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Finite element methods in analyzing contact mechanics to estimate the wear of hard biomaterials in human hip prosthesis - A review

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ABSTRACT

Hip simulator, pin-on-disc (POD) and finite element analysis (FEA) are widely used approaches to estimate in-vitro wear of implants. While former two are shown to be widely employed for predicting wear under lubricated and non-lubricated conditions, which are often not suitable for assessing wear in the event of extensive changes of hip design parameters. The parameters like radial clearance, edge loading, acetabulum cup thickness, femur head diameter and cup inclination angle have impact on wear. This article focuses on computational approaches involved in predicting hip wear of hard bearing biomaterials i.e. metal-on-metal (M-o-M), Ceramic-on-ceramic (C-o-C) and PCD-PCD (poly crystalline diamond) for human hip prosthesis. The limitations on reliability of FEA tool based on various parameters like meshing, modeling approaches and experimentally determined friction and wear coefficients from POD or hip simulator are reviewed. The continuous evolutions of modeling techniques developed using FEA tool are motivating the researchers, to improve these techniques further to predict contact stress and wear of implants in better way.

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1.0 INTRODUCTION

Due to aging, human hip joints are affected by osteolysis which leads to wear of acetabulum cup and femur head bearing surfaces which represents a typical ball-in-socket joint. Despite osteolysis, hip joints might prone to failure due to disease, injury and physically demanding gait activities. Hip prosthesis is to replace the injured and diseased acetabulum cup and femur head using bio-compatible materials, to restore normal functionalities of hip joints (Choroszyński et al., 2017). Normally human hip prosthesis is categorized into two types, namely, total hip replacement (THR) and hip resurfacing (HR). THR involves replacement of femur head, acetabulum cup and femur stem with artificial materials, while the latter, involves only the removal of acetabulum cup and femur head. Earlier, THR surgery was performed for many patients, which showed dislocation because of crack in bone cement leading to loosening of acetabular component (Ring, 1967). Therefore, use of computers in assisting the surgery of hip resurfacing resulted in the reduction in failure of components and had better accuracy in positioning of femoral component (Seyler et al., 2008). The use of cementless hip prosthesis enabled the cup to be secured firmly with the help of threads and screws, thus eliminating the use of bone cement. It also promoted bone growth and helped in attaching the prosthesis to bone surface (Morscher, 1983). The different types of THR techniques (Wiles, 1958) showed that these operational techniques for THR mainly depend on patient's age and gait activity of the person being involved frequently. Crack formation developed in cement mantle due to circumferential stresses was the most reported post operative problem (Stauffer, 1982). In this regard, the idea of small femoral head diameter to reduce the total load exerted on artificial joint was investigated (Charnley, 1970). Research revealed that the percentage of patients required revision surgery were high and most of the dislocation was mainly due to the malposition of acetabulum cup (Ali Khan et al., 1981). The femur head with larger diameter was found as the alternate approach for reduction in dislocation (Berry et al., 2005). Demographically, the rate of dislocation was found to be higher for female patients and remained as a major concern for long term post operative conditions (Berry et al., 2004). Further, older patients are more vulnerable compared to younger patients and dislocation after surgery was mainly due to femoral fracture (Newington et al., 1990).

The various post operative failures of total hip arthroplasty were due to the poor or inadequate cementing technique, femoral head loosening and improper acetabular cup positioning (Callaghan et al., 1985). Another study (Schmalzried et al., 1992) reported that the loosening of the cement was mainly, due to the biological issues of patients' body reaction to the cement material and not due to the mechanical failure. Age related factor had greater influence in cemented total hip arthroplasty technique, and it had to be carefully addressed (Dorr et al., 1994). Loosening and inflammation ranging from, mild to severe were the common issues, encountered in THR (Charosky et al., 1973).

An alternative technique called hip resurfacing was used instead of THR, because the failure rate of THR was found be more, compared with former (Langton et al., 2011). The retrieved failed M-o-M hip prostheses result showed that the edge loading was the common factor affecting the wear of the bearings in hip resurfacing as well as THR (Matthies et al., 2011). Some of the advantages for hip resurfacing include preservation of bone, effective stress transfer, less dislocation and effective revision of implants when compared with THR (Mont et al., 2006). The factors affecting the selection of patient for hip resurfacing includes gait activity, leg length, gender etc., (Nunley et al., 2009). Hip resurfacing technique was widely used in younger patients and the possible difficulties encountered in hip resurfacing include the effect of release of metal ions and complex procedures (Quesada et al., 2008). Similar kind of observation was reported (Treacy et al., 2005) wherein, the hip resurfacing involving (M-oM) was found to be more effective for younger patients since no revision occurred for minimum follow up of five years. The various factors affecting the design of implants for hip resurfacing like range of motion and dislocation needs to be analyzed for younger patients since the range of motion of femur head inside the acetabulum component was limited (Kluess et al., 2008). The use of smaller radial clearance between the acetabulum cup and femoral head had significant effect, in the reduction in level of metal ions found in the blood (Langton et al., 2009). The femur head diameter and age of the patients were the major factors causing the failure of hip resurfacing arthroplasty (Corten and MacDonald, 2010). Another factor causing the failure of hip resurfacing technique was the use of steep inclination angle for acetabulum cup which resulted in increase in level of metal ions in blood and it led to edge loading (De Haan et al., 2008).

All the above-mentioned problems lead to wear of biomaterials and it remains the major concern for surgeons and academicians, to look for new highly biocompatible materials to reduce the consequences of wear. The wear was mainly influenced by hip joint contact force, rotation of femur head inside the cup due to different gait activities, implant mal-positioning, steep cup inclination and micro separation leading to edge loading (Bergmann et al., 2001; Charnley and Halley, 1975; Schmalzried and Callaghan, 1999; Schmalzried et al., 1998; Varady et al., 2015). Wear particles generated from implants lead to tissue inflammation and revision of implants.

There are three approaches which are mainly used in analyzing the wear of hip implants namely, clinical measurement (in-vivo), in vitro simulator test and computational wear analysis. The first approach involves patients' consent in estimating the wear as it requires long term follow-up. The techniques adopted include radiographic, computerized tomography and hip analysis suite. The risk of dislocation in terms of inclination and anteversion of acetabular component leading to wear could be identified with the help of above techniques (Rieker and Köttig, 2002; Sadhu et al., 2017). Without active participation of patients, these approaches could not be used to diagnose ailment or to estimate wear to improve the success rate of hip surgery. Sometimes follow-up of patients would be lost and willingness to participate also would be a problem. The second approach involves using equipment like hip simulator, POD, ball-on-disc (BOD) and circularly translating Pin-on-Disc (CTPOD) in estimating wear of biomaterials under dry and lubrication conditions (Capitanu et al., 2019; Leslie et al., 2008; Rao et al., 2018; Razak et al., 2016; Razak et al., 2015; Saikko and Keränen, 2002). In-order to estimate the wear of such biomaterials using above mentioned equipments, there are many factors like loading similar to normal walking, effect of bio-lubricants as similar to synovial fluid and test cycles or duration needs to be determined. This requires extensive design modifications leading to high cost in designing of equipments as well as cost spent for biomaterials in manufacturing femur head and acetabulum cup in order to estimate wear.

The problems or difficulties faced in above two approaches could be easily addressed with the finite element wear approach. Using this approach, modeling of implants with respect to different parameters could be investigated in better way. Also cost incurred in estimating wear using above two approaches could be greatly reduced. Moreover, suitable design parameters in designing the hip implants to improve the success rate of hip surgery are possible in this approach. There are two modeling techniques used in FEA i.e., global and local model to estimate the contact pressure, von Mises stress etc which are discussed in detail in FE modeling section. In order to estimate the wear, two wear laws namely, Archard's wear law and dissipated energy wear law are used in analyzing the wear of the implants based on obtained contact pressure using finite element concepts. The wear law is discussed in detail in section 2.5 for calculating linear, volumetric, fretting and fatigue wear.

