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Research Article

Finite Element Investigation of the Effects of the Cup Orientation on Implant Load-bearing in Total Hip Resurfacing

Amin Komeili[1]*, BehzadVafaeian[2], SamerAdeeb[2], Ravi Gudena[3], James Powell[4] and Marwan El-Rich[2,5]

[1]*Faculty of Kinesiology, University of Calgary, Canada*

[2]*Civil and Environmental Engineering, University of Alberta, Canada*

[3]*Whiston Hospital, UK*

[4]*Orthopaedic Trauma and Lower Extremity Reconstruction, University of Calgary, Canada*

[5]*Department of Mechanical Engineering, Khalifa University, United Arab Emirates*

***Corresponding Author**: Amin Komeili, Postdoctoral fellowship, University of Calgary, University Drive NW, Calgary, Alberta, Canada.

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Abstract

Metal-On-Metal hip resurfacing technique is becoming widely accepted as a viable alternative to other forms of total hip replacements. The technique has many proven benefits in terms of ease of the surgical operation, operation outcome and subsequent revisions. However, recent reports suggest that there is an effect caused by the different femoral implant and acetabular cup relative orientations on the release of debris into the blood stream. These reports suggest that an "optimum" orientation needs to be sought. The search for an "optimum" position to reduce wear of the metallic components has been explored clinically and experimentally, however, there are no reports to explain the mechanical mechanisms behind an "optimum" position or lack thereof. In this paper, a finite element analysis (FEA) was conducted to investigate the effect of the acetabular cup orientation on the implant behavior under weight-bearing load. The FEA model incorporates all the components of the Birmingham Hip Resurfacing implant along with the bone tissue. A wide range of anteversion and inclination angles were investigated and the effect of these orientations on the contact and von Mises stresses in the implant for a one-legged stance position was studied. The results of this study suggest that, in the one-legged stance loading configuration, the optimum ranges for inclination and anteversion angles associated with the least contact stresses between the femoral component and cup of the implant are 5^0 to15^0 and 40^0 to 45^0 respectively. These findings agree with the reported clinical and experimental results and add insight on the mechanical drivers behind the optimum orientation.

Keywords: *Finite Element Analysis; Metal-on-Metal Hip Resurfacing; Birmingham Hip Resurfacing; Anteversion Angles; Inclination Angles; Contact Stress*

Abbreviations

THA: Total Hip Arthroplasty; MOM: Metal-On-Metal; IGS: Initial Graphics Exchange Specification; BHR: Birmingham Hip; MCP: Maximum Contact Pressure; GPa: Giga Pascal, FEA: Finite Element Analysis

Introduction

An alternative to Total Hip Arthroplasty (THA) is the Total Hip Resurfacing which has been reported as a more effective and beneficial method especially in younger patients [1,2]. The operation is characterized by being conservative thereby retaining more femoral head and acetabulum native bone tissue [3,4]. The conservative nature of the initial operation facilitates any required revisions [3,4]. The possible benefits of hip resurfacing are employing larger bearing surfaces identical to the natural geometry of the hip, accurate restoration of leg length and femoral offset that lead to better stability in addition to allowing more range of movement with reduced risk of dislocation and impingement [4-6]. The short-term outcome of hip resurfacing depends on different parameters among them are patients' age and activities, the surgical technique, and the implant design [7,8]. However, the main factor restricting a long lifespan of hip resurfacing is the wear of the bearing surfaces [9,10]. Wearing of the bearing surfaces is a major concern when resultant debris with high concentration are produced. Osteolysis caused by cellular reactions to the polyethylene debris lead to implant loosening in metal-on-polyethylene resurfacing techniques [11]. Recently, Metal-On-Metal (MOM) hip resurfacing has emerged as a stronger and tougher replacement for polyethylene devices. However, allergic responses, osteolysis and the formation of soft-tissue masses are adverse consequences that have been reported for MOM hip resurfacing [10-17]. These biological responses are due to metal ion concentration arising from metal worn particles. Debris concentration is related to the wear rate of the bearing surfaces [18] and the wearing phenomenon depends on the implant characteristics (geometry, material properties, orientation), lubrication conditions and the contact pressure between contact surfaces [18-21].

The orientation of the resurfaced acetabular cup in MOM hip resurfacing is an important factor in the wear of the bearing surfaces. The acetabular cup orientation affects bearing surfaces lubrication, risk of rim (edge) contact, contact stress and contact area [18,22,23]. Since the cup orientation has a wide range and is practically under the operating surgeons' control [17], the subsequent effects of the different orientations of the cup have become a deserving subject of studies.

