# A three-dimensional finite element analysis of the human hip

Mohammad Akrami [a*], Kim Craig [a], Mahdieh Dibaj [a], Akbar A. Javadi [a], Abdelmalek Benattayallah [b]

[a] Department of Engineering, College of Engineering, Mathematics, and Physical Sciences University of Exeter, Exeter, United Kingdom
[b] MR Research Centre, St. Luke's Campus, University of Exeter, Exeter, United Kingdom

*Corresponding Author
Dr Mohammad Akrami
Lecturer in Mechanical Engineering
College of Engineering, Mathematics, and Physical Sciences
University of Exeter
Telephone: 01392 724542
Address: Room 271, Harrison Building, North Park Road, Exeter, Devon, EX4 4QF

Keywords: Finite Element Analysis, Hip biomechanics

Word count:
Submitted as an *Original Article*

**Abstract**

A three-dimensional hip model was created from the MRI scans of one human subject based on constructing the entire pelvis and femur. The ball and socket joint was modelled between the hip's acetabulum and the femoral head to analyse the multiaxial loads applied in the hip joint. The three key ligaments that reinforce the external surface of the hip to help to stabilise the joint were also modelled which are the iliofemoral , the pubofemoral and ischiofemoral ligaments. Each of these ligaments wraps around the joint connection to form a seal over the synovial membrane, a line of attachment around the head of the femur. This model was tested for different loading and boundary conditions to analyse their sensitivities on the cortical and cancellous tissues of the human hip bones. The outcomes of a one-legged stance finite element analysis revealed that the maximum of 0.056 mm displacement occurred. The stress distribution varied across the model which the majority occurring in the cortical femur and dissipating through the cartilage. The maximum stress value occurring in the joint was 110.1 MPa, which appeared at the free end of the proximal femur. This developed finite element model was validated against the literature data to be used as an asset for further research in investigating new methods of total hip arthroplasty, to minimise the recurrence of dislocations and discomfort in the hip joint, as well as increasing the range of movement available to a patient after surgery.

## 1. Introduction

The hip joint is one of the most load-bearing joints in the human body and consequently must undergo a large amount of use over a human lifetime. During frequent use, the hip joint can wear down and start to erode; meaning the hip may have to be replaced with a prosthesis. In fact, instances of total hip arthroplasty are performed have increased 40.9% between 1991 and 2005 (1); therefore, the need for more efficient and effective surgery is needed more than ever. As technological and scientific advancements are made, methods of total hip arthroplasty are evolving to improve the range of movement after surgeries and minimise the risk of prosthesis dislocation.

Therefore the biomechanical behaviour of the hip joint needs to get investigated to have a better understanding of the related treatments and how different tissues work. Accessing the internal structure of the hip joint in-vivo is impossible to monitor how different segments work and sometimes for measuring the functional behaviours, the soft tissues need to be cut during or before the surgery. Therefore, computational methods are utilised to generate detailed results and analyse the biomechanics of such complex musculoskeletal structure.

There are a number of studies that already focus on finite element modelling of the hip, however, many of them analyse hip prostheses (2, 3), or alternatively focus on individual parts of the body like the pelvis (4, 5) or femur (6, 7) without considering their bio-realistic interactions. This main novelty of the current research is modelling both the femur and pelvis in the hip joint besides the cartilage and ligaments. Some of the previous studies modelled the pelvis in half, which the effects of this assumption in terms of the aesthetics have not been investigated in detail. Huiskes & Chao (8) showed an alternative method of finite element analysis on a 2D scale, through superposition of 2D medical images. They analysed the femur under the exertion of a unit force of 1 N applied ellipsoidally over the acetabulum area of the hip joint, creating a simulation of a one-legged stance. This paper measured the critical fracture load of the femur and would be a reasonable study for a project which is limited on resources, like computing power or medical imaging. Keyak et al. (9) used finite element modelling to predict the critical values of femoral load fracture by creating 3D models of 18 pairs of femoral from cadavers.

