replacement bearings with a fixed level of dynamic biomechanical hip separation. Variations in both rotational and translational surgical positioning of the hip joint replacement combine to influence both the biomechanics and the tribology including the severity of edge loading, the amount of dynamic separation, the force acting on the rim of the cup and the resultant wear and torque acting on the cup. In this study, a virtual model of a hip joint simulator has been developed to predict the effect of variations in some surgical positioning (inclination and medial lateral offset) on the level of dynamic separation and the contact force of the head acting on the rim as a measure of severity of edge loading. The level of dynamic separation and force acting on the rim increased with increased translational mismatch between the centres of the femoral head and the acetabular cup from 0 to 4mm and with increased cup inclination angle from 45° to 65°. The virtual model closely replicated the dynamics of the experimental hip simulator previously reported, which showed similar dynamic biomechanical trends, with the highest level of separation being found with a mismatch of 4mm between the centres of the femoral head and acetabular cup and 65° cup inclination angle.

Keywords

Total hip replacement, edge loading, hip separation, inclination angle, surgical position

Introduction

Hip replacement is a successful surgical intervention for the treatment of arthritis, tissue degeneration and pain in the hip. An active ageing population and earlier interventions have led to increased expectations of longer life times and lower revision rates. Wear, osteolysis and loosening are recognised as late modes of failure and causes of revision in hip prostheses[1, 2]. Edge loading of the bearings in hip prostheses occurs when the bearing surfaces of the total hip replacement are contacting or articulating outside of the intended bearing area and when the head contacts the rim of the acetabular cup.

While the causes of edge loading are multifactorial, there are two dominant modes of action. The first mode results from impingement and lever out of the head onto the rim of the cup, which can often occur in less frequent extreme activities of motion. The second mode, which is simulated in this study, is edge loading without impingement; this is often associated with dynamic separation of the centres of the head and cup and can occur frequently in many activities such as standard walking[3-6]. Edge loading of the bearings in hip prostheses has been linked to increased wear for hard-on-hard (metal-on-metal and ceramic-on-ceramic) bearings in artificial hips[3, 4, 7-13]. In polyethylene acetabular cups, edge loading may cause increased deformation and strain in the polyethylene and the potential of fatigue fracture[14]. In all bearing types, edge loading can cause an increased torque on the cup and potentially cup loosening[15]. In all types of hip prostheses, edge loading may lead to an increased rate of revision and failure.

The occurrence and severity of edge loading may be caused by variations in rotational or translational surgical positioning of the components[10, 11]. Variation in rotational positioning of the cup has been represented and simulated by the variation in the inclination (abduction/adduction) angle in the coronal plane[10, 11, 16]. Both experimentally and clinically, increased cup inclination angle has been found to be one of the causes of edge loading and increased wear[8, 10, 11, 17-22]. Variation in the translational positioning has been simulated as a mismatch between the positioning of the centre of the cup and head, along the medial-lateral axis in the coronal plane[4, 8-12, 23-25]. Edge loading may occur without dynamic (micro) separation, due to variation in rotational positioning alone, but there is increasing evidence that a mismatch between the relative positions of the centres of the cup and head, leads to dynamic separation and severe edge loading, increased wear and damage and an increased torque at the cup surface[8, 10-12, 14, 15, 26].

Clinically, the level of dynamic separation and severity of edge loading can vary considerably[27]. To be consistent with previous work, microseparation is used here to describe dynamic displacement of the centres of the head and cup of less than 1 mm and separation to describe dynamic displacements greater than 1 mm. Subluxation occurs when the separation of the centres of the head and cup increase above 5 mm and may approach 10 mm or more, resulting, in the extreme, in dislocation.

The occurrence, the magnitude and severity of the dynamic separation and resultant edge loading, damage and wear are dependent on many biomechanical factors, including joint laxity, tissue tension, muscle forces, prostheses design and also surgical (rotational and translational) positioning[10, 11]. The effect of the combinations of these variables on the level of the dynamic separation of the head and cup during different activities, the severity of edge loading and the effect on damage, wear and loosening risk of failure are not fully understood. The complex interactions between different conditions and variables require further investigation. Previous experimental work has indicated that individually both translational and rotational positioning of the acetabular cup are important factors affecting the occurrence of edge loading and level of wear [10, 11] and these two variables form the focus of this study of the combined effects of variation in rotational and translational positioning on the occurrence and severity of edge loading. Experimental studies require expensive bespoke capital equipment and parametric studies of different variables and are costly and time consuming and require prototype parts to be made, while computational models can more easily address a range of variables and be adopted during the design phase of product development.

