Finite Element Analysis of Mechanical Stability of Ceramic Acetabular Components and Evaluation of ROM in Articulating Hip Joints*

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Abstract
Ceramic articualr surfaces are now widely used as total hip replacements for young and active patients. However, while the excellent tribological properties and high biocompatibility of ceramic articular surfaces prevent loosening and osteolysis, their high stiffness and low ductility occasionally result in ceramic surface fractures. Therefore, this study investigated the effect of varying the sizes of the acetabular components on the mechanical stability. Three femoral head models and 27 acetabular cup models were designed following three size parameters: ball head diameter (28, 32, and 36 mm), acetabular cup thickness (3, 4, and 5 mm), and liner thickness (9, 10, and 11 mm). For all these models, the mechanical stability was evaluated using 3D finite element analyses. Plus, the motion of the 3 femoral head models was measured in six directions using a motion study. The results showed that the maximum stress was decreased when increasing the sizes of the cup, liner, and femoral head, where the 36 mm ball head, 5 mm cup, and 11 mm liner showed the lowest maximum stress, while the 36 mm femoral head exhibited the largest range of motion. The acetabular cup stability was also shown to be affected by the stiffness of the components, where increasing the head size or thickness of the cup and liner increased the component stiffness and range of motion. Thus, the mechanical simulation demonstrated that increasing the size of the acetabular components decreased the ceramic surface stress and risk of impingement.

Key words: Total Hip Replacement, Ceramic Articular Surface, Finite Element, Mechanical Stability, Range of Motion

1. Introduction
A total hip replacement (THR) is a widely used treatment for conditions such as degenerative arthritis, rheumatoid arthritis, avascular necrosis, hip joint infection, and congenital malformation[2]. Recognized as an effective procedure to ease pain and recover the range of motion (ROM), the THR surgery eliminates the damaged femoral head and an artificial hip implant made of metal or ceramic is inserted in the femoral region. The
artificial hip joint is composed of a socket-shaped acetabular cup that is directly joined to the pelvis, a stem inserted inside the bone-marrow cavity in the femur, a sphere-shaped ball head that substitutes for the femoral head, and a liner that connects the ball head with the acetabular cup.

The bearing surface between the femoral head and the liner was first made of polyethylene. However, polyethylene particles generated by friction at both articular interfaces became recurrent debris components, inducing loosening and osteolysis. Thus, to solve the problems of polyethylene wear, ceramic-ceramic or metal-metal bearing surfaces took over as the preferred choices for artificial hip joints. Ceramic-ceramic bearing surfaces have continued to produce satisfying results since they were first introduced in 1970 by Bouton. In addition to a very low coefficient of friction compared to a polyethylene bearing surface, the wear rate of ceramic is 4000 times less than that of polyethylene, attracting young and active patients. Plus, a ceramic-ceramic bearing produces a less intense foreign-body inflammatory response around ceramic joints, making it the most insertable biomaterial. Nonetheless, despite its theoretically improved fracture toughness over polyethylene, the rate of catastrophic ceramic-bearing-surface failure is still relatively high at the time of short-term follow-up, as the high stiffness and low ductility of ceramic articular surfaces increase the risk of fracture due to loosening and cracks. Also, an insufficient ROM can lead to impingement between the femoral neck and the rim of the cup, and continuous repetition of impingement can cause dislocation and material failure.

In response to these failures, evaluating the mechanical stability has become one of the key factors for implant designs. The mechanical stability of the acetabular cup is related to the stiffness of the components, which in turn is affected by the geometrical values of the stem neck design, femoral head size, cup type, cup sizes, and head-to-neck ratios. Many studies have already tried to evaluate the effects of the component designs. Burroughs focused on femoral head dislocation to examine the effects of the component sizes. Sychterz investigated the polyethylene liner wear. Hsu used a finite element analysis to evaluate the micromotion of the cup with different sizes of the acetabular component. Furmanski estimated the effects of various rim designs. However, relatively few studies have focused on how the acetabular component sizes affect the mechanical stability and ROM. Accordingly, this study evaluated the mechanical stability of the acetabular cup when varying the component sizes, and compared the ROM when changing the ball head diameter.