This review article consolidates the different FEA techniques adopted in analyzing the contact pressure and wear of human hip prosthesis for hard-on- hard biocompatible materials used as replacement for hip joints. The modeling techniques, input parameters to FE modeling, mesh convergence and design parameters. The process flow chart of FEA wear analysis which includes linear, volumetric, fretting and fatigue wear of hard bearing materials is clearly depicted in Figure 1. First, the contact mechanics of hip model using global and local model approach are used to generate contact pressure obtained from FEA packages including ANSYS and ABAQUS are reviewed in this study. Later wear of the bearings analyzed using different wear law namely Archard's wear, modified Archard’s wear and energy wear law approaches. The wear studies based on the hip joint design parameters which include femur head diameter, radial clearance, acetabulum cup thickness, physically demanding gait activities, cup inclination angle, microseparation with and without frictional contact were shown in Figure 2. So far, two review articles focused on wear of human hip prosthesis which dealt with lubrication wear modeling and polyethylene wear based on FEA were published (Mattei et al., 2011; Lin Wang et al., 2019). One of the recent review articles focused on tribological study of Si3N4 based on biomedical applications which dealt with in-vitro and FEA studies (Subramaniam et al., 2021).

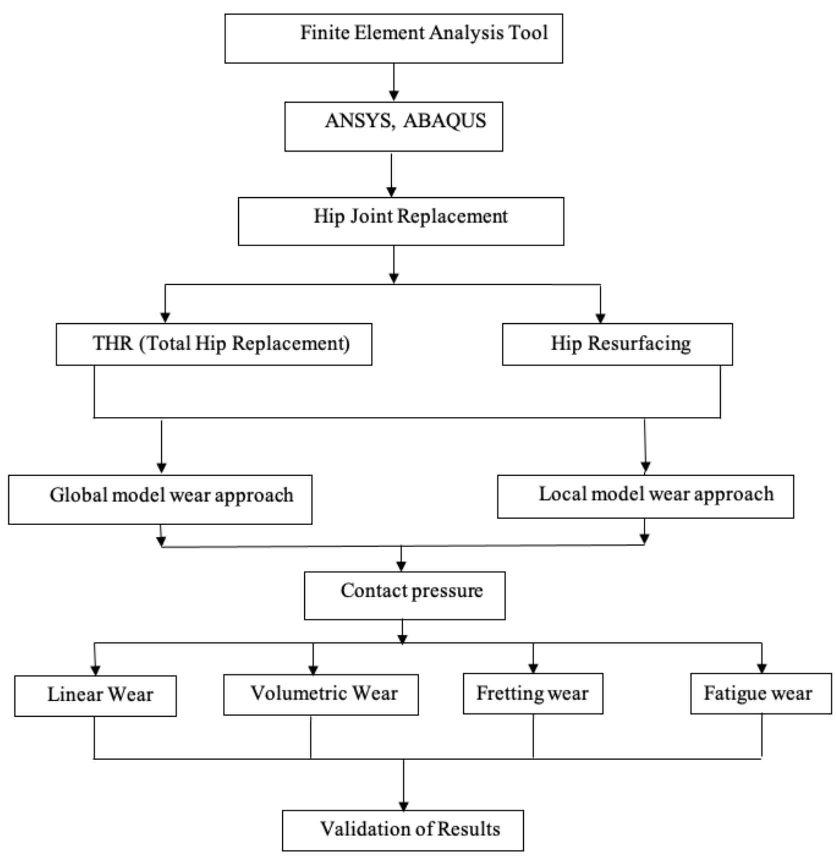

Figure 1: Overview of finite element wear analysis.

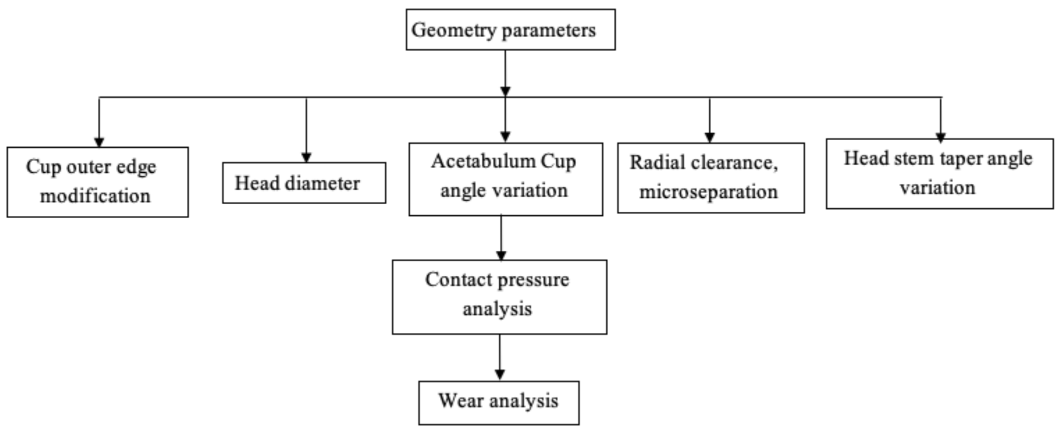

Figure 2: Parameters considered for wear modeling of hip prosthesis.

2.0 MODELING AND WEAR STUDY OF HUMAN HIP JOINT

Finite element modeling in analyzing the wear of hip implants involves different parameters like acetabulum cup thickness, head diameter of implants, cup inclination angles, radial clearance and micro-separation or head lateral displacement. The acetabulum cup thickness represents outer radius of acetabulum cup as it helps in securing joint firmly to pelvic bone. Head diameter refers to femur head typically ranging from 22 to 32 mm. Cup inclination refers to cup abduction/ante version angles based on articulation side and coordinate axis. Radial clearance refers to the clearance between the acetabulum cup and femur head for effective lubrication of bearing surfaces. The head lateral displacement or dislocation during post operative period with respect to acetabulum cup leading to edge contact refers to microseparation. These parameters were difficult to be analyzed in hip simulator or POD as it requires major design modifications to be incorporated in these equipments leading to high cost and time consumption. Initially the contact stress, von Mises stress and deformation were analyzed using FEA and later on wear of the bearings was investigated mostly using Archard's wear law and dissipated energy wear law. The former wear law was widely adopted for ball-in-socket model for volumetric wear analysis while the latter was applied for head-stem taper junction in analyzing fretting wear or fatigue wear. The Archard's wear law relies on volumetric wear of contact between two rubbing surfaces for applied normal load, wear coefficient and sliding distance of rubbing surfaces. It does not account for cross shear and relative sliding motion between two surfaces. In case of dissipated energy wear law, using energy wear coefficient (α) by considering interfacial shear and relative motion the volumetric wear is computed (Ashkanfar et al., 2017). The modeling software's like ANSYS and ABAQUS were widely used in analyzing the contact pressure of the bearings by applying loading and boundary conditions to the developed FE model. The various mechanical properties of the bearings are listed in Table 1 which mainly include Young's modulus, Poisson ratio and friction coefficient. These values were given as input to FE model based on material combination analyzed for estimating the contact pressure. The hip gait loads reported in the literatures (Bergmann et al., 2001; Varady et al., 2015) were mainly used as input parameters in analyzing the wear of the implants.

Table 1: Mechanical properties of biomaterials.

#	Bio-material	Mechanical properties				Reference
		E (GPa)	υ	COF	K_w (mm^3/Nm)	
1.	CoCrMo/CoCrMo	210	0.3	0.2	0.5×10^{-8} 0.15×10^{-8}	Harun et al., 2009
2.	Al_2O_3/Al_2O_3	375	0.3	0.1	0.2×10^{-8}	Shankar, 2014
3.	Si_3N_4/ Si_3N_4	300	0.29	0.2	0.43×10^{-8}	Shankar, 2014
4.	PCD/PCD	900	0.1	0.1	0.00459×10^{-8}	Uddin and Zhang, 2013

