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Finite Element Analysis of Stresses from Hip Implants with Different Head Sizes

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ABSTRACT

The desire of younger patients to continue to be physically active after hip replacement surgery has led to the use of hip implants with larger diameter femoral heads. Because the geometry of the socket changes with a larger head, the force transmitted to the implant stem will change. In this study, we sought to answer the following questions. Does increase in head diameter alter the stress pattern seen in bone and stem after a rotational activity? Does a larger diameter head produce greater stress? To answer these questions, finite element analysis of von Mises stress distribution under the loading condition of a single-leg stance with pelvis rotation in two computer-generated virtual models, a large diameter head metal-on-metal and a smaller diameter head metal-on-polethylene total hip arthroplasty (THA) model was performed. Higher stress concentration was seen in the large diameter head model than in the conventional smaller diameter head model, but stress patterns were similar. Peak stress was found in the bone surrounding the distal third of the implant stem and in the neck portion of the implant. A large diameter of the femoral head of a THA implant gives rise to higher stress after a simulated rotational activity than the conventional, smaller femoral head model.

Keywords: *Conventional metal-on-polyethylene total hip arthroplasty, large-diameter-head metal-on-metal total hip arthroplasty, hip rotation, finite element analysis*

INTRODUCTION

Younger, more physically active individuals are having hip arthroplasty, and these individuals would like to be able to continue their previous level of physical activity after their surgery. For this reason, it would be desirable to have implants available that allow a greater range of motion than the commonly used metal-on-polyethylene (MoP) implant.

The conventional MoP bearing used in hip replacement surgery has a head with a smaller diameter than the acetabular socket into which it is placed. This implant has been reported to have satisfactory long-term clinical and radiographic results [1, 2]. However, in the past decade, a metal-on-metal (MoM) bearing has become available. An advantage of the MoM bearing for more active individuals is that it can be obtained in a larger diameter head (LDH) size than is possible for an MoP bearing. The larger head size allows an increased range of motion and has less risk of dislocation [3, 4]. However, most reports of MoM THA have been on implants using a metal-inlay liner coupled with a smaller (28-mm or 32-mm) femoral head [5-12]. Clinical reports on large heads are relatively rare and limited to studies of the earlier McKee-Farrar implants and to later studies of resurfacing arthroplasty [3, 13-15].

Little is known about the inner stresses caused by physical activity in implants with different head sizes. Golf is an example of a physical activity that is allowed after hip and knee replacements [16, 17]. During shoveling movements or a golf swing, the head of the implant rotates in the socket of the hip joint and the force applied to the implant and through the implant to the femur should depend on the surface area of the head and the distance of each element from the center of rotation. These forces should therefore be affected by the diameter of the head. The current study was a finite element analysis of a virtual THA model, using a swinging motion (single-leg stance with pelvic rotation) as the loading condition and heads that varied in diameter, a LDH MoM and a smaller diameter MoP head.

Our questions were (1) Does use of a large head change the stress pattern in the implant and femur during pelvic rotation? (2) Are stresses on a large head THA implant during pelvic rotation higher than those seen on a conventional MoP THA?

METHODS

Overview of Experimental Protocol

Virtual total hip arthroplasty using implants with two different head sizes was created from digital computed tomography images of an artificial pelvis and femur and from images of two clinically available implants constructed with computer software. Virtual arthroplasty was then performed with each implant, and the resulting images were transferred into finite element analysis (FEA) software. After the accuracy of the FEA method was tested mathematically and by comparing the virtual results with those obtained from a physical model, the virtual models were subjected to a simulated golf swing and the von Mises stresses that resulted with each implant recorded and analyzed.