2. Methods

2.1 Mechanical stability of acetabular components using Finite element analysis

The 3-dimensional (3D) nonlinear FE model of the femoral head consisted of the stem, ball head, and support. The ball head diameter was designed with three different sizes: 28, 32, and 36 mm. A taper-type fixation was used between the ball head and the stem. Each component was generated as a solid model using CAD software (SolidWorks 2009; Dassault System, Concord, MA), and exported into Hypermesh version 8.0 (Altair HyperWorks; Altair Engineering, Inc., Troy, MI) to generate an FE mesh. The FE analysis was conducted using commercially available software (ABAQUS 6.9; Hibbit, Karlsson and Sorenson, Inc., Providence, RI). Figure 1 shows the complete FE model, consisting of an 8-node brick element, where the number of elements used in the stem, ball head, and support was about 20,000–30,000, 42,000–80,000, and 7000, respectively.
Fig. 1 Static load and boundary conditions of femoral head models. (a) ISO 7206-5 (Implants for surgery -- Partial and total hip joint prostheses -- Part 5: Determination of resistance to static load of head and neck region of stemmed femoral components) (b) Finite element analysis mesh of femoral components containing 3D 8-node brick elements. Load was applied to stem (46 kN), while support was constrained.

The material properties were selected from literature (Table 1), and the materials were assumed to be isotropic and homogeneous. The stem-head and head-support interfaces were modeled with surface-to-surface contact and a friction coefficient of 0.35 and 0.3, respectively\(^{(10)}\). The load condition for the static test of the ceramic hip implant was 46 kN, given by FDA Guidance (Guidance document for preparation of premarket notifications for ceramic ball hip systems)\(^{(11)}\), while the maximum load on the hip joint during physiological activities was 2.5 kN\(^{(4)}\). The loading conditions used in this study were about 21 times greater than normal. The boundary condition for the static test was against a 100° cone (ISO 7206-5 standard)\(^{(23)}\). An axial load was applied at the center of the stem and the ball head was sustained against an axially retained 100° support.

Table 1. Material properties used in femoral head FE models

<table>
<thead>
<tr>
<th>Part</th>
<th>Material</th>
<th>Elastic modulus (MPa)</th>
<th>Poisson’s ratio, $\nu$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ball head</td>
<td>Al2O3</td>
<td>380,000</td>
<td>0.245</td>
</tr>
<tr>
<td>Stem</td>
<td>TiAl6V4</td>
<td>105,000</td>
<td>0.3</td>
</tr>
<tr>
<td>Support</td>
<td>PMMA</td>
<td>210,000</td>
<td>0.3</td>
</tr>
</tbody>
</table>

To evaluate the effects of the component sizes on the mechanical stability, an acetabular cup with a thickness of 3, 4, and 5 mm, liner with a thickness of 9, 10, and 11 mm, and ball head with a diameter of 28, 32, and 36 mm were generated. Thus, a total of 27 models with different sizes were designed using SolidWorks, and FE models generated using Hypermesh. The number of elements used in the ball head, liner, acetabular cup, and support was about 7,000–12,000, 10,000–16,000, 7,000–16,000, and 20,000, respectively. The contact coefficient of friction between the ball head-liner and liner-acetabular cup
interfaces was 0.083 and 0.35, respectively\(^{(12)}\). The liner-support interface was assumed to be bonded (Tie). All the materials considered were assumed to be linear elastic, isotropic, and homogeneous and the material properties are given in Table 2. The loading and boundary conditions for the acetabular cup were as same as those for the femoral head models.

![Finite element models of acetabular components](image)

**Fig. 2** Finite element models of acetabular components. (a) Acetabular components consisted of ball head, liner, acetabular cup, and support. (b) Load was applied to center of ball head (46 kN), while support was constrained.

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<td>0.23</td>
</tr>
<tr>
<td>Liner</td>
<td>Al2O3</td>
<td>380,000</td>
<td>0.23</td>
</tr>
<tr>
<td>Acetabular cup</td>
<td>TiAl6v4</td>
<td>105,000</td>
<td>0.3</td>
</tr>
<tr>
<td>Support</td>
<td>PMMA</td>
<td>2,000</td>
<td>0.33</td>
</tr>
</tbody>
</table>

