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In Silico of Subject-Specific Progression of Knee Osteoarthritis: A Finite Element Analysis on Swing and Stance Phases

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Abstract

The clinical condition of joint pain and dysfunction induced by joint degeneration, osteoarthritis, affects more people than any other joint illness. Mechanical stress is a major contributor to the onset of osteoarthritis (OA). However, there is a difficulty of achieving direct quantitative measures of tissue behaviours during different grades of osteoarthritis and currently there is a lack of studies that explore the changes seen in cartilage effected by OA during swing and stance phases. Therefore, the purpose of this research is to look at the role of articular cartilage in the development of OA, as well as to evaluate and simulate the biomechanical behaviour of the knee joint under various boundary conditions by segmented knee joints from computed tomography datasets. Mimics software has been used to obtain the 3D model of the knee bones. In addition, the soft tissues were modelled using 3-matic software. Marc.Mentat software was used to correctly simulate the knee OA behaviour during the stance and swing phases for the nonlinear finite element analysis. During the stance and swing phases, the maximum von Mises stress and displacement on the femur, femoral cartilage, tibia, and tibial cartilage were collected for healthy, grade 1 and grade 2 osteoarthritis. The results reveal that when body weight load increased, so did stresses and displacements in articular cartilage and bones. This suggests that being overweight or obese may increase the risk of joint articular cartilage degeneration and osteoarthritis of the knee. In conclusion, the articular cartilage could be in a trouble if excessive forces are exerted towards it.

1. Introduction

Osteoarthritis (OA) is the most common type of medical joint disease and disability in the world, and it is one of the top five causes of disability in elderly people who are not institutionalized. It is a chronic and a debilitating condition characterized by pain, joint inflammation, and joint stiffness. OA can affect any joint in the body, including the hands, knees, and hips yet the most often affected joints in the human body are the knees. As a source of physical disability and public health burden, OA was ranked on par with chronic obstructive pulmonary disease, heart disease, and congestive heart failure. Gait analysis has been a commonly utilized technology in recent years to offer physical therapists and doctors with the kinematic and kinetic data they need to select the best treatment for their patients [1]. Patients with walking difficulties, such as knee osteoarthritis, can benefit from gait measurements. However, there is no analysis study exploring the changes seen in cartilage with several boundary conditions to represent the gait cycles for different grades of knee osteoarthritis.

Mechanical stress is a major factor in the development of OA, and computational modelling of joint degeneration can help clinicians predict the onset and progression of degeneration in the various components of the joint, allowing them to intervene sooner (e.g., weight loss, surgery, rehabilitation). From joint-level musculoskeletal models to finite element models, computational models are available. Furthermore, since musculoskeletal modelling does not measure tissue-specific stresses and strains, it can only provide an average estimation of joint loading and cannot distinguish areas of injury initiation or progression. As a result, no subject-specific predictions are permitted. The finite element model (FEM), on the other hand, can quantify stresses and strains in various joint tissues.

The purpose of this study is to explore the influence of bones and cartilage degeneration in knee osteoarthritis during two different gait cycles from a biomechanical standpoint.

2. Materials and Method

2.1 Development of the Knee Bone

To generate the three-dimensional models, CT datasets in DICOM file format was imported into Mimics (Materialise, Leuven, Belgium). The DICOM dataset has been converted into gray scale values in the software, on which a threshold method is based on previous literature [2]. The range of gray scale values of each unique structure in its related image data set is then determined to define the threshold, after which a 2D geometric model of a single structure was generated from each slice to build a separate mask [3]. During the segmentation procedure, the desired bones, the femur, tibia, and fibula, were picked up from the CT. Three segmented masks are available (femur mask, tibia mask, and fibula mask). For each mask, the threshold ranged from 226 to 3071 HU [4]. To produce a better segmented mask, the segment menu's tool called region grow was used to isolate the bones from the unconnected entities that were in the same threshold but outside of the structure's area. The lowest densities of the bones that were left out of the mask were filled with a smart fill tool from the segment menu. The calculate part tool was used to convert the finalized 2D masks to 3D models when the segmentation procedure was completed. To get a smoother look, all 3D masks underwent wrapping and smoothing processes. The mask modified to fully retain the anatomical information of the segmented structures. Finally, a stereolithographic (STL) file was created from all the 3D models to be able to open the mask in the other software. Fig. 1 shows steps for developing the finite element model of the knee bones.

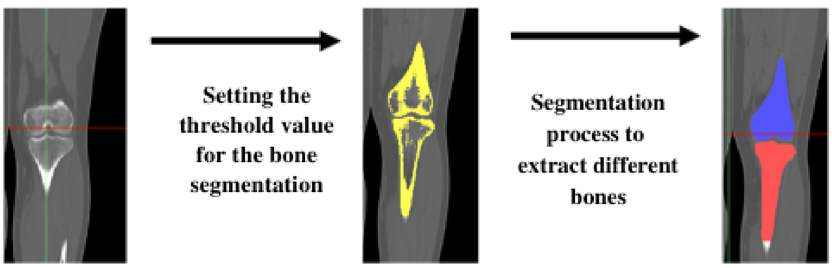


Fig. 1 *The processes for reconstructing a finite element model of the bones of the knee joint*