The aim of this study was to develop a virtual dynamic model of a hip joint simulator in order to predict the occurrence and severity of edge loading due to variations in rotational and translational surgical positioning of the acetabular cup in the coronal plane. The model was developed to predict the effect of variation in translational and rotational positioning on the magnitude of dynamic separation and the load acting on the cup-rim.

Materials and Methods

A virtual computer model of one station of the Leeds Mark II Physiological Anatomical Hip Joint Simulator was developed (Figure 1). The key components were created in the Parasolid XT format within the computer-aided design (CAD) package NX I-deas 6.1 (Siemens, UK) and these were imported into Adams/View 2013 (MSC Software Corporation, USA) to create the rigid, dynamic simulation model.

The device tested was a 36mm BIOLOX® delta ceramic-on-ceramic bearing (Pinnacle® design, DePuy Synthes, UK) with a diametrical clearance of 0.08mm. The edge of the liner (cup) was curved; a fillet with a 3mm radius was applied to the rim of the cup to replicate this curve- this was a simplification of the actual design. Surrounding setup included, a titanium shell behind the liner and PMMA cement holding the liner to the metallic fixture. The interfaces of the cup to cement and cement to cup holder were modelled as fully bonded, and no attempt was made to model clinical fixation interfaces to bone. This acetabular cup design was concurrently being tested in the laboratories[28]. The Adams model used rigid body dynamics with an elastic algorithm at the contact. The contact algorithm is based on a penalty method assuming a contact force that varies non-linearly with contact distance and a constant contact stiffness which can be adjusted by the user to give satisfactory numerical convergence and stability. No attempt was made to predict the contact stiffness, for example from Hertzian contact mechanics since such analysis is not appropriate for a highly conforming contact as pertains in the hip joint.

The femoral head was prevented from moving in all directions. The acetabular cup positioned superiorly to the femoral head and was fixed in a metallic holder which had a mass of 1.7kg. The cup and holder were free to translate in the transverse and frontal planes. A realistic contact friction value which was measured experimentally[29] was applied between the acetabular cup and the femoral head (friction coefficient of 0.05).

A gait cycle was simulated that comprised extension/flexion of -15°/+30° (applied on the femoral head) and a twin-peak dynamic vertical load (ISO14242-1; 3kN peak, 100N swing phase) that was applied through the centre of the cup at a frequency of 1Hz (Figure 3). The medial-lateral dynamic displacement was created by applying a spring element (of stiffness 100N/mm) to the lateral side of the cup holder, which caused the cup to move in the medial direction at low axial loads, as applied in the experimental simulation[28]. This spring stiffness used in the model was matched with the spring stiffness used experimentally to simulate edge loading conditions. The method used experimentally was previously validated against ceramic-on-ceramic retrievals by replicating similar stripe wear mechanisms and bimodal wear debris size distribution observed clinically[3, 4, 30, 31]. A translational mismatch between the centres of the femoral head and the acetabular cup was implemented as an initial, fixed compression of the spring.

Parametric tests were performed using the following set of inputs resulting in 30 different combinations:

- Acetabular cup inclination angles which are clinically equivalent to 45°, 55° and 65°[25].
- Translational component mismatch between the centres of rotation of the femoral head and acetabular cup of 0, 1, 2, 3 and 4mm.
- Bearing surface friction coefficient values of zero and 0.05.

The outputs measures from the model were:

- The magnitude of the dynamic separation between the centres of the femoral head and acetabular cup in the medial-lateral axis.

- The maximum load under edge loading, i.e., the maximum applied axial load (Figure 3) when the head is still in contact with the rim during the gait cycle

The magnitude of dynamic separation was defined as the maximum displacement between the centre of the cup and the centre of the head along the medial-lateral axis which occurred during the swing phase and early stance phase of the gait cycle. The maximum load under edge loading (the maximum force transmitted through the rim of the cup) occurred just after heal strike when the head was relocating back into its concentric position in the cup. The applied axial load was noted at a point where the magnitude of the reducing dynamic separation reached 0.2mm. At this point the contact between the head and the cup was still on the fillet radius of the rim of the acetabular cup and had not reached a concentric position within the acetabular cup.