Model Elements

The artificial pelvis and corresponding femur model were obtained from SAWBONES (Pacific Research Laboratories Inc, Vashon Island, WA, USA). The implants used were an LDH MoM implant and a conventional MoP implant. The LDH MoM implant created had an acetabular component with outer diameter 58 mm, inner diameter 54 mm, height 29 mm, and a femoral head with outer diameter 52 mm, inner diameter 56 mm, and height 29 mm. The MoP implant had an acetabular component with outer diameter 58 mm, inner diameter 52 mm, a polyethylene liner with outer diameter 52 mm, inner diameter 28 mm, and a femoral head with outer diameter 28 mm, inner diameter 15 mm, height 21 mm. The stem for each implant was a cementless proximal fitting stem, distal diameter 15 mm. This stem was chosen because it is the most common stem type used in modern primary THA. All constructs used in the simulation were replicas of commercially available equipment, in order that the three-dimensional computer model resemble the clinical situation as closely as possible.

Model Setting

For the mesh used in the FEA, in consideration of the geometric shape of the structure, the second order tetrahedral element (solid 92 element) of the ANSYS software was selected for the volume part. Free mesh generation was performed after the construction of the analysis parameter model.

All material settings were based on information in the literature, including the settings of cortical bone, cancellous bone, and hip prosthesis [18, 19]. They were modeled as a homogenous linear elastic continuum exhibiting isotropic properties.

The porous coating of the stem was fully bonded to the cancellous bone to simulate the condition of bone in growth. For the contact simulation settings, contact pair element (Contact174/Target170) was selected, and the non-linear contact element was constructed using the node-to-surface mode to simulate the phenomenon of contact and friction between the LDH and the prosthetic cup, between the conventional head and polyethylene liner (PE liner), and between the femoral stem (non-coating part) and the inner wall of the femoral cortex. The coefficient of friction between the stem and the endosteum was 0.4 [20]. The coefficients of friction of LDH-prosthetic cup (CoCr-CoCr) and conventional head-PE liner (CoCr-PE) coupling were 0.15 and 0.07, respectively [19, 21].

Generation of Virtual FEA Model

Digital computed tomography images (*. DCM) of the artificial femur and pelvis were obtained and processed using medical image processing software (Amira 4.1, Mercury Computer Systems, Chelmsford, MA, USA). The resulting files were then transferred into the finite element analysis software (ANSYS Ver. 11.0, Canonsburg, PA, USA) used to construct the entity model of the femur and pelvis.

The acetabular component and two femoral heads were constructed by CAD software (Pro/Engineer, Wildfire 2.0, Parametric Technology Corp., Needham, MA, USA). A three-dimensional model of the cement less U2 stem prostheses (model #1104-1078, distal diameter of 15 mm) was provided by the manufacturer (United Orthopedic Corp., Hsinchu, Taiwan).

After constructing the femur, pelvis, and prosthesis, the prosthesis was implanted in the femur and pelvis to the appropriate location. The range of the cancellous bone that was not replaced by the prosthesis was from the greater trochanter to the distal end of the femoral stem porous coating area (Figure 1).