2.1 Global and Local Modeling

The global modeling (GM) represents the modeling of entire geometry of acetabulum cup, femur head and stem. Also in GM, non-uniformity in meshing of FE model was widely adopted due to high computation time which leads to the less accurate estimation of contact pressure. Earlier studies considered modeling of cement mantle and stem support along with femur head and acetabulum cup in estimating the wear of hard bearing materials (Liu et al., 2008). One of the studies further simplified the model and showed that pelvic bone had negligible effect on contact pressure (Barreto et al., 2010). So, most of the recent literatures used simplified acetabulum cup and half of femur head in estimating contact pressure (Nithyaprakash et al., 2018; Shankar and Nithyaprakash, 2014; Shankar et al., 2020; Uddin and Chan, 2019; Uddin and Zhang, 2013). For this, the radius of femur head was taken as 14 mm with radial clearance of 0.05 mm. The acetabulum cup thickness was considered to be 5 mm. Although most of the literatures adopted above mentioned parameters as standard dimensions for many FEA study, head diameter depending upon patients i.e. it ranges from 18 to 34 mm (Berry et al., 2005). Similarly for radial clearance also, it ranges from 0.001 to 1 mm (Wang et al., 2019). Sometimes, microseparation between the bearing alone was considered to estimate the contact pressure developed during edge loading (Uddin and Chan, 2018). Ellipsoidal bearing, instead of sharp corner, round corner geometries were modeled using FEA to investigate the contact pressure developed for hard bearing materials (Uddin and Chan, 2018; Wang et al., 2014). Similarly, the acetabulum cup inclination and anteversion angles were varied to estimate the contact pressure and wear of hard bearings. Intermediate shell was sometimes modeled between the femur head and acetabulum cup (Besong et al., 2001; Liu et al., 2003) to analyze wear and contact pressure. Some of the literatures used FE model to investigate stress and wear using full hip joint model (Aghili et al., 2021; Bitter et al., 2017; English et al., 2016; Toh et al., 2021). One of the recent literatures developed FE model to investigate the wear by incorporating friction, induced due to vibration (Askari et al., 2015). For all these studies, global modeling approach i.e. full FE modeling was used to analyze the contact pressure and wear. Normally in global modeling (GM) approach, non-uniformity in selection of meshing elements was widely adopted in several works. The finer elements were used at the contacting interfaces of GM while coarser elements were used for the remaining regions. Figure 3 shows GM having different meshed element sizes.

The new and efficient way of analyzing the contact pressure and wear of biomaterials with less computational time and improved accuracy was proposed for first time using local modeling (LM) technique (Shankar et al., 2020). Figure 4 provides the details about the various models developed in FEA until now, to mimic hip implant design (like full model, half model etc.). The hip stem and taper junction model was mainly used in predicting the fretting wear for different biomaterial combinations. To analyze the linear and volumetric wear of bearings for different parameters like radial clearance, cup inclination angle and femur head diameter simplified half femur head and acetabulum cup model was considered. To analyze the fatigue wear of different material combination and to identify the suitable stem to withstand fatigue wear, FE model consisting of acetabulum cup, femur head and hip stem were used. The acetabulum cup along with UHMWPE liner model was used to estimate the contact pressure between the hard bearing surfaces to estimate its influence in stress shielding effect. Among these FE models, the simplified model of half of femur head with acetabulum cup as shown in Figure 3 was widely adopted in predicting the linear and volumetric wear.

The detailed procedure for submodeling or local modeling was shown in Figure 5, in which region of interest (RoI) being selected where one needs to find the contact pressure between the bearing combinations. This minimizes the problem of modeling the entire geometry wherein, the contact pressure was estimated at contacting interface only. This eliminates extra time being spent to refine the elements outside the contact interface to develop GM whereas in LM such tedious modeling can be avoided. In this approach, cut boundary region or region of interest (RoI) was selected from GM such that the contact pressure developed should be within the cut boundary region. Thus, by incorporating boundary conditions from GM result file, the LM was solved in faster manner. Earlier, only stress analysis was investigated using GM and LM for hard bearing combination using various cup inclination angles (Elkins et al., 2011). The finding revealed that increase in cup inclination increased contact stress.

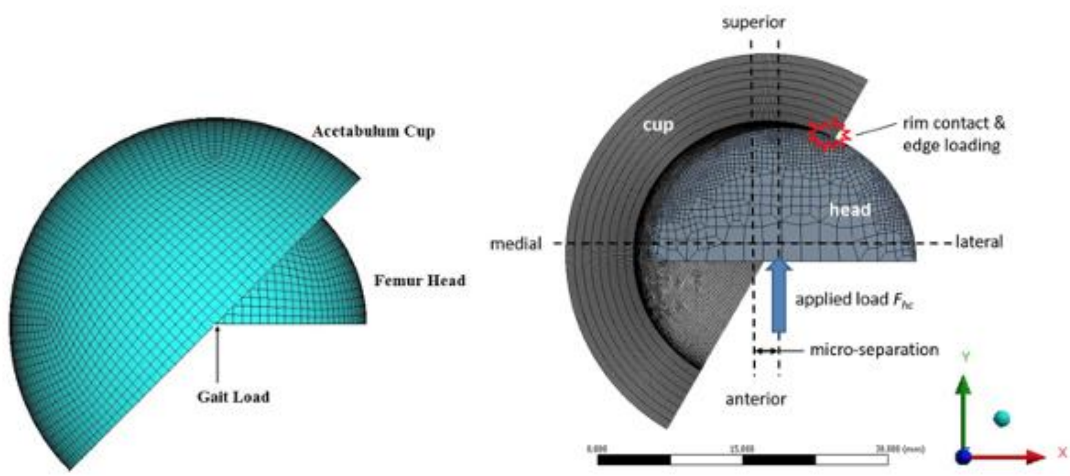

Figure 3: Global FE Model with uniform and non-uniform mesh (Nithyaprakash et al., 2018; Uddin and Chan, 2018).

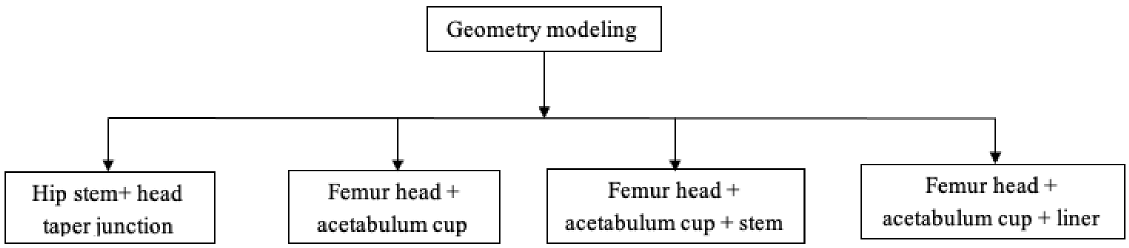

Figure 4: Geometric modeling types adopted for simulation.

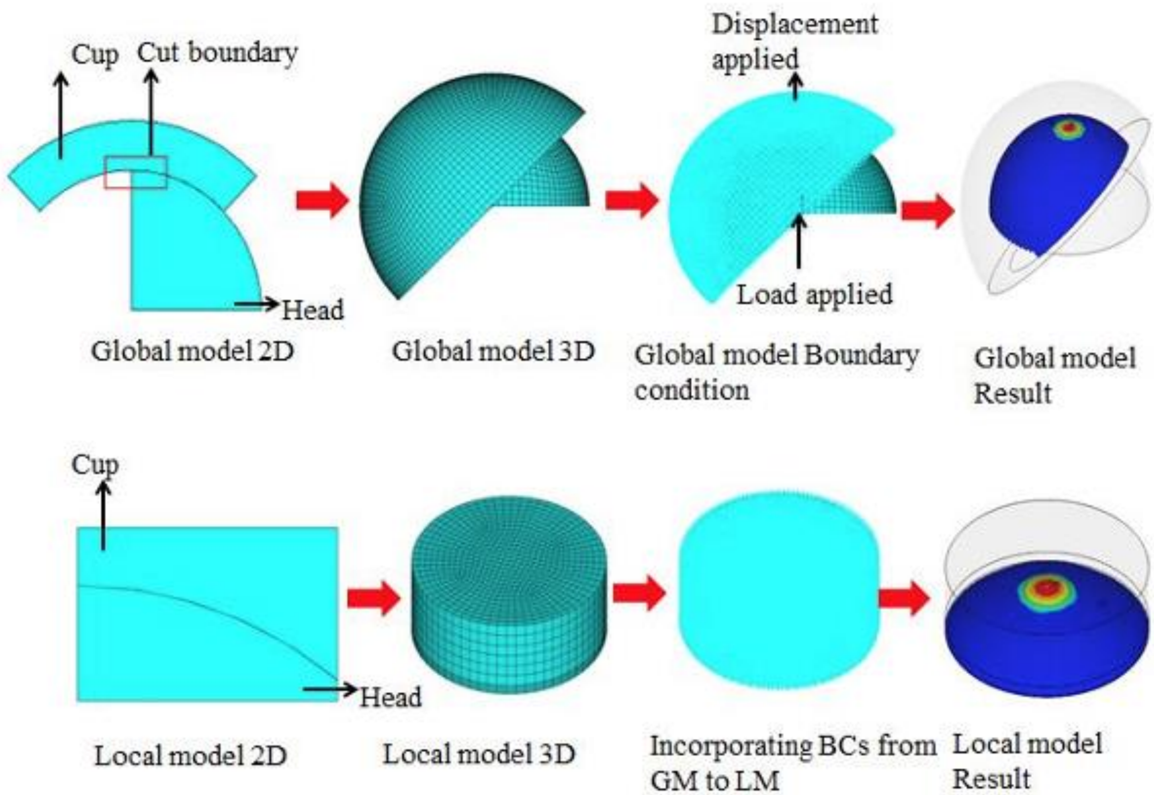

Figure 5: Development of LM from GM (Shankar et al., 2020).