The effect of implant orientations on bearing surfaces wear has been studied clinically in order to examine the relation of the wear of the bearing surfaces and the metal ions concentration. Langton., *et al.* [18] measured the metal ions levels in the blood samples of a group of patients having had MOM total hip resurfacing. They found a nonlinear relationship between the chromium ion concentration and the acetabular cup orientation. The minimum ion concentration, indicating lower wear, was associated with the cup inclination and anteversion angles of 45^0 and 20^0 respectively. Hart., *et al.* [24] found that higher metal ions concentrations were associated with inclination angles more than 50^0 in the Birmingham MOM hip resurfacing technique. These clinical studies suggest that a mechanism leading to increased wear occurs at higher inclination angles.

The effect of the inclination angle of the cup in MOM hip resurfacing on the behavior of the implant has also been investigated experimentally and numerically. Williams., *et al.* [16] used dry sawbones to investigate impingement between the MOM hip resurfacing components. They found that impingement occurs at a variety of combinations of inclination and anteversion angles. The range of motion increased with the increase in the ratio of the femoral component diameter to the femoral neck diameter [16]. Angadji., *et al.* [22] investigated experimentally the effect of cup inclination angle on the wear performance of MOM hip resurfacing. Using a hip joint simulator, they found that larger acetabular inclination angle (more than 50^0) corresponded to higher wear. The experimental studies agree with the clinical ones in realizing that higher inclination angles produce higher wear rates for the MOM hip resurfacing technique.

Clinical methods are not able to measure mechanical variables such as stresses while the experimental studies lack comprehensiveness due to their inherent limitations and costs. An alternative to those is the use of numerical methods such as Finite Element Analysis (FEA) to understand the mechanical behavior of the implant in MOM hip resurfacing technique. This study aims to evaluate the effect of the cup orientation on the implant load-bearing in hip resurfacing using FEA. In particular, the contact and von Mises stresses are evaluated under gravity and muscle loads for several inclination and anteversion angles in the one-legged stance loading configuration.

Materials and Methods

A 3D finite element model of a semi pelvic bone -portion between the sacroiliac joint and the symphysis pubis- and the related right femur bone along with a typically MOM hip-resurfacing prosthesis was generated. Modeling the pelvis and femoral bone was based on an anonymous CT-Scan data of a hip joint.

Geometry Construction

The geometry of the bony structures was extracted from the CT-Scan images using Mimics software [25]. In order to obtain properly smoothed surfaces, the binary STL file generated by Mimics was imported into Geomagic software [26]. Non-uniform rational b-spline based surfaces produced by Geomagic were exported to the Initial Graphics Exchange Specification (IGS) graphics format. Since the geometry of the hip structure was assumed to be completely symmetric about the sagittal plane, half of it was considered (Figure 1). This geometrical reduction would facilitate the finite element modeling and reduce the calculation time by reducing the required digital memory allocation. The final geometry of the bony parts consisted of half of the right pelvis, related femoral head, and neck.

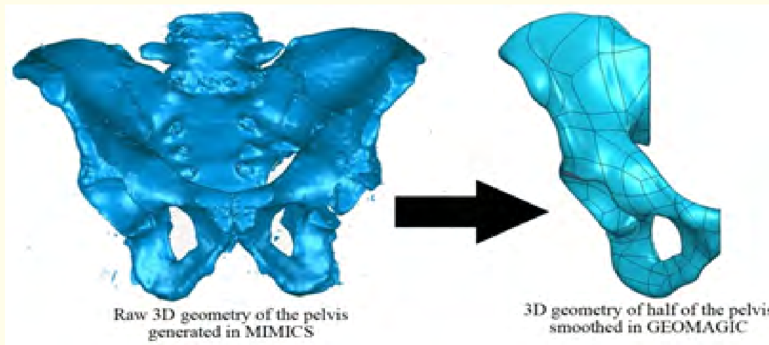

Figure 1: *Schematic of 3D geometry of the pelvis constructed in Mimics and smoothed in Geomagic.*

Recognized as one of the most widely-used hip resurfacing arthroplasty, the Birmingham hip (BHR) joint implant (Smith & Nephew Orthopaedics Ltd, Bromsgrove, United Kingdom) [27-29] was chosen for this study. The prosthesis geometry was extracted from the implant size chart proposed by the manufacturer. The cup had a nominal outer diameter of 60 mm and a uniform thickness of 3 mm. A 54 mm femoral head component was used with a radial clearance of 8 μm between the bearing surfaces. The implant geometry was reconstructed using SolidWorks software [30].