In order to use the computational models for analysing the bio-realistic behaviours, loading and boundary conditions need to be applied precisely. Boundary conditions are proven to present different results depending on the location where the hip joint is fixed. Watson at al. (4), showed that the location of the boundary condition significantly changes the stress distribution and the magnitude of stress across the bones. A 2007 study by Phillips et al. (5) modelled and analysed the hip by setting muscular and ligamentous boundary conditions. This caused the stress distributions around the hip to appear significantly different from that of ordinary fixed boundary conditions. Throughout the literature, mostly the pubic symphysis is fixed as the main boundary condition. Another study compared deformation under loading by modelling the pubic symphysis as both a rigid and deformable boundary condition (7). Both approaches showed similar pressures and stresses, however, the rigid model showed high stresses in the cancellous sections of the inferior pubis bone and the superior dome. Phillips, et al. (5) also considered the different options when modelling the pubic symphysis, but ultimately decided that rigid modelling is more practical when focussing on the strength of the hip joint itself in order to decrease the computational cost. There were several assumptions made in previous studies that should be assessed, the first, a common one, was to assume the perfect spherical shape of the femoral head and the corresponding acetabular socket (5). However, this assumption is disputed and unrealistic; the femur head is known to have a concave depression within it named the fovea capitis, which is the location where one of the ligaments in the hip attaches to the femur. Also, previous studies neglect the effect of changed in the shape of the femoral head and cartilage thickness due to joint degeneration which may have siginificant effect on stress distribution (10). In addition to this, several papers assume unique values for the thickness of cartilage and cancellous bones (11, 12). The frictionless surface-to-surface interaction was applied in studies in which the effects of the frictionless assumption has not been investigated in detail (13). Therefore, the main purpose of this study is to determine the biomechanical properties of the hip joint during different types of loading and how these will cause the functional parameters of the cortical and cancellous bones of the hip joint for increase-decreasing the loads. This can help to understand the roles of these bones in everyday locomotion patterns.

## 2. Methods and materials

### 2.1. Medical image analysis

Three-dimensional versatile geometries of the current hip model were reconstructed from the medical MRI data. The MRI scanning was performed on a 1.5 Tesla Phillips Intera system using T1 3D Gradient Echo sequence (TR/TE = 57 ms/21 ms, flip angle 90°, 360º slices and spatial resolution with voxels size of $0.5 \times 0.5 \times 0.5$ mm$^3$). The geometries were captured by scanning the right hip of a healthy female subject (20 years old, with no history of upper limb injuries or hand abnormalities) in the neutral position with a 2 mm slice interval. All the images were segmented manually to reconstruct the boundaries of bones and soft tissues using ScanIP software (Synopsys, Mountain View, USA). Automatic smoothing was carried out in order to omit the rough surfaces and sharp edges; thereby a surface made from the average nodal positions was produced, which more accurately represented bone geometry. Cancellous and cortical tissues of the femur and the whole structure of the pelvis have been modelled to have a three-dimensional structure of its anatomy (14). In order to model the articulations between the two bones, cartilage layers were designed. There are two key areas of cartilage in the hip, both found in the articular joint section; the first is on the head of the femur and the second in the pelvic acetabulum. Cartilage that appears on articular surfaces like the hip is hyaline cartilage; which providing a smooth lubricated surface for joints as well as supporting soft tissues. In this study, the cartilage layers on the surface of the acetabulum and femoral head were reconstructed based on the MRI data to maintain their bio realistic structure. So far, based on the knowledge of authors, there is no study which has designed the cartilage topological structures for the hip joint based on the bio-realistic image-driven data.