2.2 Mesh Convergence Study

Mesh convergence refers to the selection of suitable element size to estimate the variations in contact pressure such that, the difference in strain and stress variation between two successive element size which should be minimum. For this, suitable mesh convergence study needs to be performed in predicting the contact pressure between the bearings. The element size was varied suitably for GM and LM. For GM, finer element size was used at contacting region, where contact pressure needs to be determined and for remaining regions coarser elements were used. For this, element size has to be varied suitably, so that the change in contact pressure and stress variations between element sizes was considerably small. The acceptable limit variation in contact pressure for change in element size should be within 5%. The problem of non-uniformity in meshing, leading to time consumption in GM, while for LM, uniform meshing of model helped in minimizing the time being spent on convergence study. One of the recent literatures (M. Uddin and Chan, 2018) varied element size from 1.5 to 0.25 mm and showed that for 0.25 and 0.5 mm element size, percentage difference in stress and strain was negligible. So, the authors adopted element size of 0.25 mm for contacting region and for remaining regions 0.5 mm element size as shown in Figure 3. (Liu et al., 2015) analyzed stress for different element size from 0.25 to 0.015625 mm and showed that 0.0625 mm had resulted in lesser percentage difference in predicting stress and strain values. The problem of using different element size for GM to minimize computational time, had led to less accuracy in predicting contact pressure. This problem was eliminated in case of LM as shown in Figure 5 in which uniform meshing of 0.75 mm element size was used.

2.3 Contact Mechanics in FE Modeling

Contact mechanics refers to establishing suitable contact between the interface of the cup and head model to estimate the contact pressure with the help of 2D and 3D contact and target elements. Sometimes the model was generated from image captured using CT scan and meshed using ABAQUS or ANSYS software. In ANSYS for 2D modeling, the element used was PLANE 42 while for 3D modeling SOLID 186 element was widely used. Contact was generated between the head and cup using contact and target elements. For 2D modeling, CONTA171, 172,175 and TARGE 169 elements were used to generate the contact between femur head and acetabulum cup. While for 3D modeling, CONTA 173,174,175 and TARGE170 elements were used. Mostly, augmented Lagrangian contact algorithm as well as penalty method were adopted to solve the contact model (Nithyaprakash et al., 2018; Shankar, 2014; Shankar et al., 2016; M. S. Uddin and Zhang, 2013). The initial gap between the cup and head was set to zero as point contact was established between cup and head to eliminate the convergence issue. Once the geometry update was over, the initial adjustment setting for contacting pair could be removed to capture contact penetration which represents worn out mechanism. The contact between cup and head could be established as surface-to-surface contact.

2.4. Gait Loads and Boundary Conditions

The human hip joint helps humans to execute different level of gait activities like normal walking, stair ascending and descending, sitting down or getting up (Bergmann et al., 2001; Kolk et al., 2014; Mellon et al., 2011). Sometimes even physically demanding gait activities like lifting and carrying load also exhibited by humans (Giarmatzis et al., 2015; Varady et al., 2015). Due to these gait activities, hip joint contact force (HJCF) would act on hip joints. Also due to these gait activities, hip joint also exhibits three types of rotations namely flexion-extension, abduction-adduction and internal external rotation. Due to these rotations, wear of materials take place inside the artificial hip implantation. Some of the literatures included these three rotations in FE modeling in computing the sliding distance (Harun et al., 2009; Uddin and Zhang, 2013). However, recent literatures included only flexion-extension angle in computing sliding distance as magnitude of this angle was found to be higher when compared with remaining angles (Nithyaprakash et al., 2018; Shankar et al., 2016; Shankar and Nithyaprakash, 2014). Using this method of calculating wear resulted in the reduction of computational time to greater extent and error in this procedure which was also quite negligible. Many literatures used peak force of different gait activities in estimating contact pressure, for different cup geometry as well as for different hard-on-hard material combinations (Shankar and Nithyaprakash, 2014; Shankar et al., 2020; Wang et al., 2014). These load values were generally taken from hip simulator study which predicted wear under adverse loading conditions (Nevelos et al., 2000; Stewart et al., 2003). The displacement of acetabulum cup was constrained in all three directions as the cup was attached firmly to the pelvic bone. The outer surface of the cup was constrained not to move in three directions while peak load was applied to the centre node of the femur head.

2.5. Wear Calculation

The wear of human hip prosthesis was predicted using the obtained contact pressure from FE model with the help of Archard wear law (Archard, 1953) and the modified Archard wear law (Marshek and Chen, 1989) for sliding bodies given by:

$$H = K_W \times P \times S \tag{1}$$

Where K_W is the wear coefficient in mm^3/Nm, P is the contact pressure (MPa) obtained from FEA and S is the sliding distance as the product of hip flexion angle with radius of femur head in (mm) due to rotation of femur head inside the cup. The wear coefficient is defined as the ratio of volume of material removed to the product of applied load and sliding distance.

$$K_W = \frac{\Delta V}{F \times S} \tag{2}$$

ΔV is a volume loss in mm^3, F denotes applied load or normal load in Newton. The cumulative linear wear of the bearing for particular million cycles was obtained by multiplying linear wear depth obtained with million cycles.

$$V_L = K_W \times P \times S \times N \tag{3}$$

Where, V_L is a linear wear (mm/μm).

Once the contact pressure was obtained, then the geometry of the surface was updated after certain N (million cycles). Contact pressure tends to decrease once the gait cycle starts increasing. Sometimes update of the geometry to capture running-in wear was estimated to be 0.2 million cycles if wear was estimated for shorter lifetime of 2 years or less. If the wear of hip implants were to be estimated for more than 2 years, then up to 2 million cycles, update interval was chosen as 0.2 million cycles, after that, update interval of 1 million cycles was adopted (Nithyaprakash et al., 2018; Shankar et al., 2016, 2017; Uddin and Zhang, 2013). The detailed FEA wear procedure for GM and LM approach are shown in Figure 6 and 7.

The cumulative volumetric wear of the bearing was obtained by multiplying the linear wear and contact are of the bearings

$$V_W = V_L \times A \tag{4}$$

Where, A is the contact area of the bearings.

To analyze the fretting wear at the head-stem taper region a new wear law called dissipated energy wear law is used. This uses single wear coefficient (α) wherein for single cycle of loading, cyclic wear depth given by

$$W_c = \sum_{i=1}^{n} \alpha \tau_i S_i \tag{5}$$

W_c denotes cyclic wear depth, τ_i is a contact shear stress and S_i is contact slip for time interval i.

$$W_d = \sum_{j=1}^{N/\beta} \beta \sum_{i=1}^{n} \alpha \tau_{i,j} S_{i,j} \tag{6}$$

Here, W_d denotes linear wear depth for loading cycles N for specific analysis j and β denotes wear scaling factor.