Although clinically there are variations in joint laxity from one patient to another which may affect the occurrence and severity of edge loading, in this study the level of swing phase load and the spring stiffness driving the medial-lateral force were kept constant. In the presentation of results, predictions of separation displacements below 0.2mm were not shown as it was not clear that the model was able to predict movements or deformations at this level of sensitivity. The virtual simulation results are compared to previously reported experimental observations[28] of the hip simulator in the Discussion. The experimental simulation used 36mm diameter ceramic-on-ceramic bearings (BIOLOX® delta, DePuy Synthes, UK) which were tested on the Leeds II hip joint simulator. The ceramic liner was inserted into a titanium shell (Pinnacle®, DePuy Synthes, UK) and the femoral head was locked onto a stainless steel stem (C-Stem® AMT, DePuy Synthes, UK). The test ran under a standard gait cycle with a Paul-like twin peak loading profile with a maximum load of 3kN and swing phase load of approximately 70N. Extension-flexion (-15°/+30°) was applied to the femoral head and internal-external rotation (±10°) was applied to the acetabular cup. The lubricant used between the femoral head and acetabular cup was 25% new-born calf serum with protein concentration of approximately 15g/L). The axial load and the medial lateral spring load were measured using a compression only load cells and the medial-lateral cup displacement was measured using a linear voltage differential transformer (LVDT).

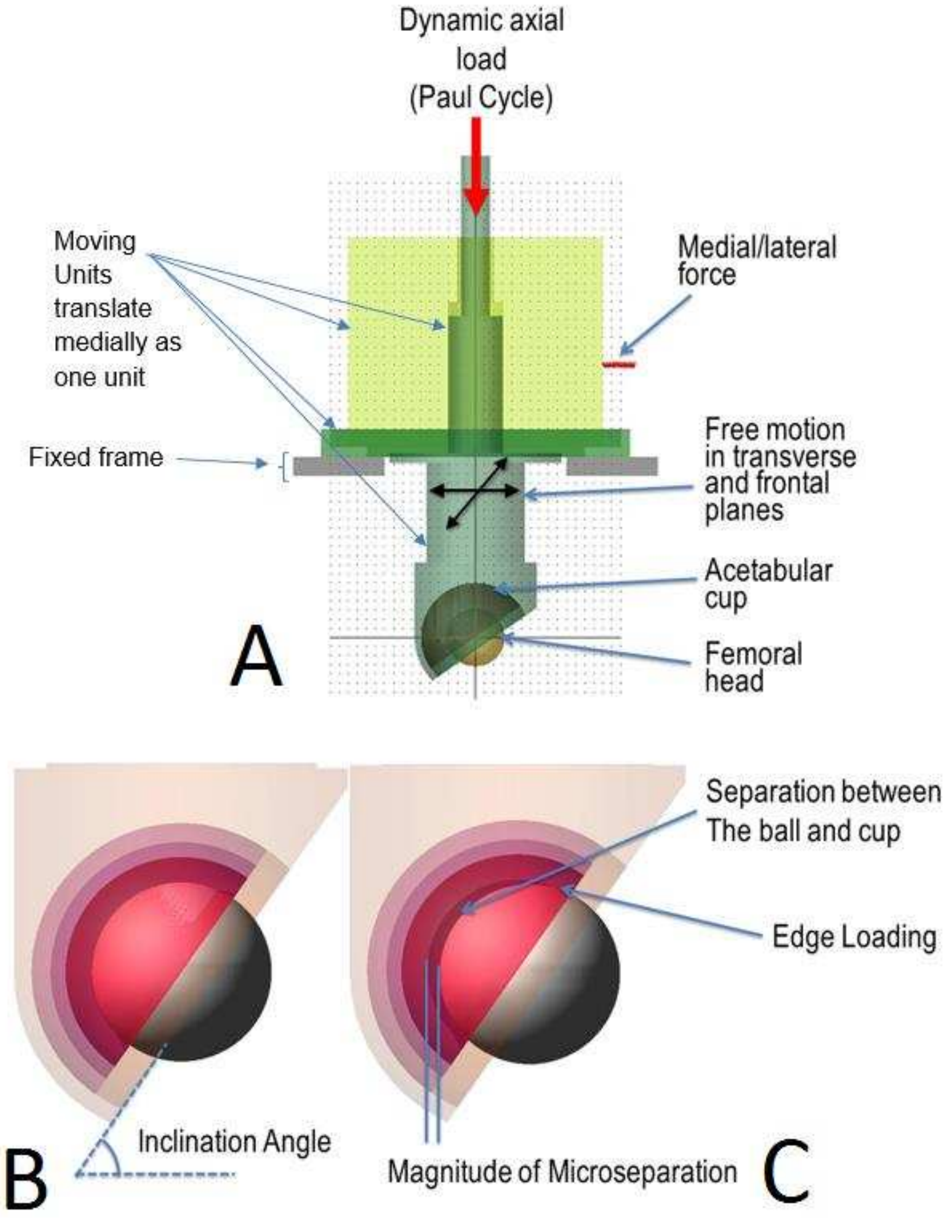


Figure 1: (A) The model in Adams™ software representing one station of the Leeds Mark II Simulator. The cup holder is allowed to move in the superior-

inferior axis while translating with the machine frame holding it in the medial-lateral axis. (B) The position of the femoral head and acetabular cup at peak loads. (C) The position of the femoral head and the acetabular cup under edge loading condition.