### 2.2 Measurement of range of motion when articulating hip joints

The ROM is influenced by both the ball head diameter and the neck-shaft angle. The stem was selected (neck-shaft angle 125°) from Corentec. Co., Ltd. considering the neck-shaft angle of a normal person. The diameters of the ball head were 28, 32, and 36 mm. The basic position of the acetabular cup, which defines the location of the anatomical hip joint, was a 45° inclination and 15° antversion, while the stem had a 6° adduction and 10° antetorsion. When changing the ball head diameter, six motions of the hip joint were measured: flexion, extension, abduction, adduction, internal rotation, and external rotation. The ROM was measured using the Motion Study function of SolidWorks and the results compared with the average ROM of a normal person\(^{(27)}\).
Inclination 45° Anteversion 15° Antetorsion 10° Adduction 6° Internal / External rotation Flexion / Extension Abduction / Adduction

Fig. 3 Anatomical position of hip implant (45° inclination, 15° anteversion, 6° adduction, and 10° antetorsion) and six motion directions (flexion, extension, abduction, adduction, internal rotation, and external rotation)

Table 3. Average ROM of normal adult male*

<table>
<thead>
<tr>
<th>Motion type</th>
<th>Flexion</th>
<th>Extension</th>
<th>Abduction</th>
<th>Adduction</th>
<th>Internal rotation</th>
<th>External rotation</th>
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<tbody>
<tr>
<td>Motion angle</td>
<td>113°</td>
<td>28°</td>
<td>48°</td>
<td>31°</td>
<td>45°</td>
<td>45°</td>
</tr>
</tbody>
</table>


3. Results

3.1 Stability according to femoral head size

In general, excessive stress enhances the possibility of failure. Also, the major structural damage to the ball head is caused by tensile hoop stress in the taper region. In this study, the stress distribution was analyzed for the principal stress in the taper region.

The FE analysis results when changing the size of the ball head were evaluated, and the change of stress described as a percent change according to the ball head (Fig. 5). When increasing the ball head diameter, the maximum stress decreased (Fig. 4). When compared to the 28 mm model, the stress on the 32 mm ball head was about 16% lower, while the stress on the 36 mm ball head was about 31% lower. The highest maximum stress was measured with the 28 mm model (395 MPa), which was still 32% lower than the failure strength (580 MPa). Thus, all the ball head models were confirmed as mechanically safe.

Fig. 4 Distribution of tensile hoop stress in ball head for static test under axial loading (46 kN) against 100° support. Ball head diameters: (a) 28 mm (b) 32 mm and (c) 36 mm
Fig. 5 Maximum principal stress with different sizes of ball head. Increasing the diameter of the ball head reduced the maximum stress value. Maximum principal stress for each ball head: 395 MPa (28 mm), 332 MPa (32 mm), and 273 MPa (36 mm).

3.2 Stability according to cup and liner thickness

The change of stress is described as a percent change according to the acetabular cup and liner thickness (Fig. 6). For all three sizes of ball head, the maximum stress decreased as the liner and acetabular cup thickness increased. When changing the acetabular cup thickness, the maximum stress occurred in the case of the 28 mm ball head and 3 mm acetabular cup. Meanwhile, when changing the liner thickness, the maximum stress occurred in the case of the 28 mm ball head and 9 mm liner. Thus, the most unstable model was the 28 mm ball head with a 3 mm acetabular cup and 9 mm liner. In contrast, the maximum stress was almost 37% lower for the most stable case: the 36 mm ball head with a 5 mm acetabular cup and 11 mm liner.

Fig. 6 Comparison of maximum principal stress with different liners. (a) Maximum stress with liner thickness of 9, 10, and 11 mm and (b) maximum stress with acetabular cup thickness of 3, 4, and 5 mm
3.3 Comparison of ROM according to femoral head size

The average ROM of a normal person is presented in Table 3. The ROM is generally influenced by sex and age, plus the maximum movement range of a hip joint differs between the active-range-of-motion (AROM) and the passive-range-of-motion (PROM). Thus, since THRs are mostly provided for adults, the results are compared to the PROM values of an adult male over 19 years old\cite{13,14,15,16}.

The results of the virtual movement of the hip joint are shown in Figure 7. The ROM tended to increase with a larger ball head diameter. The smallest ROM model was produced with the 28 mm ball head. When compared to the average ROM of a normal person, the movement values with the 28 mm ball head were increased as follows; 0.71% flexion, 7.35% extension, 3.77% abduction, 1.45% adduction, 5.45% internal rotation, and 3.42% external rotation.

![Comparison of ROM according to ball head diameter.](image)

**Fig 7.** Comparison of ROM according to ball head diameter.

**If L1 < L2 then δ1 < δ2**

![Relation between different ball head diameters and ROM.](image)

**Fig. 8.** Relation between different ball head diameters and ROM. ROM was greater with larger ball head than smaller ball head.