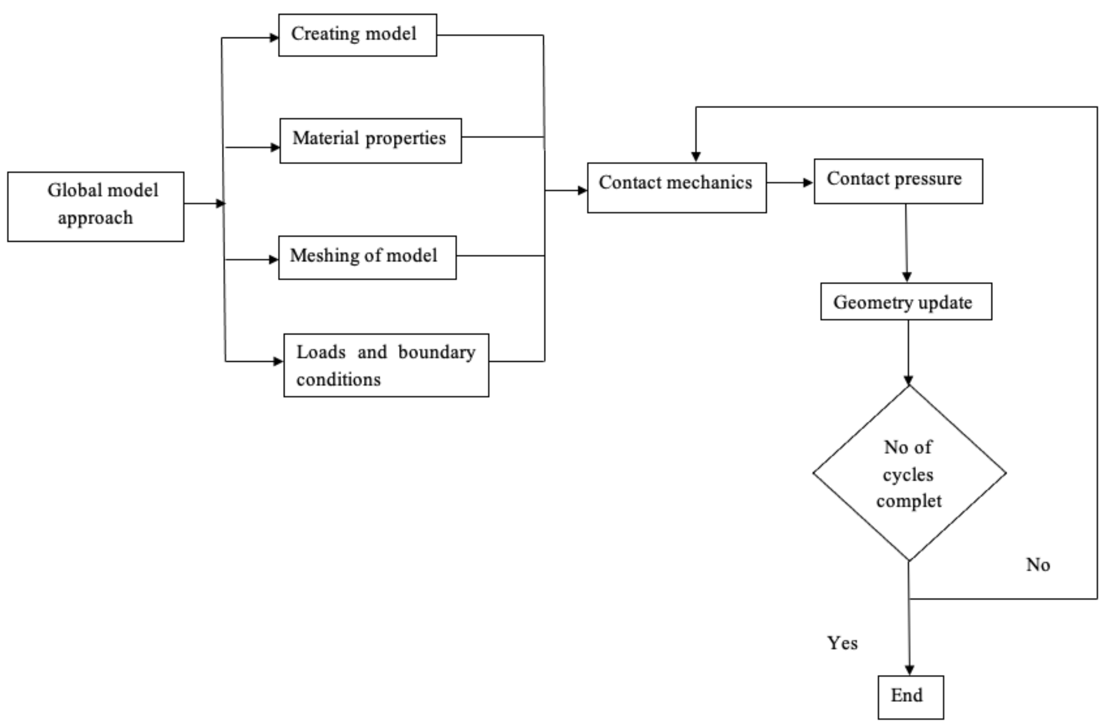

Figure 6: Global model procedure for contact stress estimation (Shankar et al., 2020).

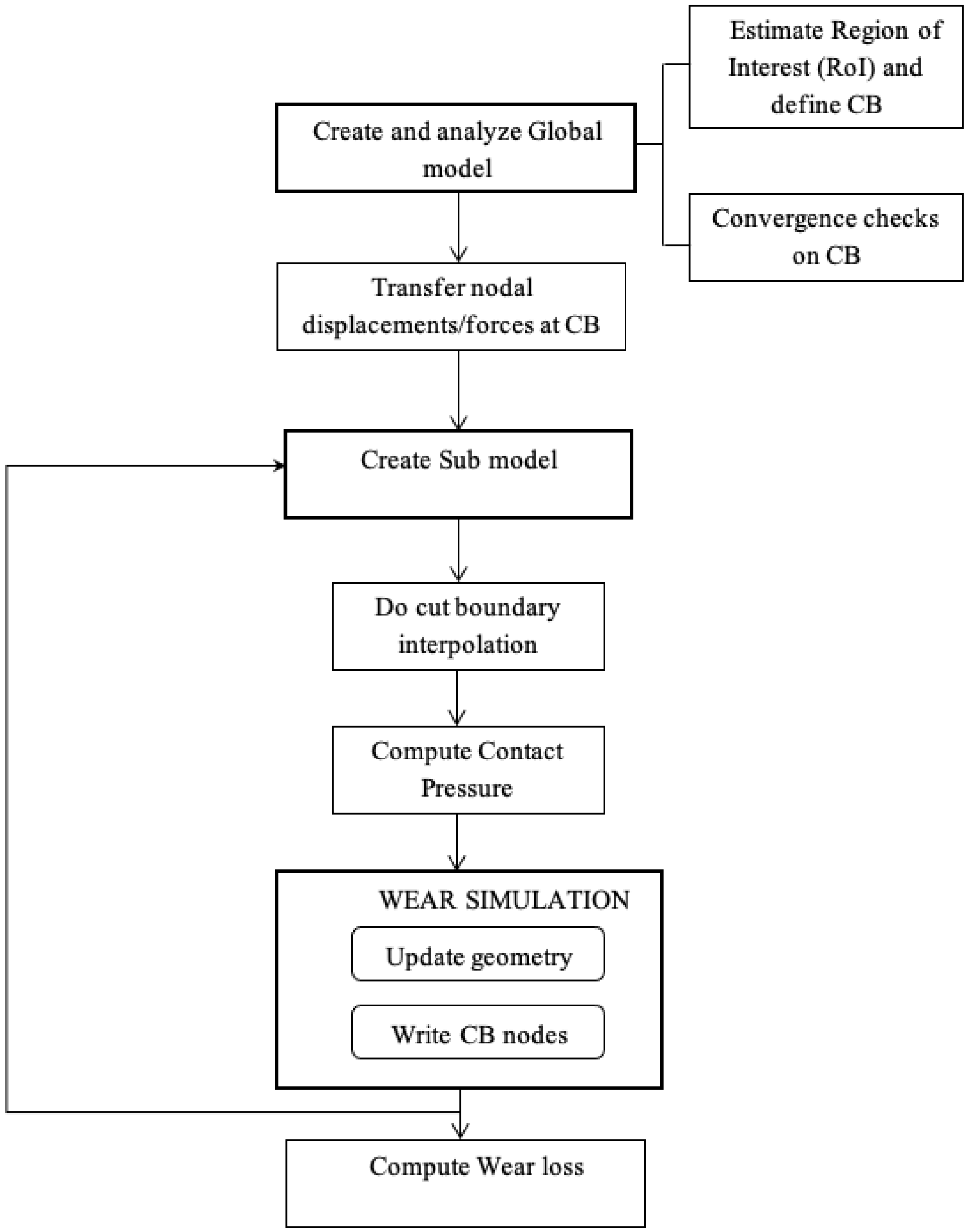

Figure7: Local model procedure for contact stress estimation (Shankar et al., 2020).

3.0 CONTACT PRESSURE AND WEAR IN M-o-M

3.1 Effect of Microseparation

The micro-lateralization or separation involves head lateral displacement from the original position during post operative period and it leads to edge loading and development of higher contact stress. The contact mechanics study of M-o-M under standard and micro-lateralization condition was analyzed for ellipsoidal geometry instead of standard spherical geometry. Findings revealed that no edge contact occurred for ellipsoidal geometry and approximately 80% reduction in contact pressure was noted for ellipse geometry of acetabulum cup (Wang et al., 2014). The use of spline geometry as an alternative to circular arc under different micro-lateralization for acetabulum cup showed greater reduction in contact pressure for increased micro-lateralization (Uddin and Chan, 2018). Another study investigated the contact pressure for M-o-M with different cup lip radius and cup orientations found that edge loading occurred for all cases beyond cup radius of 1 mm with increase in magnitude of contact pressure (Elkins et al., 2012). For micro displacement of greater than 0.1 mm, the magnitude of contact pressure higher than the yield stress of material was observed (Liu et al., 2015). The contact pressure generated for different microseparation values were shown in Table 2. From the contact pressure values, it was quite clear that the higher contact pressure was obtained for lateral displacement of head inside the cup leading to edge loading. Therefore, in order to counter the head lateral displacement, many researchers modified the cup geometry which included spline, ellipse and rounded corner for cup instead of sharp corner. They showed better reduction in contact pressure (Wang et al., 2014).

Table 2: Contact pressures analysis based on edge loading using FEA.

#	Material Combination	Biomaterial	Micro-separation	Cup Geometry	Contact Pressure (MPa)	Reference
1.	M-o-M	CoCr	0.25mm	Ellipsoidal	927.3	Wang et al., 2014
2.	M-o-M	CoCr alloy	≤2 mm	sharp corner	3600	Elkins et al., 2012
3.	C-o-C	Al_2O_3	80-250 μm	Spherical	800	Mak et al., 2002
4.	M-o-M	CoCrMo	0.1-2mm	Round corner	500-2500	Liu et al., 2015
5.	M-o-M	CoCr alloy	1-2.5mm	Sharp corner	1205	Uddin and Chan, 2019
6.	M-o-M	CoCrMo	0.5-2mm	Spherical	650	Liu et al., 2019

3.2 Effect of Radial Clearance, Cup Geometry and Loading

The radial clearance is defined as the clearance between the acetabulum cup and femur head which helps in lubrication to reduce wear. The effect of radial clearance influencing the contact pressure for different femur head size and acetabulum cup thickness using ABAQUS for M-o-M was investigated (Udofia et al., 2004). Reduction in radial clearance showed a drastic reduction in contact pressure, while reducing cup thickness and increasing femur radius was investigated using two-dimensional axisymmetric and 3D model (Udofia et al., 2004). The effect of radial clearance for M-o-M along with cup thickness, lip and studs outside the acetabulum cup showed that, the radial clearance highly influenced the contact pressure developed (Yew et al., 2003). Another study, using ABAQUS, reported the contact mechanics of M-o-M backed with titanium shell showed that the contact pressure was lower for larger radial clearance because of taper connection between bearing (Besong et al., 2001). The effect of UHMWPE backing used for M-o-M showed that relatively low contact pressure was observed due to UHMWPE backing and also showed loading direction had no significant change on contact pressure (Liu et al., 2003).