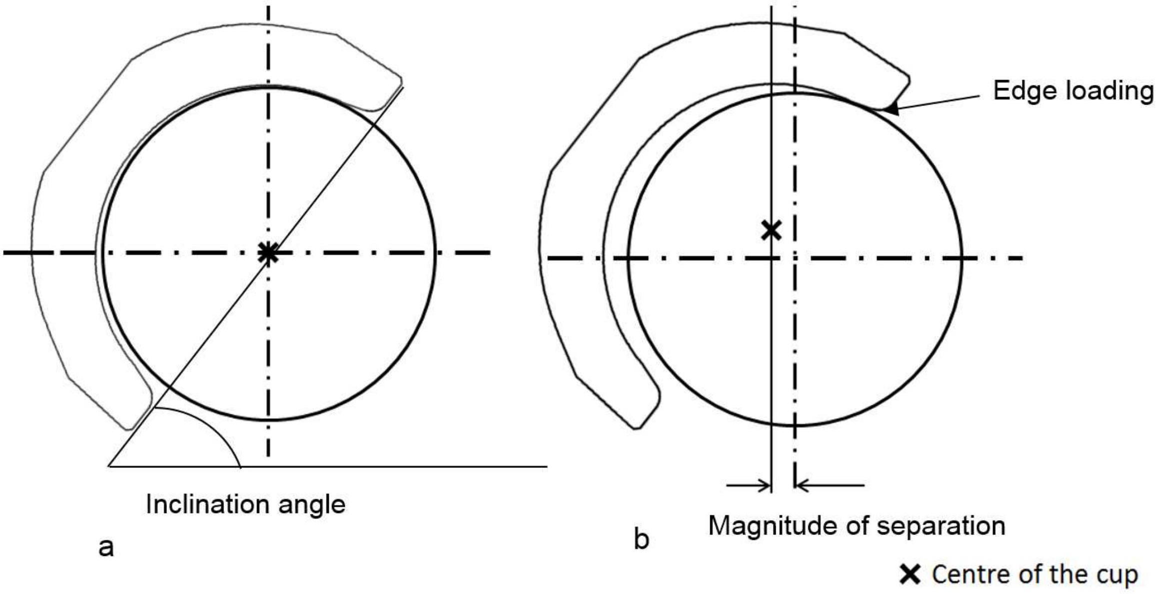


Figure 2: (a) a joint in its conforming, neutral position. (b) a joint in its non-conforming, edge loaded position.

Results

Both variation in rotational component positioning (steep cup inclination angle) and translational component positioning (mismatch in the head and cup centres of rotation) affected the occurrence and severity of edge loading conditions.

An increased level of mismatch between the centres of the head and cup produced an increased magnitude of dynamic separation for all cup inclination angle conditions (Figures 3 & 4, showing maximum separation for each in silico case). A positive linear relationship was observed between the magnitude of dynamic separation and the level of the translational mismatch between the centres of the head and the cup for each of the inclination angle conditions (Figure 4). These trends were similar for each cup inclination angle condition and for both of the contact friction cases, in that all were linear and has similar gradients. For a given variation in translational position (mismatch between the centres), the magnitude of the dynamic separation increased as the cup inclination angle increased (Figure 4). For any given translational mismatch, the magnitude of dynamic separation was found to be greatest for the largest inclination angle.

A combination of both a high medial-lateral translational mismatch and steep cup inclination angle caused the most severe edge loading condition, i.e., the load on the rim was greatest (Figure 5). Each line shows a positive relationship between the maximum force under edge loading and translational mismatch between the centres of the head and cup. The highest inclination angle provided the greatest maximum load under edge loading for each value of translational positioning. The results for maximum load under edge loading display the expected plateau towards the maximum applied load of 3000N.

An increase in translational mismatch of 1mm, e.g. from 2 to 3mm, increased the magnitude of dynamic separation by 1mm and produced an increase in the maximum force acting on the edge of the cup. The separation magnitude was more sensitive to the translational mismatch than to the cup inclination angle, within the range tested.

An increase in contact friction from zero to a contact coefficient of friction value of 0.05 caused a decrease in the magnitude of separation and also an increased maximum load under edge loading, i.e., the force on the rim was greater (Figures 4 & 5).