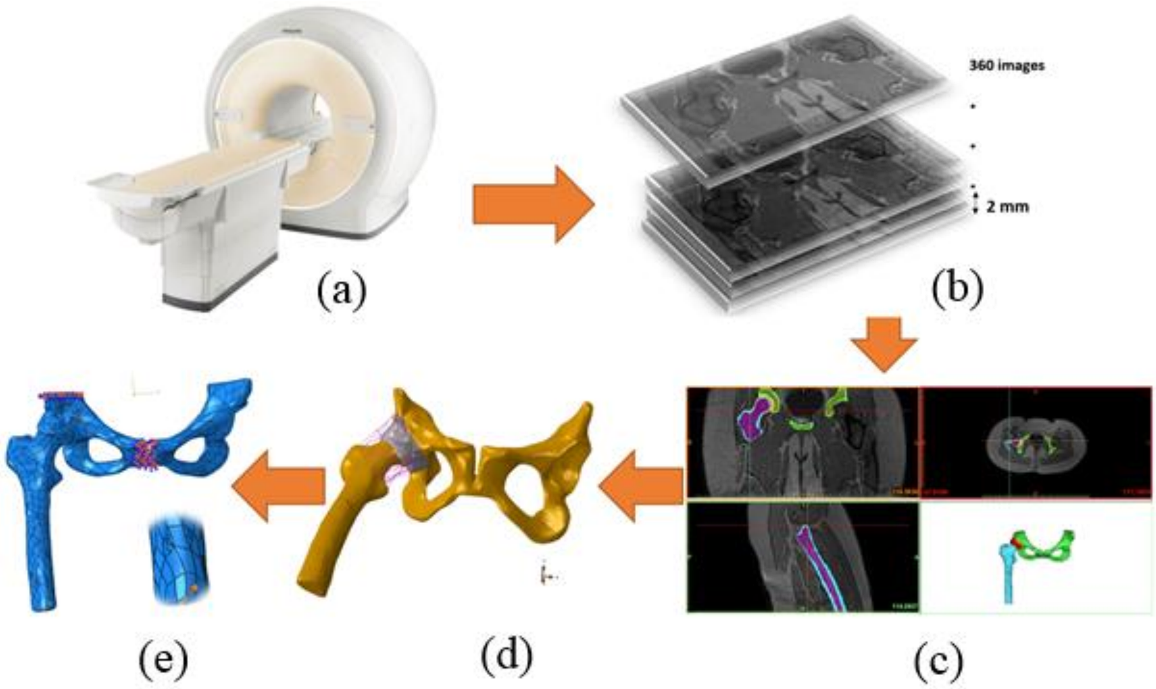


**Figure 1** Developing process of the human hip model: (a) Scanning the subject (b) Generating DICOM files (c) medical image processing to segment the bones and cartilages (d) adding ligaments to the model in SOLIDWORKS (e) applying loads and boundary conditions to the finite element model in ABAQUS

### 2.2. Finite element modelling

After the medical image data were processed using medical image processor software (ScanIP), the STL format files were transferred to Solidworks software (Dassault Systèmes, SolidWorks Corp., USA) to assign the boundary surfaces and assembling the solid models for the bones and the designed soft tissues based on Bio-CAD Image-Based technique (15). The whole pelvis was designed besides the femur which was designed partially to analyse the hip joint. These two bony tissues were divided into cortical and cancellous to have the bio-realistic representations of this anatomical structure. The material properties derived from literature by distinguishing the cortical and cancellous properties for the hard tissues (See Table 1). For the ligaments, the stiffness values are assigned in detail from the literature (See Table 2). Each springs' stiffness was calculated using the parallel springs rule. The total stiffness of a group of parallel springs is the sum of each springs' individual stiffness. The material properties and element types used for modelling different components of the hand complex are listed in detail in Table 1. The mesh sensitivity study was applied through the convergence analysis by the gradual increase of the mesh quality until deviations in the evaluated stresses reached <5% (16).

**Table 1** Material properties and element types of the human hip finite element model

| Components | Materials | Element types | Young's modulus (MPa) | Poisson's ratio | References |
|---|---|---|---|---|---|
| Cortical bone | Solid, linear elastic | Tetrahedral | 17000 | 0.3 | (4, 11, 13, 17-19) |
| Cancellous bone | Solid, linear elastic | Tetrahedral | 70 | 0.2 | (13) |
| Cartilage | Solid, linear elastic | Tetrahedral | 15 | 0.45 | (7, 13) |