The contact pressure comparison between THR and HR was investigated and found that the latter had better contact area with reduced contact pressure using FEA (Liu et al., 2005). The wear analysis based on contact pressure developed was investigated for two types of M-o-M joints and found that the wear volume ratio between two types agreed with the theoretical wear volume (Cosmi et al., 2006). The wear of M-o-M for 50 million cycles was computed using FEA and the results agreed well with experimental hip simulator results. Moreover, the result also showed decrease in contact pressure for increase in gait cycles (Liu et al., 2008). Using two wear coefficients i.e., running-in and steady state wear coefficients (Harun et al., 2009), the wear of M-o-M using FEA for 50 million cycles for both cup and head was investigated, and the volumetric wear loss was found to be quite small. The effect of frictional contact in estimating the contact stress was investigated and showed that using friction coefficient, reduced linear wear rates and no change in volumetric wear was observed (Mattei and Di Puccio, 2013). The contact pressure and contact mechanics for M-o-M hip resurfacing for different cup angle was investigated and found that the edge contact occurred for lower cup inclination angles below 65° (Wang et al., 2012). Table 3 shows the wear of biomaterials based on FEA tool. From the table, it was quite clear that, higher magnitude of linear wear as well as contact pressure values were obtained for higher radial clearance and negligible difference in volumetric wear was observed. Also, for higher cup inclination angles increase in wear was observed for M-o-M bearing. Frictional contact had negligible influence in estimating the wear.

Table 3: Wear of biomaterials based on FEA study.

#	Bio-material	Mill. cycles/ Years	Linear Wear Per year/million cycles	Volumetric wear Per year/million cycles	In-vitro validation		Ref.
					Volumetric Wear/ million cycles	Ref.	
1.	CoCrMo	50	0.19 and 0.16 µm	0.24 mm^3 for running-in and 0.074mm^3 for steady state	0.22 $mm^3$and 0.65 mm^3 for running-in and steady state	Chan et al., 1999	Harun et al., 2009
2.	Al_2O_3/Al_2O_3	2	0.25 µm for 0.03 mm RC 2.375 µm for 0.75 mm RC	0.0375 mm^3	0.014–0.05mm^3	Nevelos et al., 2001	Shankar, 2014
3.	Si_3N_4/Si_3N_4	2	--	0.443mm^3	--		Shankar, 2014)
4.	PCD/PCD	2	0.0635 µm	0.00292 mm^3	0.0036mm^3		Uddin and Zhang, 2013
	Al_2O_3/Al_2O_3		1.317 µm	0.17275 mm^3	0.014–0.05mm^3	Nevelos et al., 2001	
	CoCrMo alloy		1.725 µm	0.1425 mm^3	0.22 $mm^3$and 0.65 mm^3 for running-in and steady state	Chan et al., 1999	
5.	Si_3N_4/Si_3N_4	20	--	--			Shankar et al., 2016
	PCD/PCD		0.039 µm	--	0.0036mm^3	Harding et al., 2011	
	Al_2O_3/Al_2O_3		1.060 µm	0.0413 mm^3	0.014–0.05mm^3	Nevelos et al., 2001	
6.	PCD/PCD	2	0.0935 µm for 0.03 mm RC 0.123 µm for 0.05 mm RC	0.00459 $mm^3$for 0.03 mm RC 0.00455 $mm^3$for 0.05 mm RC	0.0036mm^3	Harding et al., 2011	Shankar, 2014

			0.158 µm for 0.075 mm RC 0.18 µm for 0.1 mm RC 0.231 µm for 0.15 mm RC 0.371 µm for 0.3 mm RC	0.00466 $mm^3$for 0.075 mm RC 0.00467 $mm^3$for 0.1 mm RC 0.00470 $mm^3$for 0.15 mm RC 0.00471 $mm^3$for 0.3 mm RC			
7.	PCD/PCD	10	Normal walking with peak load 2410 N was 0.014 µm Normal walking with peak load 3327 N was 0.02 µm Sitting down/getting up 0.049 µm Carrying load 25 kg 0.02 µm Carrying load 40 kg 0.024 µm Stair up 0.03 µm Stair down 0.013 µm Ladder up (70°) 0.04 µm Ladder down (70°) 0.035 µm	Normal walking with peak load 2410 N was 0.001 mm^3 Normal walking with peak load 3327 N was 0.0012 mm^3 Sitting down/getting up was 0.0024 mm^3 Carrying load 25 kg was 0.0013 mm^3 Carrying load 40 kg was 0.0016 mm^3 Stair up was 0.0019 mm^3 Stair down was 0.0009 mm^3 Ladder up (70°) was 0.0025 mm^3 Ladder down (70°) was 0.0024 mm^3	0.0036mm^3	Harding et al., 2011	Nithyaprakash et al., 2018

3.3 Fretting and Fatigue Wear

The fretting wear or crevice corrosion occurs at modular junction of hip joint for large head diameters. The implant assembly load and taper junction influencing the fretting wear were investigated with the help of FEA. Besides assembly loading and taper junction, several other parameters like head size, taper size and implant junction positioning influence the fretting wear, as it could not be analyzed in-vivo. The influence of geometry and taper mismatch at head-neck junction led to the smaller mismatch that could ensure firmer connection at junction (Fallahnezhad et al., 2016). The finite element wear model using adaptive meshing to study the fretting wear at taper interface of hip prosthesis showed that the predicted result agreed well with experimental results (Bitter et al., 2018). The effect of assembly load at head stem taper interface was investigated using FEA and findings showed that the exact fretting wear loss was not computed using FEA due to lack of geometry update (Bitter et al., 2017). The different gait activities influencing the fretting causes more damage to joints which was investigated using FEA and the contact pressure values for different gait activities at head-neck interface were reported (Fallahnezhad et al., 2018; Farhoudi et al., 2017). Another study investigated fretting wear and suggested that the fretting wear could be reduced by upper contact offset at taper junction which reduced relative micro motion (Ashkanfar et al., 2017). The use of micro-grooved trunnions at taper junction was investigated and found that the smooth taper surface had better reduction in fretting wear (Ashkanfar et al., 2017). The fretting wear for taper angle mismatch was investigated and found that a decrease in contact pressure was noted for reduced taper angle mismatch. The fretting wear was found to be more at proximal mismatch angle (Fallahnezhad et al., 2017). The assembly loads applied at taper junction which develops contact pressure was investigated and found that the load applied should be below 4kN to minimize fretting wear (English et al., 2016). Larger head diameter showed increase in fretting wear for head-neck taper trunnions reported in another study (Elkins et al., 2014). Table 4 shows different parameters investigated to analyze the fretting wear using FEA.

The fretting-fatigue wear model was analyzed for 10 years computationally and found that the linear wear depth of 1 μm was observed for DMLS Ti–6Al–4V alloy due to better wear resistance characteristics (Zhang et al., 2013). For the fatigue wear study, initial press fit load of 1044 N was applied to femur head of FE model followed by peak load of 2648.2 N. The loads are applied and released in a cyclic manner which represents a typical hammer used by surgeons for press fit (Duda et al., 1997). The limitations in these approaches include the use of static load while in real scenario surgeons apply cyclic impact load during implant insertion and assembling. The lack of wear modeling still exists based on Archard's wear law which predicted wear as a product of contact pressure, micromotion and contact area.