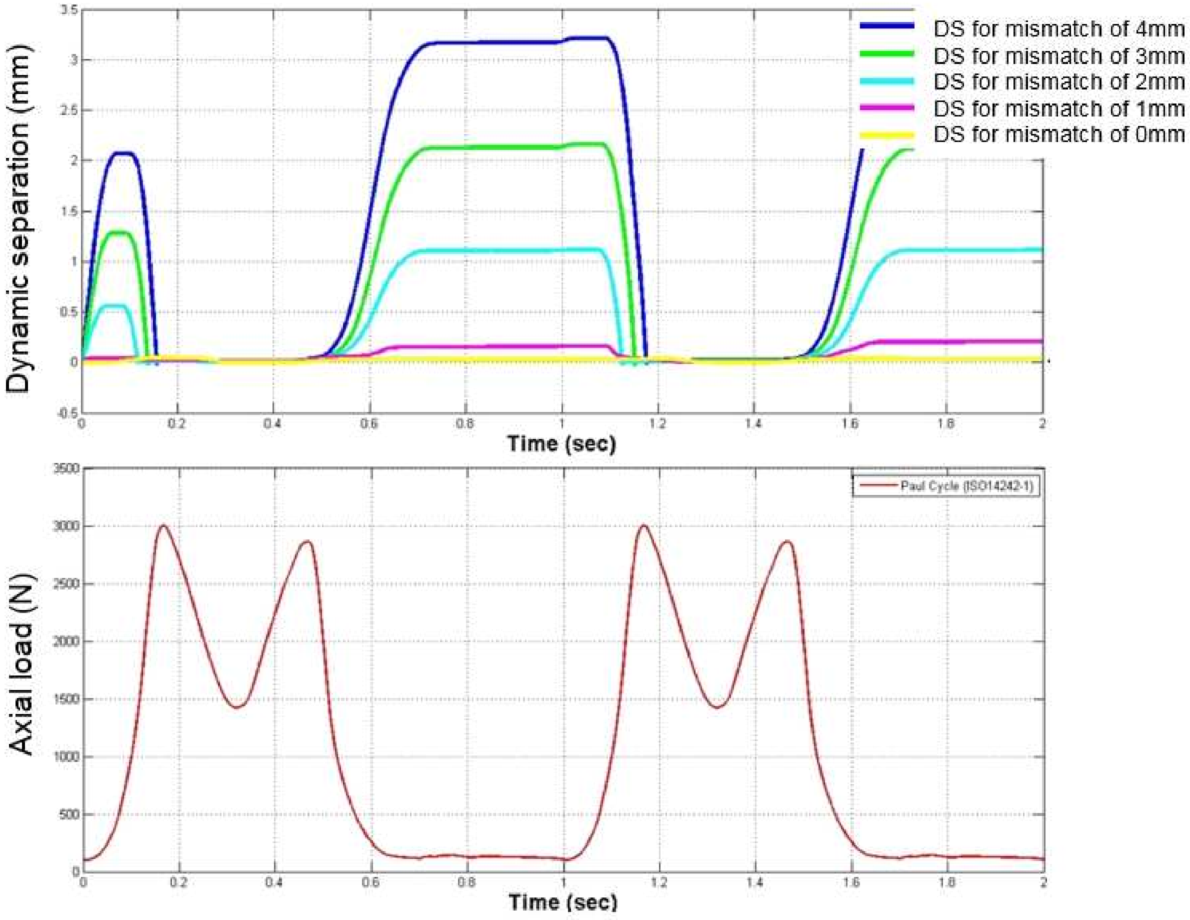
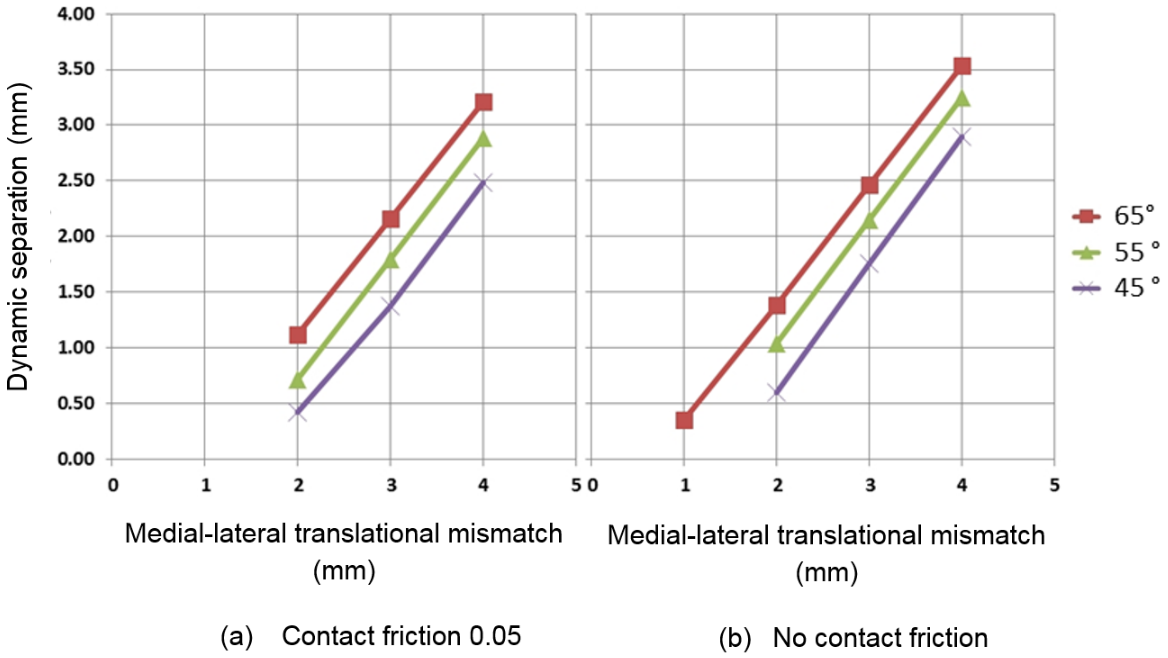