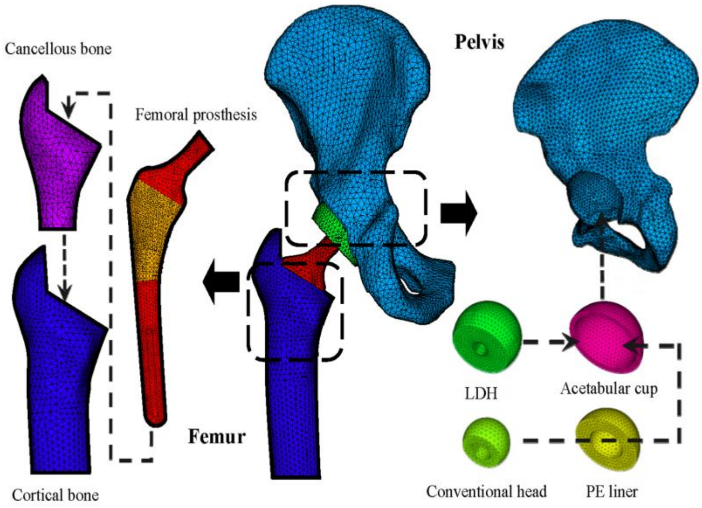

Figure1: *Construction of THA prosthesis and corresponding bony structure.*

Convergence Test

To ensure that the algorithms used in the analysis gave robust and reliable numerical values, analyses were performed using a series of mesh sizes from size 9mm to size 4mm to test convergence. The load condition used for this test was a 2000 N downward vertical (that is, perpendicular to horizontal) load at the center point of the hip prosthesis. The kinematic boundary condition for all nodes on the distal cross section were rigidly fixed (i.e. displacements and rotations were equal to zero). The output variable monitored was the total strain energy and the vertical displacement of selected points within the cortical bone. The results for each mesh size were compared to the results for mesh size 3mm. The difference between mesh size 4 and mesh size 3 was < 0.3%. Therefore, mesh size 4 was used as the mesh element size in the analysis; there were a total of 138,499 elements and 202,655 nodes in the over-all model using this mesh size.

Comparison with A Physical Model of an Artificial Femur

In addition to testing the reliability of the virtual model simulation by testing convergence over a range of mesh sizes, we tested its reliability by comparing its results to those of a physical simulation, using an artificial femur and a tensile testing machine. In the virtual simulation used for comparison with the physical simulation, a single downward vertical force of 2000 N was applied at the most superior edge of the hip prosthesis, and zero degrees of freedom was set at the distal node of femoral stem as the boundary condition. In the physical simulation, a downward force was applied to an artificial femur at a velocity of 0.05 mm/s until 2000 N was reached. Strain values at corresponding locations on the virtual and physical model were then compared. The results of the physical experiment and the computer simulation were within an acceptable range (20%) and were a further verification of the reliability of the finite element analysis.

Loading and Boundary Conditions for the Golf Swing Simulation

For the virtual pelvic rotation, the acetabular and femoral components were set in standard position with 45° inclination and 15° anteversion of the cup and neutral alignment of the stem. A loading condition based on the downswing-to-impact phase of an arm and shoulder swing with a single-leg stance on the leading foot was applied on the hip. The joint reaction force was 2,872 N together with a 1,237 N abductor muscle force (about 3.4-times body-weight), as described by Wang et al. [22] The pelvic rotation angle at impact was set at 29.4° and the rotation velocity at impact was set at 258.8° per second. The settings for the swing were based on the results of a motion analysis of a low-ball-velocity golfer (mean ball velocity of 55.7 meters per second; mean age of 58.5 years;

mean body mass index of 26.6; mean USGA handicap of 15.1 strokes) [23, 24]. The loading configuration is shown in Figure 2.

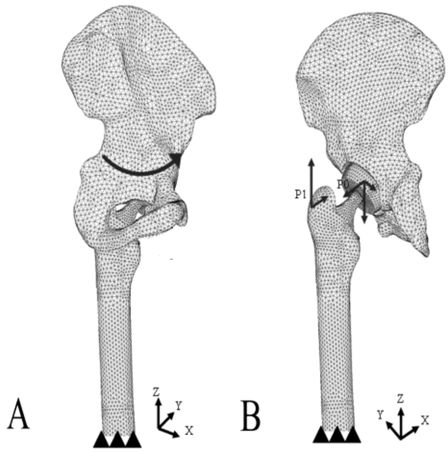

Figure2.*Golf swing loading condition of the finite element model (demonstrated with the LDH MoM THA). (A) The distal end of the construct (▲) was constrained in all directions as a boundary condition. The coordinate system (X, Y, Z) represents rotation of the pelvis (arrow) along the Z-axis on the left hip. (B) The joint reaction force (P0) is the resultant force of 616 N, 171 N, and 2800 N on the (X, Y, Z) coordinate, respectively. The abductor force (P1) is the resultant force of 430 N and 1,160 N on the (X, Z) coordinate, respectively.*

The von Mises stresses were analyzed by the ANSYS software on the periprosthetic bone and the stem, using the seven regions of interest (ROI) described by Gruen et al. [25] The von Mises stresses on each ROI were averaged to represent the stress distribution under the loading condition.