Table 4: Fretting wear analysis based on different parameters

#	Material Combination	Software platform	Head Diameter (mm)	Angle Mismatch	Load/ Micro-motions	Wear /Friction coefficient	Contact pressure (MPa)	Linear Wear	Volumetric wear	Load cycles	Reference
1	CoCr/ CoCr	ABAQUS 6.13		0.024°	4 kN/0-38 µm	1.31×10^{-8} MPa-1	1940-2500				Fallahnezhad al., 2018
2	CoCr/ CoC	ABAQUS (6.14-3)	36	6°			350	0.100 to 62.03 µm	0.020-2.241 mm^3/yr	10 million	Ashkanfar et al., 2017
3	CoCr/ CoCr	FORTRAN code	36	0.024° to 0.124°	4KN / 0.54 - 1.65 µm		130			4,080,000	Fallahnezhad et al., 2017
4	CoCrMo/CoCrMo	FORTRAN code(UMESHMOTION subroutine)			0 N, 20 N, 44 N, 53 N, 70 N and 81 N /50 µm					150	Fallahnezhad et al., 2018

5	CoCr/ Ti6Al4V	ABAQUS 6.10	28, 36 and 44	5°40'	4kN/ 10–50 µm						Dyrkacz et al., 2015
6	Ti6Al4V alloy	ABAQUS 6.14	36		15 - 27 µm	0.3-0.45 MPa^{-1}	800-1400		0.79 ± 0.52 mm^3	10 million	Bitter et al., 2018
7	Co–28Cr-6Mo and Ti–6Al–4V	ABAQUS 6.13		0.015°		0.21 MPa^{-1}	375 - 715				Fallahnezhad et al., 2016
8	CoCrMo/Ti	ABAQUS/CAE 6.14–1	40	2.88°	<4 kN		928				Raji and Shelton, 2019
9	CoCr/ Ti	ABAQUS 6.13 (3DS Dassault Syste`mes Simulia Corp.)			4 or 15 kN/ 0.0744-1 µm		1202–930			10 million	Bitter et al., 2017

10	CoCr/ CoCr	ABAQUS	32	0.024°	4.5 to 6 kN / 0 – 38 μm	3.34×10^{-9} MPa^{-1} to 4.16×10^{-8} MPa^{-1}	900			100	Farhoudi et al., 2017
11	CoCr	ABAQUS	36		10 N	1.13×10^{-8} mm^3/Nm	68		0.374 mm^3		Zhang et al., 2013
12	CoCrMo	ABAQUS 6.13	36		0.1-3 kN				0.13-2.25 & 1.13-2.58$mm^3/10^6$ cycles		Di Puccio and Mattei, 2015
13	CoCr	ABAQUS 6.7-1			4 kN/15-20 μm		927.3			150	Wang et al., 2014

4.0 CONTACT STRESS AND WEAR IN C-o-C AND C-o-M

The effect of radial clearance and ceramic insert thickness influencing the contact pressure was investigated and found that the maximum contact pressure was obtained for large radial clearance while smaller clearance showed reduced contact pressure (Cilingir, 2010; Mak and Jin, 2002). The contact pressure distribution was investigated for acetabulum cup with various anteversion, and inclination angles showed that the microseparation with steep cup inclination would cause edge loading irrespective of steep cup inclination (Mak and Jin, 2004). Microseparation caused severe edge loading leading to higher contact stress for different acetabulum cup geometry (Liu and Fisher, 2017; Mak et al., 2002; Mak et al., 2011). The effect of cup angle and head lateral displacement showed increase in contact pressure of almost 210% compared with normal model (Sariali et al., 2012).

In general, the contact pressure was found to be quite high for C-o-C bearing when compared with M-o-M. However, later showed decrease in linear as well as volumetric wear because of improved wear resistance property. Al2O3-Si3N4 combination was investigated for 2 million cycles and results showed that for higher load or physically demanding gait activities, volumetric wear was found to be minimum. The reduction in volumetric wear was due to the reduced wear coefficient obtained for this combination (Shankar et al., 2020). The wear of ZrO2-Al2O3 combination was investigated under dry and saline lubrication and results revealed that, for physically demanding gait loads, higher volumetric wear was observed for dry as well as lubrication condition (Shankar et al., 2020). The wear of Si3N4-Ti6Al4V combination was investigated under five different bio-lubricants for some selected gait activities for 2 million cycles using sub modeling technique. The findings showed phosphate buffer solution bio-lubricant exhibited reduced wear among other bio-lubricants for selected gait activities (Shankar et al., 2020).

5.0 CONTACT STRESS AND WEAR ANALYSIS IN PCD-PCD

The wear of three hard material combination of CoCrMo-CoCrMo, Al2O3- Al2O3 and PCD-PCD was investigated using FEA for 2 million cycles and result showed that wear of PCD bearing had lower wear rate compared with remaining combinations (M. S. Uddin and Zhang, 2013). Another study using FEA investigated wear of Si3N4-Si3N4, Al2O3- Al2O3 and PCD-PCD for long term with different cup abduction and inclination angles. The findings revealed that PCD had least wear when compared with the remaining combinations (Shankar et al., 2016). The wear of PCD-PCD bearing for different gait activities was analyzed for 10 million cycles and result showed PCD had very minimal wear rate even for physically demanding gait activities (Nithyaprakash et al., 2018). The Si3N4- Si3N4 couple coated with nanocrystalline diamond (NCD), diamond-like carbon (DLC) and PCD was analyzed using 2F FE model with uniform coating thickness of 0.01 mm on head and acetabulum cup. The wear was estimated for 20 million cycles and findings revealed that PCD coating showed least wear (Shankar et al., 2018).

In all these works, PCD showed maximum contact pressure followed by ceramic material and minimum value obtained for metallic combination. Moreover, the wear was found to be least for PCD when compared with the other metallic and ceramic biomaterials. The contact pressure obtained for PCD material both in global and local model is shown in Figure 8.

6.0 VALIDATION OF FEA WEAR FINDINGS WITH IN-VIVO AND IN-VITRO APPROACHES

Most of the wear studies carried out using FEA validate their wear with in-vivo and in-vitro approaches. The in-vivo approaches mainly use plain radiograph to measure the wear after certain years of post-implantation. This mainly depends on patient age group; type of activity being carried out. The validation of FEA study with this approach leads to difference in wear estimated, because of, many parameters taken into account in analyzing wear in in-vivo, leading to error. The parameters include gender, age, obesity, type of activity being carried out and follow-up period etc. For validation of FEA results, it would be better to compare with experimental results from hip simulator or POD as the loading and other parameters like cup inclination angle, femur head size, microseparation etc could be easily replicated in FE model. The experimentally obtained wear of implants from hip simulator as well as POD/BOD and predicted FEA wear results are shown in Table 3 and it clearly indicates that predicted FEA wear result (Harun et al., 2009; Mattei and Di Puccio, 2013; Nithyaprakash et al., 2018; Shankar et al., 2020; Shankar et al., 2020; Shankar et al., 2020; Shankar et al., 2018; Uddin and Zhang, 2013) agreed well with experimental result (Chan et al., 1999; Harding et al., 2011; Nevelos et al., 2001; Stewart et al., 2003).