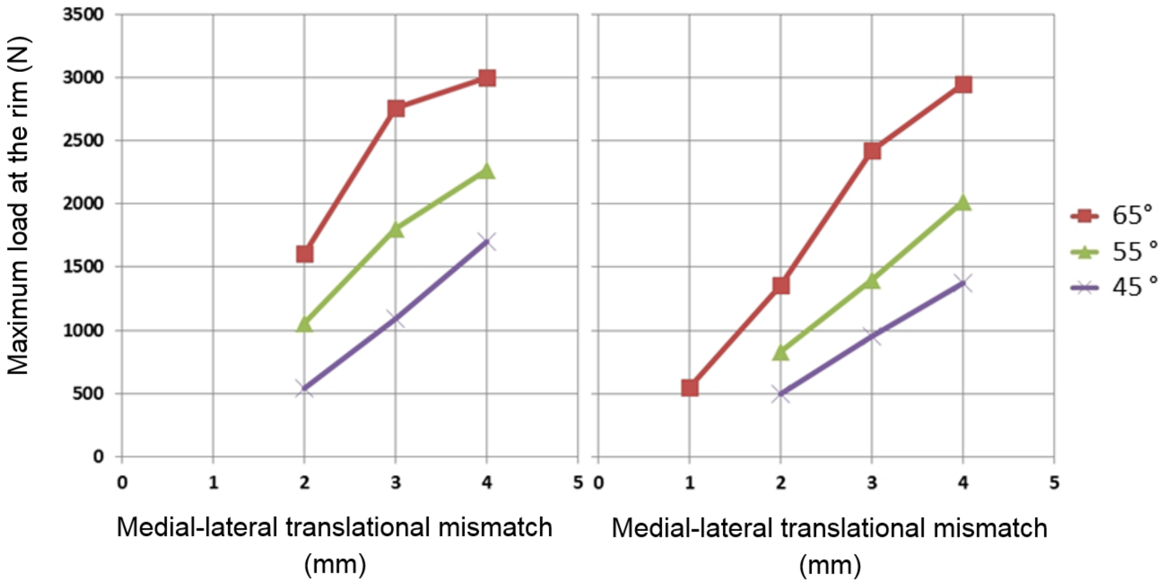
Figure 3: Plot of dynamic separation (DS) for 0, 1, 2, 3, 4 (mm) translational mismatch for the Pinnacle cup design at a cup inclination angle of 65° and with a swing phase load of 100N when the model had realistic contact friction coefficient between the acetabular cup and femoral head; shown with the twin peak dynamic downward load (Paul Cycle ISO14242-1).



(a) Contact friction 0.05

(b) No contact friction

Figure 4: Plots of the maximum dynamic separation versus medial-lateral translational mismatch between the centres of rotation of the head and the cup for the three different cup inclination angle conditions. Subplot (a) shows the results when the model had a realistic contact friction between the acetabular cup and the femoral head while subplot (b) is when the model had no contact friction. Each graph has three lines, one for each of the inclination angles. Each line describes a positive linear relationship between the magnitude of dynamic separation and the translational mismatch for each inclination angle.