**Table 2** Ligament properties for the hip joint ligaments (13)

| Ligament | Ligament Stiffness (N/m) | Number of Spring elements |
|---|---|---|
| Teres | 68000 | 1 |
| Ischiofemoral | 39600 | 10 |
| Pubofemoral | 36900 | 6 |
| Inferior Iliofemoral | 100700 | 4 |
| Superior Iliofemoral | 97800 | 4 |

The cortexes and pubic symphysis were fixed throughout the analyses. A force of 600 N applied perpendicularly to the acetabulum at the end of the femur length to recreate a one-legged stance (5). Constraints and interactions were added to the model to ensure that each of the components in the model reacts with each other accurately. A frictionless surface-to-surface interaction was created between the surfaces of the cartilage and bones; between the hip and femur cartilage free movement was allowed, however between bone and cartilage for both the hip and the femur a “tie” constraint was used to fix the surfaces together (20).

## 3. Results

Loading values were captured from literature (5, 11) in order to compare the results for the validation process. Furthermore, the addition of cartilage and ligaments to the model altered the results away from those previously presented, especially due to different material properties or geometry. Higher stresses on the femur within the loading plane were predicted (see figure 2), as these were expected due to the bending applied to the femur in this direction. The maximum stress occurred in the cortical femur was close to 75 MPa, which is similar to the reported 70 MPa (5**).**

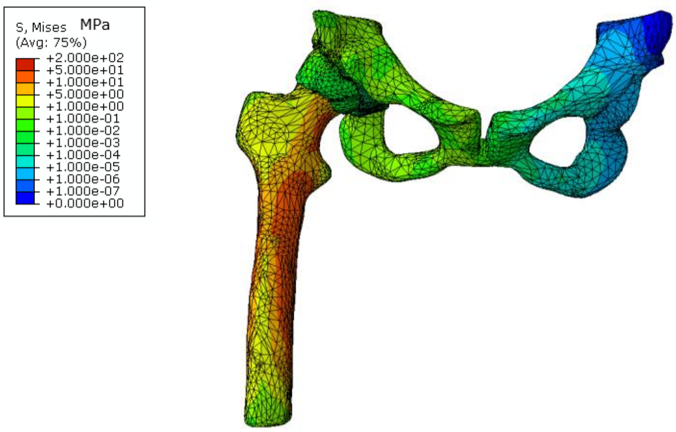


**Figure 2:** Stress distribution within the bones and cartilage layers in normal standing condition

It can be interpreted that most of the weight from the upper body is supported in the legs rather than the pelvis and lower spine. The cortical proximal femur sustained the highest stress values of any component in the hip joint. Both cancellous bones show lower values of stresses as it was expected (21). The cancellous pelvic bone had a maximum stress of

0.1827 MPa, while the cancellous femur had a maximum stress of 12.42 MPa which shows the role of the femur in sustaining the loads before transmitting to the knee and ankle joints. The maximum stress occurring in the cancellous femur was on the femoral head; while the cortical femur had a high amount of stress at this point as well since this was where the Teres ligament was located. The cortical pelvic bone presented a maximum stress of 9.581 MPa around the acetabulum. Regarding the cartilage, they sustain most of the stresses in the contact region. The femur and pelvic cartilage each had a maximum stress of 31.95 MPa and 13.7 MPa, respectively. The stress plot for the cartilage shows medium to high stresses in the direction of the femur's rotation, while other areas showed arbitrary locations stress (see Figure 3).