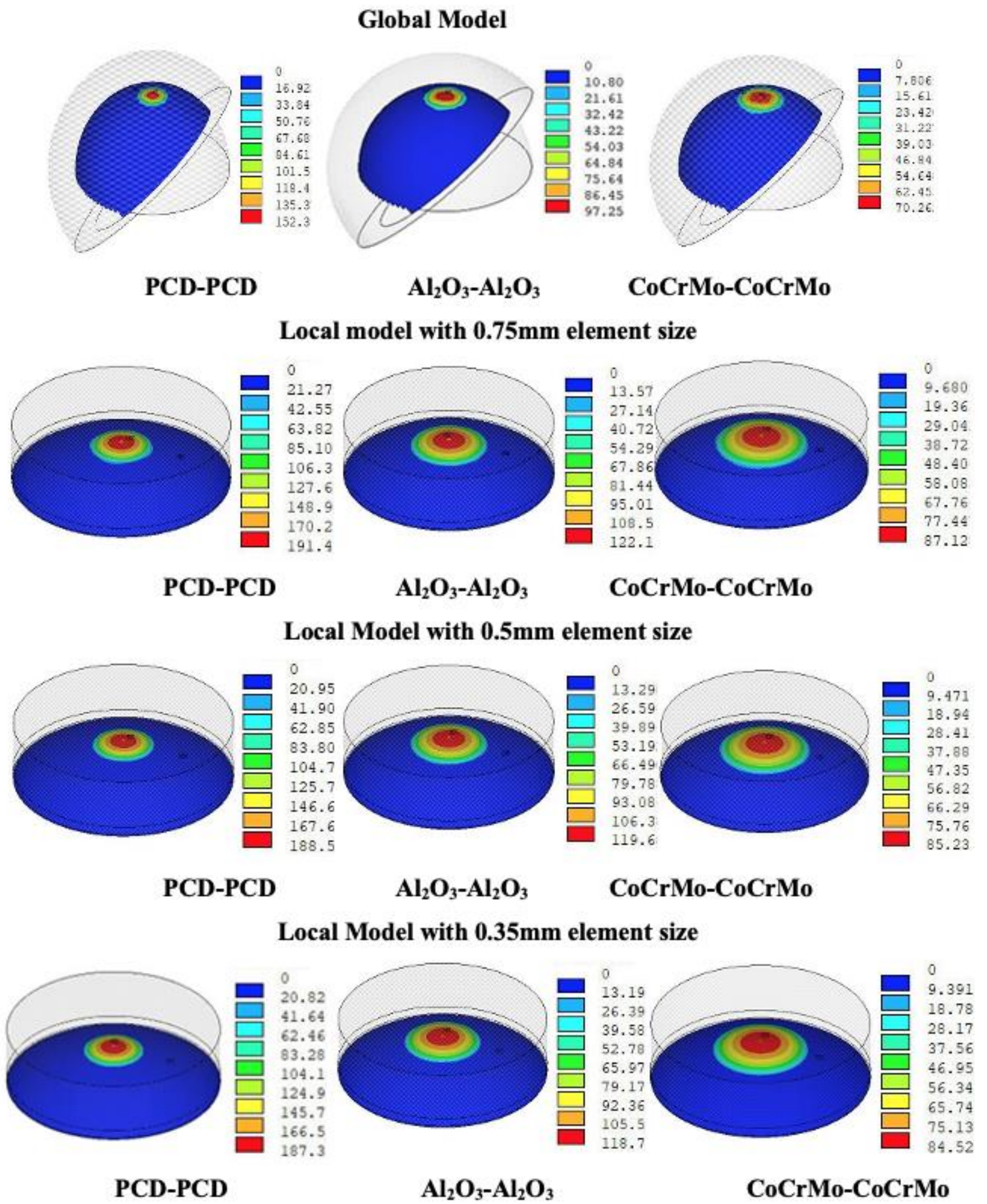

Figure 8: Contact pressure (MPa) for initial gait cycle of global mode vs. local model (Shankar et al., 2020).

7.0 DISCUSSION

The generation of contact pressure highly influences the linear wear in human hip implants. The change in predicted contact pressure was mainly due to the parameters like radial clearance, mesh refinement at contacting zones causing huge variation in contact pressure. This leads to change in linear wear, however, no change in volumetric wear was observed in most of the studies (Shankar and Nithyaprakash, 2014; Shankar et al., 2020). The new approach called LM or submodeling resulted in more accurate prediction of contact pressure compared to GM. The submodeling approach showed higher contact pressure leading to increase in linear wear compared to GM (Shankar et al., 2020; Uddin and Zhang, 2013). More than 26% increase in contact pressure was noted for poly crystalline diamond (PCD) for LM approach when compared with GM. Similarly increase in linear wear was found to be more than 17% for LM when compared with GM (Shankar et al., 2020). Moreover, time taken to solve LM was very minimum compared to GM with more refined mesh than GM as it minimizes the requirement of high-end computing system (Shankar et al., 2020).

Mesh refinement is the major task in FEA wear study as improper selection of element size leads to less accuracy in contact pressure generated for FE model. This leads to inaccurate wear estimation as reported in literatures showing variations in wear prediction when compared with experimental study (Harun et al., 2009; Nithyaprakash et al., 2018; Shankar and Nithyaprakash, 2014; Uddin and Zhang, 2013). For proper mesh refinement, researchers have to spare time for convergence study in choosing the finer element with minimum stress and strain difference. This requires high end computing system and requires in-depth knowledge in proper selection of element size at contacting region where contact pressure needs to be analyzed. To minimize the problem of mesh refinement, local modeling technique could be adopted, and uniform meshing of model is possible. This eliminates the need for high end computing system and computational time is also greatly reduced.

8.0 LIMITATIONS OF CURRENT APPROACHES AND FUTURE SCOPES

8.1 Based on Wear and Friction Coefficient from Experimental Study

The friction coefficient plays a very minimal effect in predicting wear using FEA as it affects the linear wear depth and volumetric wear remains same (Mattei and Di Puccio, 2013). The mild variation in contact pressure leading to change in linear wear was also reported. However, the wear coefficient plays a major role in estimating the wear of human hip joint using FE model. The wear and friction coefficient of bearings were estimated with the help of in-vitro studies namely POD and hip simulator (Aherwar et al., 2018; Barbour et al., 1999; Guezmil et al., 2018; Kaddick and Wimmer, 2001; Lu and McKellop, 1997; Luo et al., 2013; Saikko, 2005; St. John et al., 2004). The wear coefficient values were estimated under dry and lubrication conditions using these approaches. Differences in test procedures exist between these two techniques; in hip simulator it is quite possible to simulate realistic wear happening inside the human hip joint. Therefore, wear and friction coefficient values obtained from the hip simulator study was quite different from POD tribometer study, yet most of the researchers also prefer POD tribometer considering equipment cost. All the finite element studies used these wear coefficient values in estimating the wear of human hip implants. However, most of the wear coefficients analyzed using hip simulator or POD tribometer focused only on normal walking gait activity with acetabulum cup inclination of 45°. The wear coefficient also varies with respect to acetabulum cup inclination angle and gait activities carried out in everyday life. The FEA wear study focused on analyzing wear for different gait activities like stair ascending, descending and carrying load using wear coefficient of normal walking gait activity (Nithyaprakash et al., 2018). However, it was not realistic approach in estimating wear as equivalent gait activity wear coefficient need to be applied in FEA to estimate the wear. One of the recent studies estimated the wear of different gait activities using FEA with the help of realistic wear coefficients for silicon nitride against alumina combination (Shankar et al., 2020). Likewise, using hip simulator for risky gait activities and other parameters influencing wear like radial clearance and microseparation etc., wear coefficient has to be determined. This will effectively enhance the wear prediction in more accurate manner.

8.2 Based on Hip Joint Modeling Parameters

The FE modeling approaches that are discussed in the current review has its own limitations. The parameters like microseparation, radial clearance and cup inclination, ante version angle and taper angle mismatch were analyzed using global and local model approaches. These modeling approaches have limitations in terms of computational time, mesh refinement and accuracy in estimating the wear while analyzing these parameters. Besides computing wear, FEA plays a major role in selection of suitable biomaterial based on von Mises stress and contact pressure for human hip implants. Though, they are helpful in predicting the wear and in design of implants, they still have their own limitations based on expert's knowledge.

The FEA tool could not replicate the realistic wear behavior of the implants, instead it is helpful in analyzing the above parameters which were difficult to be analyzed either in-vivo or in-vitro. Real human body conditions were difficult to be simulated in FEA studies. Therefore, taking into considerations of all these factors, a new better modified and improved FEA approach needs to be developed for accurate and realistic hip wear estimation.

CONCLUSIONS

The present review article focuses on consolidating the works carried out in wear analysis of human hip prosthesis using FEA approach. The use of FEA tool in analyzing the wear of bearings by varying different parameters like radial clearance, cup thickness, microseparation, head- stem taper junction and head diameter for different gait loading conditions with the help of obtained contact pressure were discussed. These parameters were difficult to be analyzed in experimental approach as they were costlier and time consuming. Earlier global model approach was used in analyzing the wear of implants. This global model approach leads to non- uniformity in meshing of model leading to less accurate wear analysis. With the continuous evolution of FEA, recently developed local model approach was used in analyzing the wear of implants in more accurate manner. Also local model approach reduced computational time and improved accuracy of wear. Most of the FEA results agreed well with the experimental results. Thus, FEA tool could be the most reliable tool and hence it is widely recognized by many experts in analyzing the wear of the bearings. In near future, with gaining ground in knowledge in FEA tool among experts, the limitations and barriers in modeling and estimating wear of human hip joint using FEA tool could be improved.

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