(a) Contact friction 0.05

(b) No contact friction

Figure 5: Plots of the maximum load under edge loading versus medial-lateral translational mismatch between the centres of rotation of the head and the cup for the three different cup inclination angle conditions. Subplot (a) shows the results when the model had a realistic contact friction between the acetabular cup and the femoral head while subplot (b) is when the model had no contact friction. Some values at 1mm translational mismatch are not shown as the medial-lateral displacements were below the 0.2mm threshold taken as a minimum for a reliable measurement from the model.

Discussion

This is the first report of a computational model to predict the combined effect of both rotational and translational positioning of a hip implant on the dynamic separation and severity of edge loading. A virtual model was developed which represented the set-up established in the laboratory hip simulator to reproduce separation and edge loading in the absence of impingement[28]. This dynamic virtual simulation of the experimental hip simulator confirms the dependency of the severity of edge loading, the level of dynamic separation and magnitude of the load acting on the rim of the cup on both the rotational and translational positioning (observed in the experimental simulation) and predicts the relationships between the variations in the two positioning variables and the severity of edge loading.

The results of this virtual model showed very clearly that, as the mismatch between the head and cup centres increased, there was an increase in the severity of edge loading, the level of dynamic separation and the maximum load acting on the rim. A higher dynamic separation means it takes longer at the end of the swing phase and following heal strike for the head to move back off the rim of the acetabular cup, and during this longer period of time, the heal strike load on the rim is increasing rapidly (Figure 2). For 3 to 4 mm of translational mismatch, the maximum load on the rim starts to plateau at the peak load in the gait cycle. The cup inclination angle also has an effect on the severity of edge loading, with an increase from 45° to 65° increasing the level of dynamic separation by 1 mm and the maximum force by over 1000 N.

The limited experimental work reported to date on the coupling of the biomechanical and tribological effects has shown similar trends on the severity of edge loading, in the effects of the surgical positioning on the level of separation and the force acting on the rim[28]. Dynamic separation and edge loading could be clearly observed in the virtual model replicating the mechanics and dynamics produced in the experimental simulator (Figure 6). The experimental simulator results have also shown a substantial non-linear increase in wear as the severity of edge loading as the combined duration of dynamic separation and maximum load during edge loading both increase[28].

Figure 6 shows the comparison of this virtual prediction for severity of edge loading and dynamic separation and the experimental observations in the simulator published to date[28]. Reasonable agreement was found between the virtual model predictions and the experiment results, but the experimental observations of dynamic separation were lower than that predicted in the model. In the experimental model, the separation does not return to zero between cycles, while in the computer model it does. If the experimental results are measured as the displaced distance from zero, a closer agreement is observed between the experiment and the model (Figure 6). This variation observed in the experimental simulator may be due to additional friction in the experimental simulator mechanisms that reduce the rapid separation motions. The experimental results to date also indicate that there may be some non-linear effects on the relationship between the force acting on the rim and the increase in surgical mismatch. Experimentally a delay in relocating to a concentric position (possibly due to friction) increased the force acting on the rim and the wear, indicating the magnitude of dynamic separation alone may not be an adequate predictor of the severity of edge loading.

The computational model was matched to the experimental set up in many ways. Experimentally, 36mm ceramic-on-ceramic bearings were used similar in geometry to the one used in the model. The output loading profile from the experimental simulator was used as the input loading to the model. The weight of the cup holder was matched as well. There were also differences with key ones, including, the lack of internal/external rotation of the cup in the model compared to the experimental simulator. The simulator set up experienced friction within its moving bearings and bending that the model did not experience which might contribute to differences in the level of dynamic separation experienced by the components.

While clinical studies have previously shown an association between cup inclination and edge loading and wear[17, 18], other variables clearly come into play, such as the variation in the translational positioning of the head and cup centres[3-6, 10]. This remains the focus of ongoing research and consideration of development of approaches to improve the positioning of the centres of the head and the cup in vivo, the importance of translational positioning is a possible explanation of the additional variation observed in some clinical studies, which simply consider cup inclination.

Both this computational model and the experimental simulator model have limitations. Although both are three dimensional models, at the moment consideration is only given to positioning and dynamic medial-lateral separation in the coronal (medial-lateral) plane. Future work will develop models in three dimensional motion and separation associated with anterior-posterior loads and motions and variation in anterior-posterior translation positioning and variation in both version and tilt angles of the cup. The swing phase load in this study was set at 100N in line with the experimental simulator and in future computational studies this will be varied to investigate joint laxity and soft tissue tension. The medial-lateral spring stiffness was set at the value of the experimental simulator and this will be varied in the computational model in the future to represent a wider range of possible clinical conditions. Future modelling will also move to use finite element methods, which have the potential to predict the time dependent contact mechanics, and material stress and strain.