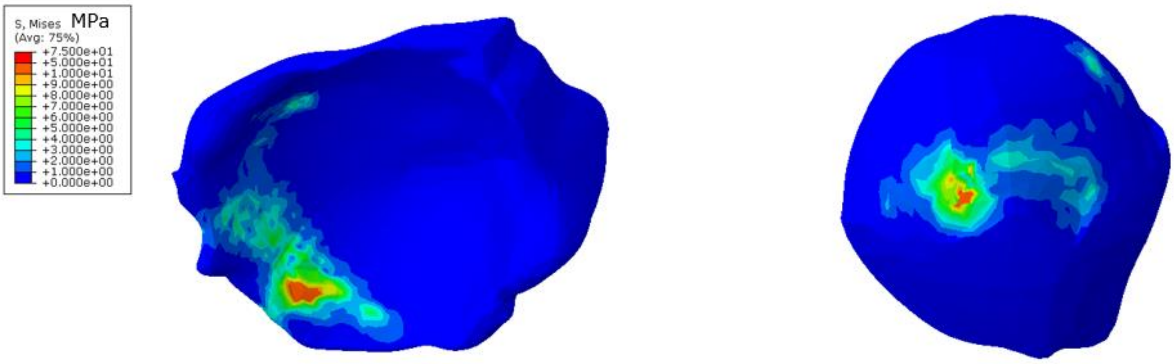


**Figure 3** Stress distributions of the pelvic cartilage (left) and femur cartilage (right)

The dissipation of stress throughout the joint from the point of contact revealed that the stress from a standing load is not very widespread, since the majority of the force was distributed on the femoral head, revealing that frequent use and high loads concentrated on a small part of the body would eventually cause failure risks, especially in the elderly people which the cortical tissues are thinning (22) . This also reveals that a large majority of the hip joint, especially the pelvis, exists for the protection of internal organs rather than for structural integrity in supporting the body. Other loading conditions may present alternative results in the overall structural stresses, for example, a large amount of the pelvis experienced little to no stress, alternative loading and boundary conditions around these areas could produce differing results; however, the hip joint is more commonly under stresses during walking or standing (the current loading), which may explain why falls are more likely to cause a hip dislocation or injury (9) due to higher magnitude and multi-directional forces causing high stresses in commonly low-stress areas, which the hip joint may not be equipped for. The ligaments experienced the majority extension in the plane of rotation of the femur with the maximum strain in the model was 0.009543%. The maximum deformation under standard loading is 0.05648 mm. As expected, the deformation was the smallest close to the centre of the joint rotation and increases away from this point, this is the natural movement of the femur about the joint.

## 4. Discussions

By running the model for 10% and 20% above and below the standard load, the load sensitivity was analysed (See figures 4 and 5). The results show that the strain or deformation changes more than the stress while the loads are increased/decreased. This means most of the excess energy from extra loads can get dissipated in the cartilage layers while the bones are articulating (Figure 4).