Assembly

Femoral head and implant

Accurate assembly and positioning of the different components is crucial for obtaining results that are clinically relevant. Particularly, position of the femoral component is an important pre-operative consideration as mal-positioning of the femoral head may lead to high stress and consequently femoral neck fracture [31-33]. The bony structures and the implant components were assembled and positioned using SolidWorks which provided appropriate tools to perform a facilitated and precise assembly procedure. As previous studies showed that by increasing the angle between femoral shaft (the center line of the femoral head) and the implant stem the strains in the superior femoral neck could be reduced and may help to reduce the risk of fracture [34,35], thus, the femoral component stem was aligned with the femoral shaft with 5^0 valgus angle (Figure 2a) [36]. The geometry of the femoral head was modified to simulate the cylindrical and chamfer reaming performed during the surgical operation. Since single leg stance position that is a stage of the normal gait was considered for analysis, a 5^0 lateral angle between the center line of the femoral head and coronal plane (Figure 2b) was modeled.

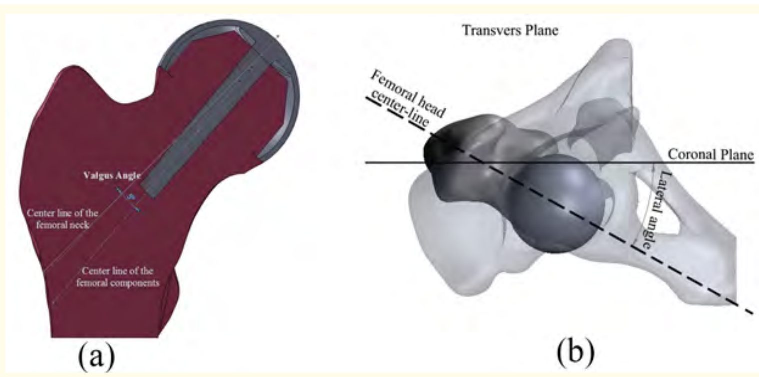

Figure 2: *Positioning of femur and implant (a) Varus angle between implant and femur, (b) Femoral head lateral angle.*

Acetabulum and Cup

The reaming of the acetabulum that is carried out during the hip resurfacing operation was simulated by removing the appropriate parts of the acetabulum cavity geometry. The position of the cup was based on Murray's radiographic definition of orientations [37] in which the acetabular inclination angle is defined as the angle between the face of the cup and the transverse plane (Figure 3a). The cup anteversion angle is defined as the angle between the cup axis and the coronal plane (Figure 3b). 30^0, 35^0, 40^0, 45^0, 50^0 and 55^0 of inclination angles combined with 0^0, 5^0, 10^0, 15^0, 20^0, 25^0 and 30^0 anteversion angles were considered with a total of 42 finite element models.

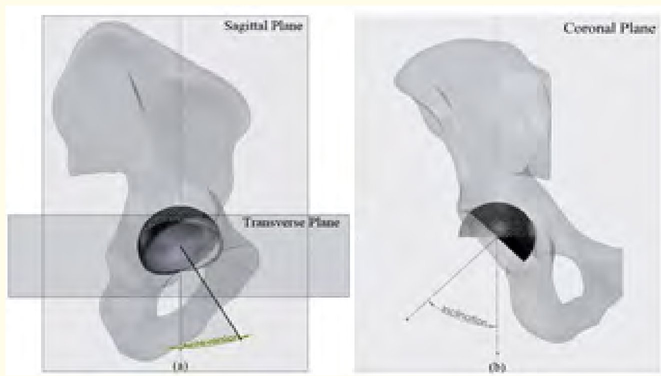

Figure 3: *Cup orientation based on Murray's radiographic definitions (Murray): (a) Anteversion angle (b) Inclination angle.*

Mesh Generation

Prepared geometries were exported from SolidWorks as IGS files and imported into Hypermesh software (Altair Engineering, Troy, USA) [38] where the finite element mesh was generated for the fully assembled model. The bony parts of the model (pelvis and femoral part) consisted of a cortical bone layer surrounding cancellous bone. The cortical bone layer was considered to have a constant thickness of 1.5 mm [39,40] and was represented by triangular shell elements. Meshing the cortical layer by shell elements provided a mesh pattern on the outer surface. This pattern was utilized for meshing the cancellous part using tetrahedral solid elements.

Two different types of elements were used to generate the mesh of the prosthesis parts. The femoral component was meshed using tetrahedral solid elements while the cup was meshed with linear brick elements to improve the performance of contact analysis. The cement that is used to fill the gap between the reamed femur head and the inner surface of the femoral component was meshed using tetrahedral elements. A denser mesh was employed on the acetabulum cavity and the prosthesis components to accurately capture the curved geometry and model the contact between the components. The final model consisted of 42126 linear shell elements and 416980 linear solid elements (Figure 4). All materials were assumed to be homogenous, isotropic and linear elastic [41-45]. The material properties and the number of elements employed for each part are summarized in Table 1.