Variations in cup positioning and the resulting severity of edge loading, whether it be at a level of microseparation, separation or subluxation, the consequential additional wear, fatigue or increased torque remain important considerations in short and long term success of hip prostheses. Edge loading due to separation in the absence of impingement, which may occur under standard and frequently used activities such as walking as a result of variations on cup positions, is clearly extremely important. Work in the last decade has identified conditions which cause edge loading and investigated the effect of a fixed and predetermined level of edge loading (under fixed biomechanical conditions) on wear and torque. This study for the first time investigated the effect of both rotational and translational positioning on the resultant adverse biomechanics, level of separation and severity of edge loading. These results can be combined with experimental studies to predict wear and the loosening torque on the cup. The study reinforces the importance of translational positioning alongside rotational positioning and the surgical precision needed for both parameters to reduce the severity of edge loading and resulting adverse consequences. The methods and results can be used in the design of new prostheses, in guidance on the level of surgical precision required in the use of available surgical offset adjustments and to define the need and precision for additional surgical positioning of assistive technology.

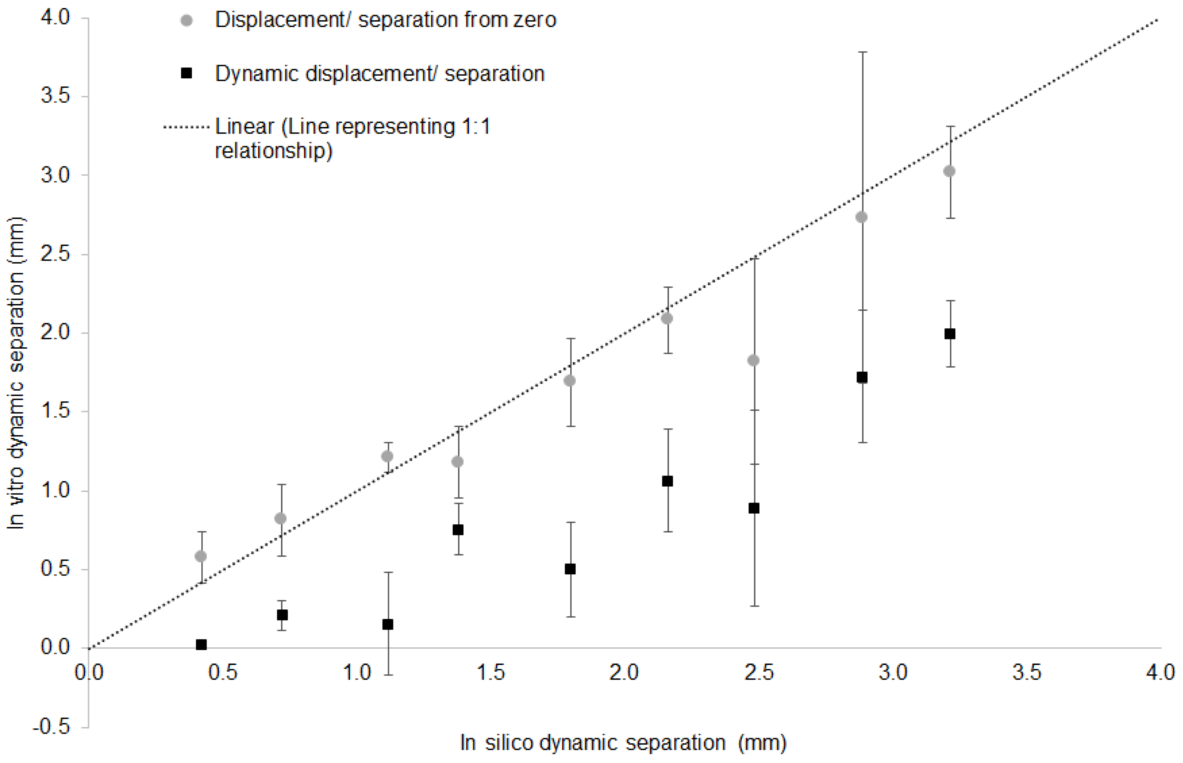


Figure 6: In vitro dynamic separation[28] verses in silico dynamic separation obtained in this study. The dotted line indicates a one to one relationship. The circled marker represent the displacement of the centre of the cup from zero position while the squared markers represent the dynamic separation during the cycle where the cup does not return to absolute zero position[28].

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Disclosure

J. Fisher is a paid consultant to DePuy Synthes Joint Reconstruction, Invibio, Tissue Regenix Group plc, Simulation Solutions and a shareholder of Tissue Regenix Group plc.

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