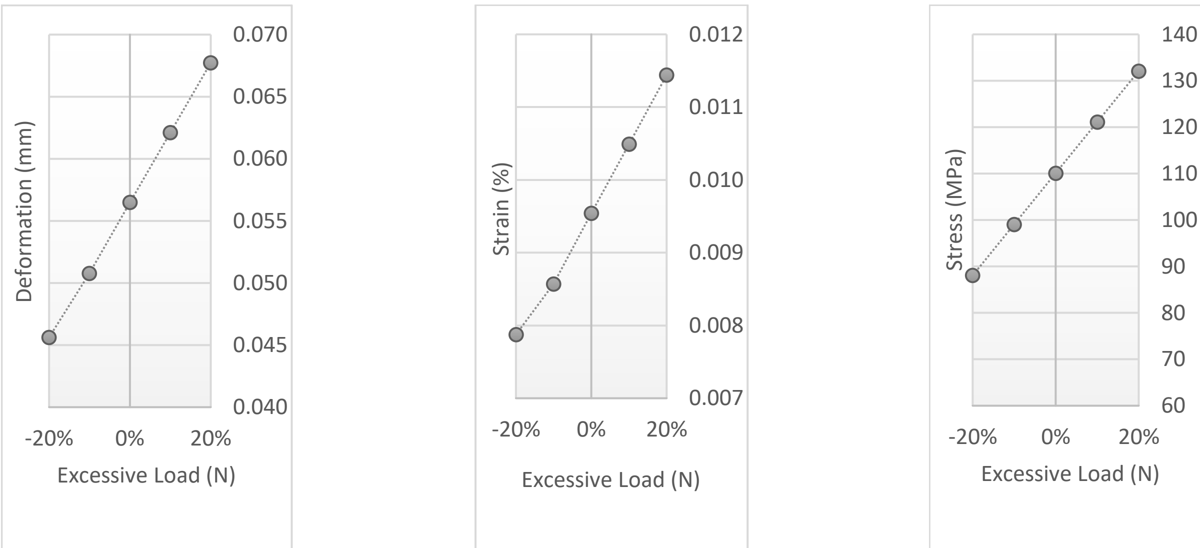


**Figure 4:** Functional parameters of the hip bones for increase-decreasing the loads

When the loads were applied, the stress is not evenly distributed within the different bony tissues (See figure 5). The cortical femoral head sustained more stresses comparing with the spongy cancellous to stabilise the body during different daily activities. While the load is being increased by 10%, excessive stresses are then localised on the femur neck which causes fracture risks as the intertrochanteric areas are releasing some of these stresses and therefore the extra energy is merely sustained by the femoral head. Less stress was observed around the ventrocranical region of the acetabulum and ventrolateral of the head as the cartilage is thicker (23). Though some studies (24, 25) show there is no correlation between aggregate modulus of articular cartilage and its thickness (26), thicker cartilage retains more excessive energy to reduce the effects of sudden shocks. For the normal pressures, the stresses were evenly distributed between the acetabulum and the acetabular cartilage to maintain the existed solidarity between the head of the femur and the acetabulum.