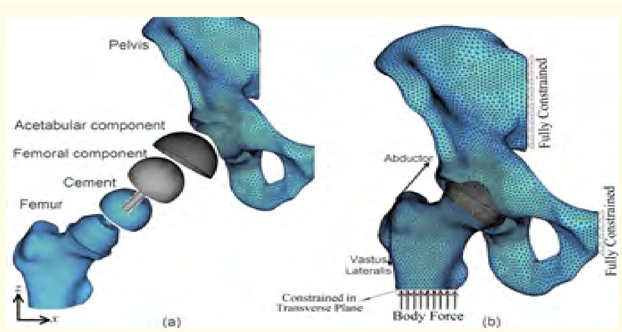

Figure 4: *(a) Hip joint with prosthesis components and cement. (b) 3: D FE model of the hip joint with loading and boundary conditions.*

Components	Element Type	No. of Elements	Material Properties (Shultz., *et al.*)	
			Young Modulus E (GPa)	Poisson Ratio ν
Cortical pelvis	Shell	17574	17	0.3
Cortical femur	Shell	24552	17	0.3
Cancellous pelvis	Solid	139212	0.8	0.2
Cancellous femur	Solid	73737	0.8	0.2
Cup	Solid	17968	210	0.3
Femoral component	Solid	158911	210	0.3
Cement	Solid	27152	7	0.3

Table 1: *Material properties of prosthesis components and non-metallic materials with associated element types.*

Boundary Conditions

As mentioned earlier single-leg stance was the chosen body stature for this analysis. Since it was assumed that the symmetrical structure of the hip remained un-rotated in any planes by that position, half of the pelvis was modeled and fixed boundary conditions at the sacroiliac joint and symphysis pubis joint were applied. The degrees of freedom at the distal end of the femur neck were constrained in the transverse plane.

The loading scenario consisted of the body weight force and the muscle forces involved in the single-leg stance of normal gate [44,46-49]. The magnitude of the ground reaction force applied vertically at the distal end of the femoral part was equal to a total body weight of 80kg. Since this study was focused on obtaining the structural behavior of the implanted hip joint, the attachment details of the muscles distant from the joint were simplified; the muscle forces were applied in the direction of their lines of action as concentrated forces on limited number of nodes representing their attachment and insertion points. Two prominent muscle groups (Abductor and Vastus Lateralis) of the hip joint acting in the single-leg stance phase of gate were considered [46] (Table 2).

Load Case	Type of Force	Force Components* (Percentage of Body Weight)			Magnitude
		F_x	F_y	F_z	F
Single-leg Stance	Abductor	58.0	4.3	86.5	104.23
	Vastus lateralis	0.9	18.5	92.9	94.73
	Ground reaction	0	0	100	100

Table 2: *Applied forces, expressed in percentage of body weight, corresponding to single-leg stance posture (Bergmann., et al) strength and fixation stability of hip implants. Measurements of hip contact forces with instrumented implants and synchronous analyses of gait patterns and ground reaction forces were performed in four patients during the most frequent activities of daily living. From the individual data sets an average was calculated. The paper focuses on the loading of the femoral implant component but complete data are additionally stored on an associated compact disc. It contains complete gait and hip contact force data as well as calculated muscle activities during walking and stair climbing and the frequencies of daily activities observed in hip patients. The mechanical loading and function of the hip joint and proximal femur is thereby completely documented. The average patient loaded his hip joint with 238% BW (percent of body weight.*

**expressed with respect to the global reference X, Y, Z corresponding to the medial-lateral, the anterior-posterior and the proximal-distal directions, respectively.*

Contact Properties

The femoral component and head, and the cement were meshed together in order to generate common boundary nodes, which made them fully bonded. The mesh for the cup and the pelvis acetabular cavity, however, was generated separately, therefore, the nodes on the outer surface of the cup were tied to the inner surface of the acetabular cavity.

The cup and femoral component were modeled as surfaces sliding on each other with an adequate coefficient of friction [20,50]. The FEA solver ABAQUS (Dassault Systèmes SolidWorks Corp.) [51] was employed to conduct a proper contact analysis. The finite-sliding formulation (allowing separation after contact) was chosen to allow the implant surfaces to slide freely relatively to each other. Contact simulation was done using the "Surface-to-Surface Contact Interference" option. The contact properties of the surfaces consisted of normal and tangential behavior. In the case of normal contact, an exponential function was imposed as the pressure-overclosure relationship to avoid pseudo locally concentrated contact forces. The exponential function with a softened contact was utilized to minimize the clearance between the contact surfaces pair and prevent the over-closure of contact surfaces as well. The penalty method was utilized to manage the tangential contact behavior. A coefficient of friction of 0.006 was used, simulating full joint mixed lubrication [21]. The cup was partitioned into 16 regions to facilitate the addressing of the location of maximum contact pressures in different models (Figure 5).