4. Discussion

Ceramic articular surfaces have a low coefficient of friction and high biocompatibility. Yet, ceramic surface failures continue to occur due to geometrical defects and limitations of
the ROM. Therefore, this study evaluated the mechanical stability and risk of impingement according to the size of the acetabular components using a finite element analysis.

The simulation results revealed the stability of the ball head and liner in relation to the maximum principal stress. A taper-type ball head was generated, which allows the penetration depth of the stem to be adjusted by changing the stem head diameter in the taper-type ball head. The head offset is also modifiable by changing the stem head diameter, making it more anatomically suited to each patient. However, the drawback is that the self-locking conical taper transforms compression stresses into bending stresses in the dome part and tangential (hoop) stresses in the rim zone of the ball head. While the stress concentration and distribution vary depending on the surface roughness and friction, most of the stress concentration and damage occur in the taper, since the ball head and stem head experience friction-contact. Meanwhile, the stress distribution occurs axial-symmetrically in a tangential direction, as shown by Weisse\textsuperscript{109} and Higuchi\textsuperscript{177}. Cracks and fractures of the ball head around the taper have also been shown clinically by Higuchi\textsuperscript{177} and Arenas\textsuperscript{18}. 

As stress occurs around the taper according to the stiffness, contact area, and friction coefficient of the ball head, increasing the ball head diameter would be expected to increase the structural stiffness. Thus, a ball head with a larger diameter would increase the acetabular cup stability. The friction force is expressed as $F=\mu N$ (where $F$ refers to the force loaded on the stem head, $\mu$ indicates the friction coefficient at the interface, and $N$ refers to the normal force loaded on the ball head from inside to outside in a tangential direction). Therefore, increasing the coefficient of friction causes less stress loaded on the ball head, resulting in a high fixation between the ball head and the stem. This also minimizes the stress and improves the fixation. Thus, it is expected that the use of friction models with a high coefficient could be clinically safer.

For the mechanical stability of the acetabular cup, Hsu\textsuperscript{19} reported that the safety of the acetabular cup depends on the stiffness of the structure and distance between the loading point of the ball head and the fixation screw. Also, increasing the thickness of the acetabular cup has most influence on its stability. Therefore, Hsu\textsuperscript{19} recommended choosing a thick acetabular cup to enhance its stability. The results analyzed in this study showed that increasing the size of the liner, cup and ball head increased the structural stiffness and reduced the maximum stress. The maximum stress that occurred in all the models was under 60 MPa, which is only about 10% of the yield stress of 580 MPa. Furthermore, minimizing the stress reduces the possibility of damage with long-term use.

For further optimization of the mechanical stability of the acetabular cup, the ROM was measured. Fractures of ceramic articular surfaces also occur due to a limited range of movement, which in turn is affected by the ball head diameter, inclination, and anteversion. In actual hip joint movements, the physiological effects of tissue, muscles, and ligaments also need to be considered. Therefore, actual movement prediction is limited without considering physiological effects, flexibility, and individual characteristics. Nonetheless, despite the difference in the ROM for an actual patient, the ROM change according on the ball head diameter can be objectively measured. As such, increasing the ball head diameter was found to increase the gyration angle of the ball head and the ROM (Fig. 8). Therefore, using a ball head with larger diameter can reduce fractures due to impingement.

The limitation of the current study was that only one type of load was considered based on FDA Guidance\textsuperscript{11} standards. However, when an acetabular cup is inserted in a human body, a variety of physiological loads are also applied to the joint. Thus, further mechanical analyses under various load conditions are needed to minimize the failures of a THR.
5. Conclusion

The current study evaluated the stability and ROM according to the size of the acetabular components using finite element analyses. Increasing the diameter of the ball head or thickness of the liner and acetabular cup reduced the maximum stress. Also, increasing the diameter of the ball head increased the ROM of the ball head. For the acetabulum sizes tested, the acetabular cup stability was enhanced when increasing the ball head diameter and thickness of the liner and cup.

Acknowledgements

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References

(11) Guidance document for the preparation of premarket notifications of ceramic ball hip systems.
(12) Kim, S. Y., Kim, Y. G., Yeo, J. Y., Kim, D. H., and Ihn, J. C., Ceramic-on-Ceramic Bearing Total Hip Athroplasty in Young Patients with Osteonecrosis of Femoral Head,