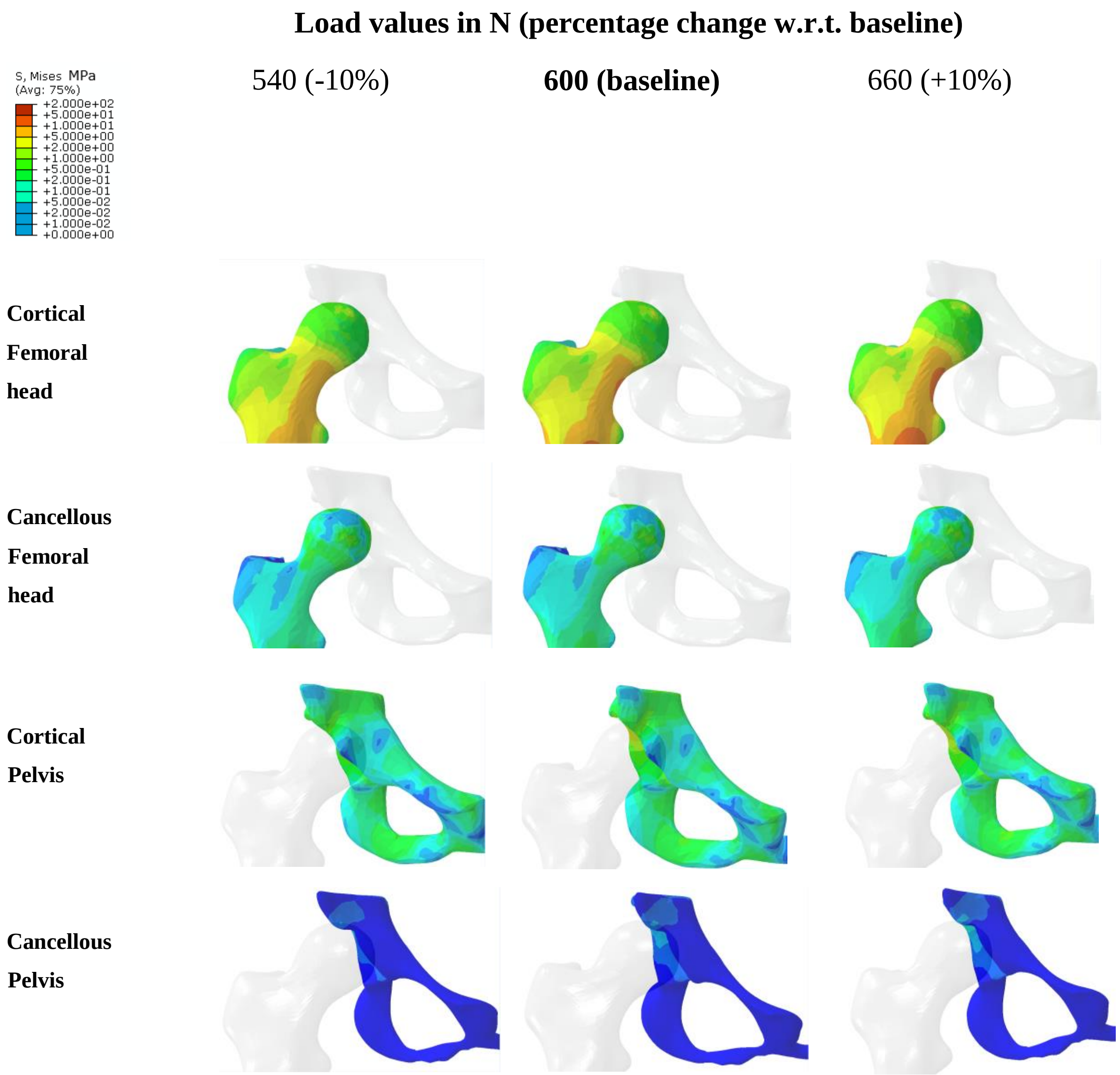


**Figure 5** Stress distributions of the cancellous and cortical femur and pelvis bones by different loading values

## 5. Conclusion

This model predicted how the loads are distributed within the different tissues of the hip bones and interconnected tissues. The model predicted the excessive loads will have more stress concentration around the femur neck which cause can cause fracture in these regions. Less stress was observed around the ventrocranical region of the acetabulum and ventrolateral of the head.

This developed model can be used as an asset to understand the effects of sudden excessive loads lead to the hip dislocations so the stress patterns after such injuries can be compared against the healthy cases, for the surgical or treatment planning. The constructed FE model of the hip can improve our understanding of this major musculoskeletal complex.

Several limitations exist when developing such kinds of computational FE models. Firstly these kinds of computational models are designed on a subject-specific basis; therefore increasing the number of participants would be valuable to complement the results. The second limitation is better to design the ligaments in three dimensions rather than spring elements, however, in this study, the ligaments origin and insertion were enigmatic in the MRI data and therefore simplifications have been made for representing these. Although the model was validated, replacing the ligaments in three-dimensional structures can improve our understanding of their roles.

**Acknowledgement**

The authors would like to thank Christina Doyle and Andrew John Timperley for the knowledge and resources they provided, which helped define the project. Also, we appreciate the “Simpleware software solutions” for providing the ScanIP software, especially Philippe Young for all his support. Also, we thank the anonymous reviewers for their insightful comments and suggestions.

**Competing interests**

None declared

**Funding**

None

**Ethical approval**

Approved by the CEMPS ethics committee, University of Exeter, with the reference number: eEMPS000040 v3.0.