Results And Discussion

Stress Patterns with Large and Small Diameter Heads

The diameter of the implant head does not affect the stress pattern seen after the simulated swinging motion. The mean von Mises stresses on the bone surrounding the implant and on the implant stem (Figure 3) follow a similar pattern with both implants. In the bone, stress concentration is seen in the area surrounding and immediately distal to the distal third of the stem (ROI 3-5). The highest von Mises forces are in ROI 5. In the stem, stress concentration is highest at the distal tip (ROI 3-5) and in the uppermost zone (ROI 7) on the medial side (the area adjacent to the neck of the implant). The von Mises stress distribution and peak stress in the femoral cortex (Figure 4) show similar patterns of stress distribution with large and small diameter heads and a peak stress over the distal femoral cortex. In the longitudinal-section view of von Mises stress (Figure 5) stress concentration is seen on the lateral side of the distal stem and over the neck of the stem in both the LDH MoM THA model and the conventional MoP THA model. Maximum stress is seen around the neck of the stem.

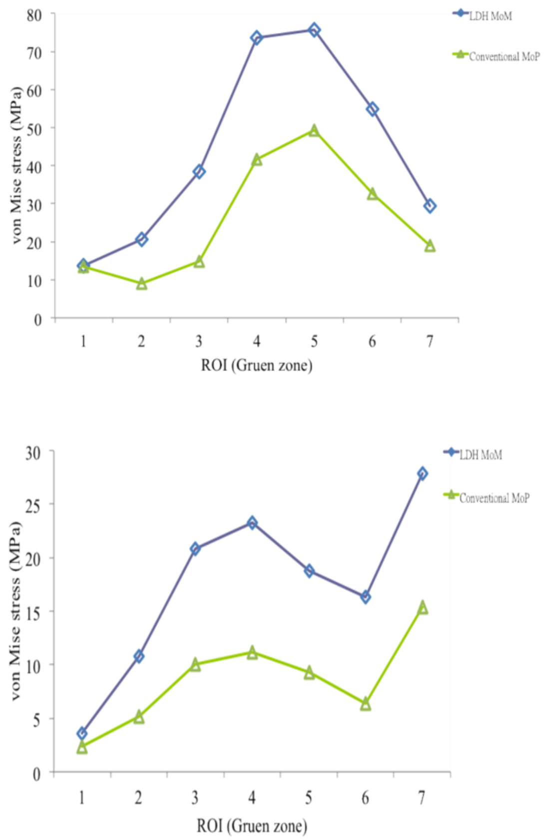

Figure3.*Mean von Mises stresses in the 7 ROI (Gruen zones). (A) periprosthetic bone and (B) stem.*

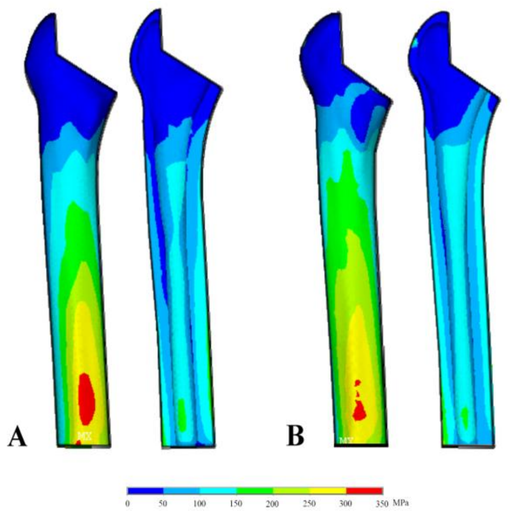

Figure 4.*von Mises stress distribution and peak stress in the femoral cortex. (A) LDH MoM THA and (B) conventional MoP THA*