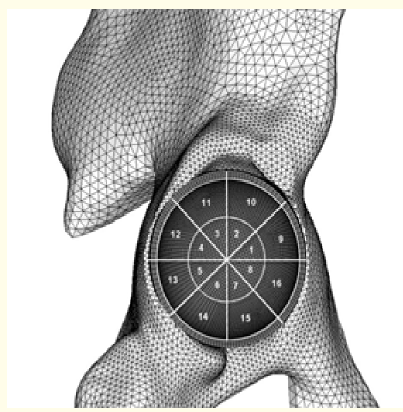

Figure 5: *The 16 defined regions of Cup.*

Results and Discussion

The contour plots of the predicted contact pressure distribution on the cup inner surface for different cup orientations (30°, 35°, 40°, 45°, 50° and 55° of inclination and 0°, 5°, 10°, 15°, 20°, 25°, 30° of anteversion) are shown in Figure 6. The finite element analysis indicates that the maximum contact pressure (MCP) on the cup inner surface ranges between 2.12 to 3.00 MPa at different cup orientations. The highest MCP occurs in the case of inclination 50^0 and anteversion 25^0 and the lowest MCP appears when the cup has 35^0 of inclination and 0^0 of anteversion. To reveal the variations of the MCP, its values as a function of the anteversion and inclination angles are presented in Figure 7a. As a general trend, the MCP increases when the inclination or/and anteversion angle(s) rises. However, in some cases fluctuations are apparent; increasing the cup inclination angle from 30^0 to 55^0 is associated with increasing in MCP values, except for the anteversion 20^0 and 30^0 at which increasing the inclination does not lead to higher MCP. In addition, corresponding to a constant inclination angle the MCP decreases locally at anteversion 10° (except for 50^0 of inclination). Anteversion angles lower than 10^0 have relatively insignificant influence on the MCP values.

The variation trend of the MCP can be also investigated on the femoral component surface. The same MCP trends as that of the cup are anticipated. Figure 7b shows the MCP values on the femoral surface which are almost the same as that of the cup.

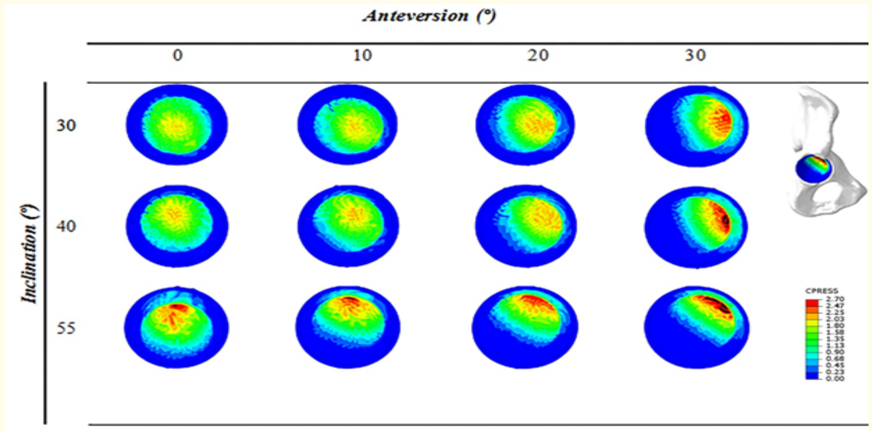

Figure 6: *Contour plot comparison of the predicted contact pressure distribution (In MPa) for different orientation of the cup.*