## References

1. Cram P, Lu X, Callaghan JJ, Vaughan-Sarrazin MS, Cai X, Li Y. Long-term trends in hip arthroplasty use and volume. The Journal of arthroplasty. 2012;27(2):278-85. e2.
2. Levadnyi I, Awrejcewicz J, Gubaua JE, Pereira JT. Numerical evaluation of bone remodelling and adaptation considering different hip prosthesis designs. Clinical Biomechanics. 2017;50:122-9.
3. Watanabe Y, Shiba N, Matsuo S, Higuchi F, Tagawa Y, Inoue A. Biomechanical study of the resurfacing hip arthroplasty: finite element analysis of the femoral component. The Journal of arthroplasty. 2000;15(4):505-11.
4. Watson PJ, Dostanpor A, Fagan MJ, Dobson CA. The effect of boundary constraints on finite element modelling of the human pelvis. Medical engineering & physics. 2017;43:48-57.
5. Phillips A, Pankaj P, Howie C, Usmani A, Simpson A. Finite element modelling of the pelvis: inclusion of muscular and ligamentous boundary conditions. Medical engineering & physics. 2007;29(7):739-48.
6. Rüdiger HA, Parvex V, Terrier A. Impact of the femoral head position on moment arms in total hip arthroplasty: a parametric finite element study. The Journal of arthroplasty. 2016;31(3):715-20.
7. Rapperport D, Carter D, Schurman D. Contact finite element stress analysis of the hip joint. Journal of orthopaedic research. 1985;3(4):435-46.
8. Huiskes R, Chao E. A survey of finite element analysis in orthopedic biomechanics: the first decade. Journal of biomechanics. 1983;16(6):385-409.
9. Keyak JH, Rossi SA, Jones KA, Skinner HB. Prediction of femoral fracture load using automated finite element modeling. Journal of biomechanics. 1997;31(2):125-33.
10. Kempson G, Spivey C, Swanson S, Freeman M. Patterns of cartilage stiffness on normal and degenerate human femoral heads. Journal of Biomechanics. 1971;4(6):597-609.
11. Anderson AE, Peters CL, Tuttle BD, Weiss JA. Subject-specific finite element model of the pelvis: development, validation and sensitivity studies. Journal of biomechanical engineering. 2005;127(3):364-73.
12. Henak CR, Ateshian GA, Weiss JA. Finite element prediction of transchondral stress and strain in the human hip. Journal of biomechanical engineering. 2014;136(2):021021.
13. Zou Z, Chávez-Arreola A, Mandal P, Board TN, Alonso-Rasgado T. Optimization of the position of the acetabulum in a ganz periacetabular osteotomy by finite element analysis. Journal of Orthopaedic Research. 2013;31(3):472-9.
14. Starly B, Fang Z, Sun W, Shokoufandeh A, Regli W. Three-dimensional reconstruction for medical-CAD modeling. Computer-Aided Design and Applications. 2005;2(1-4):431-8.
15. Sun W, Starly B, Nam J, Darling A. Bio-CAD modeling and its applications in computer-aided tissue engineering. Computer-aided design. 2005;37(11):1097-114.
16. Akrami M, Qian Z, Zou Z, Howard D, Nester CJ, Ren L. Subject-specific finite element modelling of the human foot complex during walking: sensitivity analysis of

material properties, boundary and loading conditions. Biomechanics and modeling in mechanobiology. 2018;17(2):559-76.

17. Sun J, Yan S, Jiang Y, Wong DW-c, Zhang M, Zeng J, et al. Finite element analysis of the valgus knee joint of an obese child. Biomedical engineering online. 2016;15(2):158.

18. Cruz D J, Rasgado MT A, CG B. Failure Analysis Following Osteochondroplasty for Hip Impingement in Osteoporotic and Non-Osteoporotic Bones. Journal of Osteoporosis and Physical Activity. 2016.

19. Alonso-Rasgado T, Jimenez-Cruz D, Bailey CG, Mandal P, Board T. Changes in the stress in the femoral head neck junction after osteochondroplasty for hip impingement: a finite element study. Journal of Orthopaedic Research. 2012;30(12):1999-2006.

20. Fan N, Chen G. Numerical study of squeaking suppresses for ceramic-on-ceramic hip endoprosthesis. Tribology International. 2012;48:172-81.

21. Bayraktar HH, Morgan EF, Niebur GL, Morris GE, Wong EK, Keaveny TM. Comparison of the elastic and yield properties of human femoral trabecular and cortical bone tissue. Journal of biomechanics. 2004;37(1):27-35.

22. Engelke K, Prevrhal S, Genant HK. Macro-and microimaging of bone architecture. Principles of Bone Biology (Third Edition): Elsevier; 2008. p. 1905-42.

23. Kurrat HJ, Oberländer W. The thickness of the cartilage in the hip joint. Journal of anatomy. 1978;126(Pt 1):145.

24. Athanasiou K, Rosenwasser M, Buckwalter J, Malinin T, Mow V. Interspecies comparisons of in situ intrinsic mechanical properties of distal femoral cartilage. Journal of Orthopaedic Research. 1991;9(3):330-40.

25. Athanasiou K, Agarwal A, Dzida F. Comparative study of the intrinsic mechanical properties of the human acetabular and femoral head cartilage. Journal of Orthopaedic Research. 1994;12(3):340-9.

26. Shepherd D, Seedhom B. Thickness of human articular cartilage in joints of the lower limb. Annals of the rheumatic diseases. 1999;58(1):27-34.