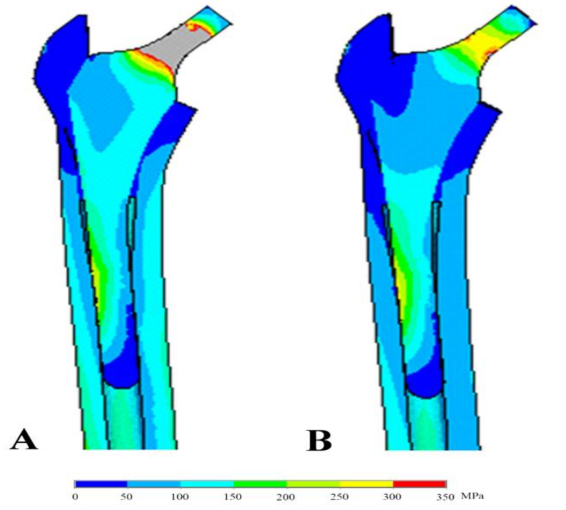

Figure 5.*Longitudinal-section view of von Mises stress distribution and peak stress. (A) LDH MoM THA and (B) conventional MoP THA*

Comparison of Stress Intensity on Large and Small Diameter Head Implants

Stresses were higher in both bone and stem in the implant with the larger head diameter (Figure 3). In the bone, the von Mises forces in ROI 5 are 76±31MPa in the LDH MoM and 49±21 MPa in the conventional MoP THA models (Figure 3). In the stem, the highest stresses are 28±1 MPa in the LDH MoM and 15±7 MPa in the conventional MoP THA model at the neck (ROI 7). The peak stress in the femoral cortex, however, (Figure 4) is 332 Mpa in the LDH MoM THA model and 343 MPA in the conventional MoP THA model.

Patient's expectations after joint replacement include relief of symptoms and improvement in physical function [26]. Increasing demands for a return to more vigorous activities after surgery have been observed in patients who choose the have their joints replaced at a younger age and clinicians have relaxed some of their previous restrictions on activity [16, 17]. One reason for this is the improvements that have occurred in joint implants over the years. However, a study of 34 patients with a mean follow-up of 6.3 years found that young patients with conventional MoP THA were in fact not as active as thought [27].With the use of LDH with metal-on-metal bearings in hip resurfacing surgery, other investigators reported an increase in the activity of patients after the index surgery [28, 29]. Encouraged by the results of hip resurfacing arthroplasty, LDH MoM THA was introduced as an alternative option for primary THA. Theoretically, LDH MoM THA provides improved stability, a greater range of motion, and a reduced surface wear on the bearing that can benefit patients and allow more activities than conventional MoP THA. The mating of LDH with a standard femoral prosthesis also eliminates the risk of femoral neck fracture associated with hip resurfacing surgery. However there have been no investigations of internal stresses produced by motion with the two implants, so we undertook a finite element analysis of stresses in virtual models of the two implants to determine whether the stress patterns produced by a simulated golf swing were similar and whether the larger head implant produced higher stresses during this type of motion.

Limitations

The current study has limitations primarily because although finite element analysis provides information to enable the reader to understand stress distribution, it cannot represent the real situation in the human body. the rotational swings of each individual will differ in terms of the swing path, rotation angle, velocity, posture, coordination between muscle groups, and so on. Because the hip rotation accompanying an arm and shoulder swing is a dynamic process and the joint reaction force constantly changes during different phases of the swing, we used the average rotation angle and rotation velocity at impact based on the results of low-ball-velocity golfer for our model [23, 24]**.** Rotation angle and velocity would differ somewhat for a similar movement such as shoveling. The loading condition in this study was a simplified model obtained by rotating the pelvis horizontally on the femoral head. It was also unknown how the stress distribution would be changed if the porous coating of the stem was not defined as fully bonded. The friction coefficient of 0.15 for metal-on-metal used in this study was based on the experimental value with a clearance of 75 □ m and bovine serum or synovial fluid as a lubrication film, and without the consideration of equatorial contact [18, 21]. Deformation of the acetabular component after press-fit implantation into the pelvis was also not modeled in this study, although such deformation would affect the stresses between the bearing surfaces [30]. Nevertheless, this finite element analysis is able to provide information for guidance before reports of longer term clinical follow-up of the LDH MoM THA are available.