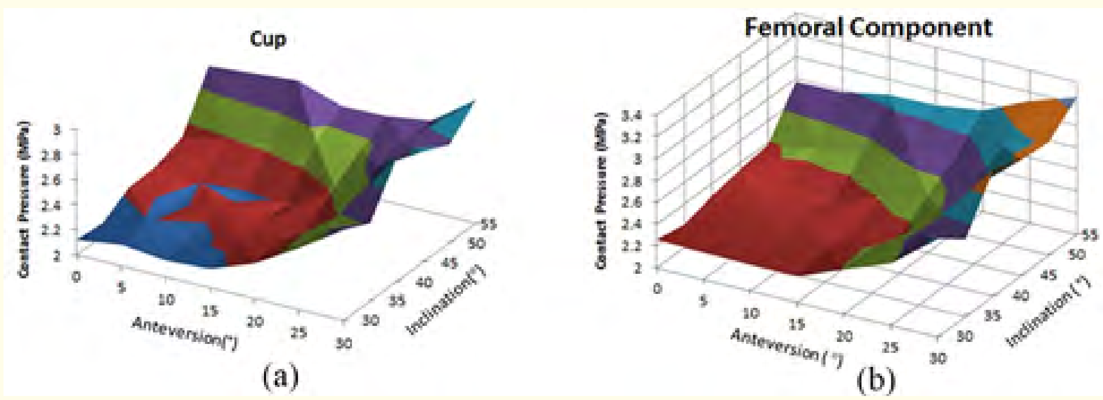

Figure 7: *Maximum contact pressure recorded on (a) cup (b) femoral component.*

In addition to the effect of the cup orientation on the MCP values, the FEA results also indicate the location of the MCP on the cup surface. For anteversion angles less than 25° and inclination angles less than 45°, the MCP is located on the regions 1 and 2 (Figure 5), closer to the center of the cup. However, for anteversion angles higher than 25° and inclination angles higher than 45° the location of the MCP moves towards the edge of the cup.

The contact region between the two implant components changes in area and location. Figure 8 depicts the change in the normalized contact area A/A_0 (where A is the area of the contact region and A_0is the total area of the cup inner surface) as a function of cup orientations. The maximum of the normalized contact area (A/A_0) for each inclination angle occurs when the anteversion angle is in the range of 5^0 to 15^0. Almost for all inclination angles, the ratio A/A0 decreases once the cup orientation reaches 10^0 of anteversion. The location of the contact region moves when the cup picks different orientations. For each inclination angle, increasing the anteversion angle shifts the contact area to the cup rim (Figure 6).

Figure 9 displays the effect of cup orientations on the von Mises stress in the cup. The range of von Mises stresses extends from 30MPa to 120 MPa in the cup, but it is fairly constant in the femoral component (27 ± 0.6 MPa). The magnitude of the maximum von Mises stress in the cup is moderately sensitive to both the inclination and anteversion angles but its location mainly depends on anteversion angles, however maximum von Mises stresses in the femoral component is not affected by changing the inclination and anteversion angles. Figure 10 shows that inclination angles of higher than 45° and anteversion angles of higher than 25° cause the maximum von Mises stress to place in the cup rim.

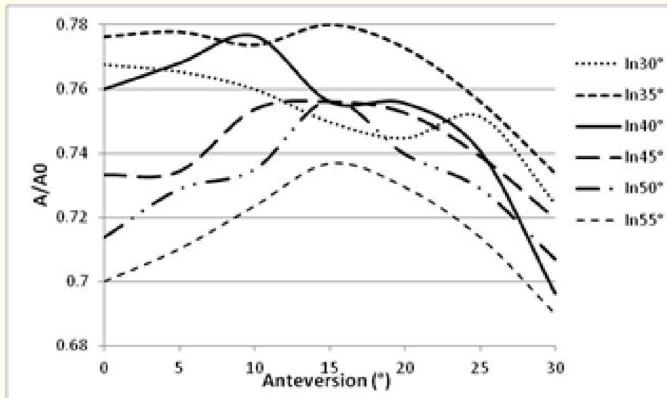

Figure 8: *Ratio of the contact area A to the cup area A_0 at different anteversion and inclination angles.*

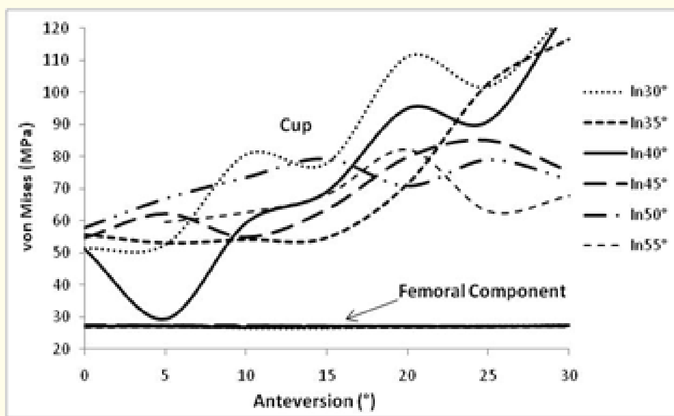

Figure 9: *Maximum von Mises stresses in different cup orientations, (a) in the cup, (b) in the femoral component.*

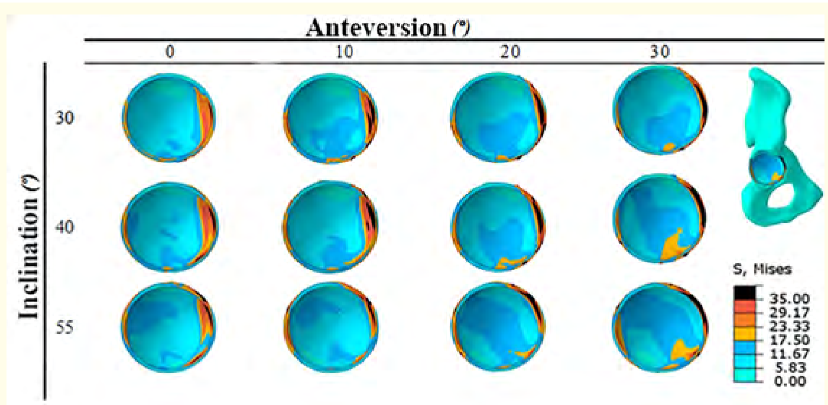

Figure 10: *Distribution of the von Mises stress (MPa) in the cup for different cup orientations.*

The aim of the present work is to investigate the effect of the cup orientations on mechanical stresses in implants utilized for Total Hip Resurfacing. Acquiring the contact pressure between the implant surfaces and von Mises stress in the implant components is the main result of this study. Complex models of a hip joint with its embedded implants under static loads (related to the single-leg stance stage of the normal gate) were analyzed using FEA. The models were identical in material and geometry but differed in the cup orientations. Varying the cup orientation under identical loadings of the system affected the contact pressure between the implant surfaces. It was expected to observe exactly the same contact pressure values on the both surfaces of the implant. However, small differences (about 5%) in contact pressure magnitude and its distribution were observed on the cup and the femoral component surfaces. These negligibly small differences (for the utilized contact analysis method) were due to the relative size of the finite element mesh which led to non-correspondent nodes on the master and slave surfaces.

The MCP is investigated with considering either the cup or the femoral component surface. The values of the MCP may seem unrealistically small if compared to other researchers' results [50,52]. However, considering these two factors clarifies the issue: 1) the body weight and the muscle forces acting on the system are assumed to be of an 80kg-individual; unlike some literatures in which the forces are assumed to be several times greater than that of an individual body weight. 2) Radial clearance is another effective factor on contact pressure magnitudes. Cilingir., *et al*. [53] showed the effect of radial clearance of the contact pressure for a ceramic-on-ceramic hip resurfacing implant. His study implied a nearly linear relationship between the MCP and radial clearance values for a definite cup orientation; such that the MCP values are relatively small for small radial distances. The small radial clearance of 0.075 mm which was used in this study can be another explanation of the order of the MCP values when compared to other studies.

The FEA results regarding to MCP indicate a general increasing trend in the values of MCP as inclination or anteversion angles increases. This result is in agreement with other computational results (Khan, Kuiper, and Richardson). According to our results, changing the cup orientations at most led to a 42% change in the MCP (from 2.12 to 3 MPa). This amount of difference in contact pressure can be a potent cause of wear rate when acting along with other factors. Williams., *et al*. [16] experimentally studied the wear of a 56 mm cup in a metal-on-metal total hip resurfacing. They showed that changing the cup inclination angel from 45° to 55° (with a constant anteversion of 0°) could result in a 60% increase in initial wear rate (cumulative wear per normal gait cycle). Our results show a 22% increase in MCP for the same cup orientations. The results on the contact pressure can also be related to clinical studies. Clinical studies are mainly focused on the level of metal ions (chromium and cobalt) released into blood by virtue of the wear occurred in the implant surfaces. Chromium and cobalt levels are reported to generally increase as the inclination angel of the cup increases [18,54]. Inclination angel greater than 55° was shown to result in higher chromium level compared to smaller inclination angel [18]. Figure 6 shows a general increase in MCP with the increase of inclination angel (with constant anteversion). It cannot be strongly inferred that the greater the MCP, the higher the metal ion level is, since the ion levels also depend on some other factors [17]. However, the MCP as one of the factors plays its role in increasing the wear rate, and the wear rate directly influences the ion levels. In addition to inclination angels, the effect of anteversion angels on metal ion levels is also studied. D. J. Langton., *et al*. [18] showed that the minimum chromium ion levels (in terms of ion concentration) reached for the range of anteversion angel from 10° to 20°. Our FEA results are also in agreement with that clinical finding. If the MCP values in the range 0° to 5° of anteversion (and constant inclination) are neglected due to the risk of impingement [55], the minimum of the MCP occurs for the anteversion angels between 10° and 20°. It must be noted that a direct relationship between the MCP and the metal ion levels may not be easily established due to the following reasons: 1) the MCP non-linearly depends on the cup orientations for a specific loading scenario, 2) the ion levels and the cup orientations was found to have a complex and non-linear relationship for a particular metal-on-metal implant [18].