Similarity of Stress Patterns

In the current study, a finite element analysis of a simulated arm/shoulder swing after hip replacement with implants with different diameter head sizes showed a concentration of stress in the distal third of the stem and the surrounding bone and a similar concentration of stress proximally, in the stem neck. The stress pattern was similar with both implants, so one might suspect that it was the specific physical activity, rather than variations in the implant design, that determined the stress pattern. This possibility should be a subject for further research.

Increased Stress in the Large Diameter Head Model

Under the loading condition used in the current study, the von Mises stress was higher in the LDH MoM THA model than in the conventional MoP THA model both in the region of periprosthetic femur bone and in the stem (Figure 3A and 3B). The highest stress concentration on the periprosthetic femur bone was noted in the ROI 4 and 5 (distal femur) region and the stress values were twice as high in the LDH MoM THA model as in the conventional MoP THA model. In clinical cases, this situation would usually mean that patients with LDH MoM THA would suffer more thigh pain after surgery than patients with conventional MoP. However, clinical reports of a LDH mated with a standard femoral stem are limited. In one previous study in 107 patients with a mean follow-up of 1.1 years, Garbuz et al. reported that clinical outcomes and improvement in quality of life were similar in patients randomized to hip resurfacing arthroplasty or LDH MoM THA [31]. The stem used in Garbuz's study was the M/L Taper stem (Zimmer, Warsaw, IN, USA) while the stem used in the current study was a distal fluted stem (U2 stem, United Orthopedic Corp., Hsinchu, Taiwan). The torsion stability of a fluted stem has been shown to be the highest in comparison with other stem designs [32], and in the current study using this stem, the highest pressure was between the distal stem and the corresponding femur bone. How the stress distribution would be changed with an M/L taper stem is of interest and needs to be examined.

Other Forces to Consider

Of course head size is not the only consideration to be taken into account when selecting an implant. When cement less stem is used, as in our model, the amount of micro motion between stem and bone during movement is important, because too great a degree of micro motion will decrease bone in growth and affect implant stability. Peterson et al (2009) studied micro motion using cadaver femurs and simulated stair climbing and found increased rotational stability and decreased micro motion with larger implants. The physical properties of the materials used are also important. Fialha et al (2007) found ceramic-on-ceramic and metal-on-metal surfaces to have higher maximum contact pressure and therefore higher frictional wear than polyethylene-on-ceramic or polyethylene-on-metal surfaces. Heat generation was higher in the metal-on-metal surface than in the other three surfaces.

Conclusions

The findings of this study may prove to be important and clinically relevant because we were able to demonstrate a high stress concentration on the prosthetic neck of a stem mated with a LDH under the loading condition of pelvis rotation. There were also higher stresses on the distal femur bone, when a LDH instead of a 28 mm conventional femoral head was used in total hip arthroplasty. Since higher stresses would be expected to be experienced during single-leg stance with pelvic rotation, patients after LDH MoM THA should be careful when performing certain physical activities.

The highest stress concentration on the stem was noted in ROI 7 (medial femoral neck of prosthetic stem) in the present study, and the stress value was almost twice as high in the LDH MoM THA model as in the conventional MoP THA model. In a report by Garbuz et al., concentrations of serum cobalt and chromium increased significantly from baseline in the LDH MoM THA group. Therefore the authors strongly recommended against further use of that particular LDH MoM THA design in the belief that the adapter connecting the LDH and the femoral stem introduced additional locations for the release of ions. The high stress concentration on the neck of the stem in the LDH MoM THA seen in the current study could in theory worsen fretting and corrosion on the head-neck junction.

Abbreviations Used

THA, total hip arthroplasty; MoM, metal-on-metal; MoP, metal-on-polyethylene; LDH, large diameter head; FEA, finite element analysis; PE, polyethylene; CoCr-PE, conventional head polyethylene liner; CoCR-CoCr, large diameter head prosthetic cup; ROI, region of interest.