Von Mises stress was obtained from the finite element analyses regarding to a specific loading scenario; namely, single-leg stance. Since the implant components are made of metal, the von Mises stress is one of the implant design parameters. Unlike the von Mises stress in the femoral component which had a nearly constant value (27 ± 0.6 MPa), its value in the cup covered a wide range from 30 MPa to 120 MPa for different cup orientations. The gap between the cup rim and the acetabular cavity causes bending moments in the cup rim which raise the von Mises stress in the cup. The complete backing of the femoral component by the cement leads to smaller bending moments comparing to the cup. The maximum values of the von Mises stress are well below the yield stress of the implant, which, according to ASTM standard [56] has to be higher than 450 MPa. It suggests that no major concern exists regarding the implant design if the loading condition is the single-leg stance phase of normal gait.

The safe range of the cup orientation based on the minimum values of the MCP and von Mises stress can be determined out of the analyses results. The coupled range as 0° to 25° of anteversion and 30° to 45° of inclination is recognized to be the safe range based on out FEA results. However, the following three other considerations on the cup optimum orientation are apparent in literatures: the risk of impingement, the possibility of dislocation and the metal ions concentration. This risk of impingement between the femoral head and the pelvic bone is reported for anteversion less than 10° or inclination less than 30° [55]. Dislocation and instability of the implant is associated with anteversion and inclination higher than 25° or 45° respectively [57]. Metal ions concentration is a clinical concern not only depends on mechanical circumstances (wear mechanism and surface lubrication) but also depends on biological variables. It has been reported that the cup inclination angle greater than 50° increases the concentration of cobalt and chromium ions in a MOM hip resurfacing [24,58]. Overall, combining our FEA results with those considerations suggest that the optimum range for anteversion and inclination angles are 10° to 25° and 35° to 45° respectively (Figure 11).

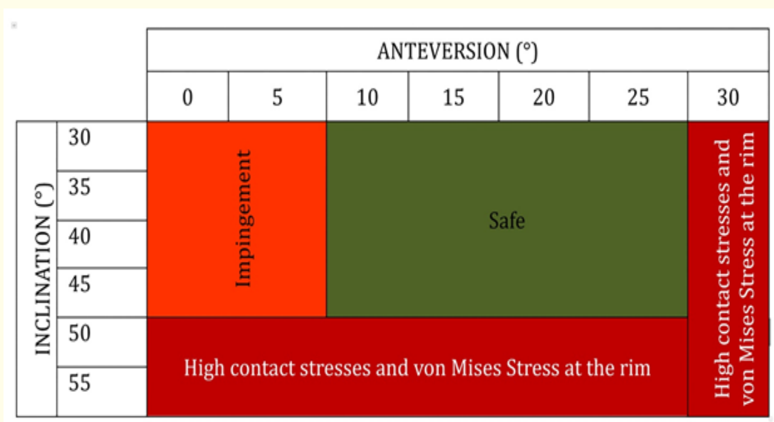

Figure 11: *Risks associated with different inclination and anteversion angles.*

The proposed region is in good agreement with the recommended safe region reported in available clinical studies in which the ions concentration, wear rate, fracture of the bone and bone remodeling were the focus of the study [59,60].

Our results are applicable to the loading scenario modeled; single legged stance, which constitutes a limitation to the generality of the results. In addition, our model was performed on one geometry set obtained from CT scans and does not include effects due to geometric variation in the pelvic bones among patients. The assumed isotropic material properties of the cancellous bone constitute a minor limitation and could have minor effects on the results. Another limitation is the idealized bone-implant interface condition with perfect contact between the cup and periprosthetic bone, assuming that the rimming procedure during the operation is carefully conducted. While the presented model has extensively covered a wide range of anteversion and inclination angles, such numerical model is necessarily limited to the modeled subject's geometry and loading configurations. The optimum orientation of the cup may vary due to other subjects' characteristics such as activities, pelvis shape and age.

Conclusion

Results of this study showed that, as a general trend, the MCP increases when the inclination or/and anteversion angle(s) rises. The inclination angles of higher than 45° and anteversion angles of higher than 25° cause the maximum von Mises stress and MCP to place in the cup rim. Overall, combining our FEA results with those considerations suggest that the optimum range for anteversion and inclination angles are 10° to 25° and 35° to 45° respectively.

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Conflict of Interest

There is no financial interest or any conflict of interest to declare.

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Volume 5 Issue 5 April 2017