Declarations

Ethics approval and consent to participate: Not applicable
Consent for publication: Not applicable
Availability of data and material: Not applicable
Competing interests: The authors declare that they have no competing interests
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Authors' contributions: DWC and CLL designed the whole project and performed the finite element analysis. MSL interpreted the data and was a major contributor in writing the manuscript. All authors read and approved the final manuscript.
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References

[1] Bozic KJ, Morshed S, Silverstein MD, Rubash HE, Kahn JG: Use of cost-effectiveness analysis to evaluate new technologies in orthopaedics. The case of alternative bearing surfaces in total hip arthroplasty. *J Bone Joint Surg Am* 2006, 88:706-714.

[2] Keener JD, Callaghan JJ, Goetz DD, Pederson DR, Sullivan PM, Johnston RC: Twentyfive-year results after Charnley total hip arthroplasty in patients less than fifty years old. A concise follow-up of a previous report. *J Bone Joint Surg Am* 2003, 85:1066-1072.

[3] Affatato S, Leardini W, Jedenmalm A, Ruggeri O, Toni A: Larger diameter bearing reduce wear in metal-on-metal hip implants. *Clin Orthop Relat Res* 2006, 456:153-158.

[4] Sikes CV, Lai LP, Schreiber M, Mont MA, Jinnah RH, Seyler TM: Instability after total hip arthroplasty. Treatment with large femoral heads vs constrained liners. *J Arthroplasty* 2008, 23:S59-63.

[5] Bozic KJ, Kurtz S, Lau E, Ong K, Chiu V, Vail TP, Rubash HE, Berry DJ: The epidemiology of bearing surface usage in total hip arthroplasty in the United States. *J Bone Joint Surg Am* 2009, 91:1614-1620.

[6] Dahlstrand H, Stark A, Anissian L, Hailer NP: Elevated serum concentrations of cobalt, chromium, nickel, and manganese after metal-on-metal alloarthroplasty of the hip: A prospective randomized study. *J Arthroplasty* 2009, 24:837-845.

[7] Dorr LD, Wan Z, Longjohn DB, Dubois B, Murken R: Total hip arthroplasty with use of the Metasul metal-on-metal articulation. Four to seven-year results. *J Bone Joint Surg Am* 2000, 82:789-798.

[8] Kretzer IP, Kleinhans JA, Jakubowitz E, Thomsen M, Heisel C: A meta-analysis of design- and manufacturing-related parameters influencing the wear behavior of metal-onmetal hip joint replacements. *J Orthop Res* 2009, 27:1473-1480.

[9] Lombardi AV Jr., Mallory TH, Cuckler JM, Williams J, Berend KR, Smith TM: Midterm results of a polyethylene-free metal-on-metal articulation. *J Arthroplasty* 2004, 19:S42-S47.

[10] MacDonald SJ, McCalden RW, Chess DG, Bourne RB, Rorabeck CH, Cleland D, Lueng F: Metal-on-metal versus polyethylene in hip arthroplasty: A randomized clinical trial. *Clin Orthop Relat Res* 2003, 406:282-296.

[11] Naudie D, Roeder CP, Parvizi J, Berry DJ, Eggli S, Busato A: Metal-on-metal versus metal-on-polyethylene bearings in total hip arthroplasty. A matched case-control study. *J Arthroplasty* 2004, 19:S35-41.

[12] Sauvé P, Mountney J, Khan T, DeBeer J, Higgins B, Grover M: Metal ion levels after metal-on-metal ring total hip replacement. A 30-year follow-up study. *J Bone Joint Surg Br* 2007, 89:586-590.

[13] Amstutz H, Grigoris P: Metal on metal bearings in hip arthroplasty. *Clin Orthop Relat Res* 1996, 329:11-34.

[14] Amstutz H, Beaulé PE, Dorey FJ, LeDuff MJ, Campbell PA, Gruen TA: Metal-on-metal hybrid surface arthroplasty: Two to six-year follow-up study. *J Bone Joint Surg Am* 2004, 86:28-39.

[15] Yue EJ, Cabanela ME, Duffy GP, Heckman MG, O'Connor MI: Hip resurfacing arthroplasty. Risk factors for failure over 25 years. *Clin Orthop Relat Res* 2009, 467:992-999.

[16] Healy WL, Iorio R, Lemos MJ: Athletic activity after joint replacement. *Am J Sports Med* 2001, 29:377-388.

[17] Healy WL, Sharma S, Schwartz B, Iorio R: Athletic activity after total joint arthroplasty. *J Bone Joint Surg Am* 2008, 90:2245-2252.

[18] Fialho JC, Fernandes PR, Eça L, Folgado J: Computational hip joint simulator for wear and heat generation. *J Biomech* 2007, 40:2358-2366.

[19] Tanino H, Ito H, Higa M, Omizu N, Nishimura I, Matsuda K, Kitamura Y, Matsuno T: Three-dimensional computer-aided design based design sensitivity analysis and shape optimization of the stem using adaptive p-method. *J Biomech* 2006, 39:1948-1953.

[20] Pettersen SH, Wik TS, Skallerud B: Subject specific finite element analysis of implant stability for a cementless femoral stem. *Clin Biomech* 2009, 24:480-487.

[21] Scholes SC, Unsworth A, Goldsmith AAJ: A frictional study of total hip joint replacements. *Phys Med Biol* 2000, 45:3721-3735.

[22] Wang CJ, Yettram AL, Yao MS, Procter P: Finite element analysis of a gamma nail within

a fracture femur. *Med Eng Phys* 1998, 20:677-683.

[23] Gulgin H, Armstrong C, Gribble P: Hip rotational velocities during the full golf swing. *J Sport Sci Med* 2009, 8:296-299.

[24] Myers J, Lephart S, Tsai Y, Sell T, Smoliga J, Jolly J: The role of upper torso and pelvis rotation in driving performance during the golf swing. *J Sport Sci* 2008, 26:181-188.

[25] Gruen TA, McNeice GM, Amstutz HC: Modes of failure of cemented stem-type femoral components. *Clin Orthop Relat Res* 1979, 141:17-27.

[26] Mancuso CA, Graziano S, Briskie LM, Peterson MG, Pellicci PM, Salvati EA, Sculco TP: Randomized trials to modify patients' preoperative expectations of hip and knee arthroplasties. *Clin Orthop Relat Res* 2008, 466:434-431.

[27] Sechriest VF II, Kyle RF, Marek DJ, Spates JD, Saleh KJ, Kuskowski M: Activity level in young patients with primary total hip arthoplasty. A 5-year mninmum follow-up. *J Arthroplasty* 2007, 22:39-47.

[28] Naal FD, Maffiuletti NA, Munzinger U, Hersche O: Sports after hip resurfacing arthroplasty. *Am J Sports Med* 2007, 35:705-711.

[29] Narvani AA, Tsiridis E, Nwaboku HC, Bajekal RA: Sporting activity following Birmingham hip resurfacing. *Int J Sports Med* 2006, 27:505-507.

[30] Ong KL, Rundell S, Liepins I, Laurent R, Markel D, Kurtz SM: Biomechanical modeling of acetabular component polyethylene stresses, fracture risk, and wear rate following press-fit implantation. *J Orthop Res* 2009, 27:1467-1472.

[31] Garbuz DS, Tanzer M, Greidanus NV, Masri BA, Duncan CP: Metal-on-metal hip resurfacing versus large-diameter head metal-on-metal total hip arthroplasty. A randomized clinical trial. *Clin Orthop Relat Res* 2009 Aug 21.

[32] 32. Kendrick II JB, Noble PC, Tullos HS: Distal stem design and the torsional stability of cementless femoral stems. *J Arthroplasty* 1995, 10:463-469.

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