2.2 Development of the Cartilage

An appropriate technique was used via 3-matic software (Materialise, Leuven, Belgium) to produce 3D models of articular cartilages as shown in Fig. 2. The cartilage layers in the articulations between the femur and the tibia were modelled by precisely finding the articulating surfaces between bones. The initial step was to create a mark on the femur bone at the precise spot. This procedure resulted in the extraction of one layer of femoral cartilage. After that, an offsetting approach was used to create the second layer. The spacing between the two layers was chosen to be (2.3mm) because this is the thickness of the femoral cartilage [5]. A construction solid surface was created to connect the two layers. The tibial cartilage was obtained using the same processes. Because the thickness of the tibial cartilage is 1.71mm [6], the spacing between the two layers of the tibial cartilage was set to that value. All global volume meshing for bone and soft tissue was done with quadratic tetrahedral elements. These elements were chosen over hexahedral elements because they give more flexibility for meshing complex geometrical subject-specific models [7]. From the 3-matic software, all of the models (femur, tibia, fibula, femoral cartilage, tibial cartilage) were exported as 3D volume Patran files.

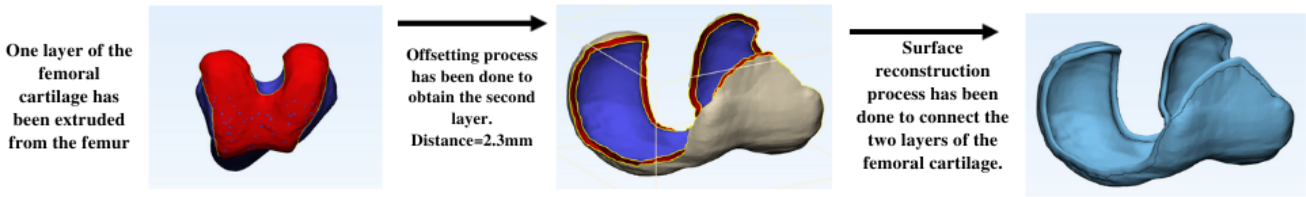


Fig. 2 *The processes for creating a three-dimensional representation of the femoral and tibial cartilage*

2.3 Development of Ligaments

The ligament models were created using the Marc.Mentat program (MSC.Software, Canada). The knee joint has four main ligaments (anterior cruciate ligament, posterior cruciate ligament, medial collateral ligament, lateral collateral ligament). The non-linear features of springs elements, which can be discovered in the link tool, were used to model them all [8-10]. The spring elements were set to true direction, and the node-to-node technique was chosen by picking the first node in the first bone and the second node in the second bone. Fig. 3 shows the location of the modelled ligaments. The spring properties stiffness was set to the same as the ligament stiffness to help mobility and force distribution through the various bones. As demonstrated in Table 1, the ligament stiffness ranges from 20 to 75 N/mm.

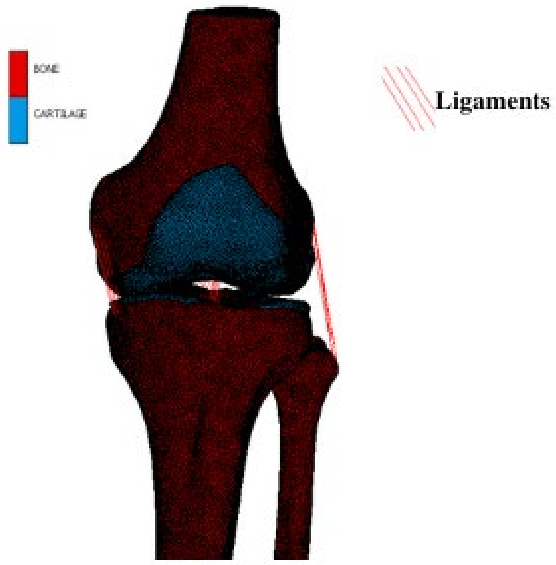


Fig. 3 *Location of the ligaments for finite element model of a knee joint*

Table 1 *Ligament Values for Stiffness Coefficient [5]*

Tissue	K (N/mm)
Anterior cruciate ligament (ACL)	75
Posterior cruciate ligament (PCL)	75
Medial collateral ligament (MCL)	70
Lateral collateral ligament (LCL)	20

2.4 Finite Element Analysis with Different Boundary Conditions

The model was exported to Marc Mentat software to execute the simulation for different grades of osteoarthritis. The loads and constraints that represent the model's relationship with the environment are described by the boundary condition. Several loads that may replicate physiological activities had to be assigned before FE analysis could be done. Only once the FE models had been completed and their material parameters set to represent osteoarthritis grades could this method be carried out. The gait cycle is commonly separated into two phases during walking: stance and swing. The swing phase is defined as the time when the considered foot is not in touch with the ground. The stance phase refers to the time when the foot in concern is in contact with the ground [11]. In this study, the two loading conditions of a normal everyday physiological state were chosen: the swing phase and the stance phase of a gait cycle.

The value of forces generated from muscle and tendon loading was applied axially to the femur bone to represent these loadings. In this investigation, the type of load applied was face load. The weight of our subject is 75 kilograms (750 N). According to the literature, during the standing stance phase, half of the body weight was applied to the foot, resulting in a face load of 375 N [12]. However, approximately 10% of the body weight was generated during the swing period which is equal to (75N) [13-14]. The boundary conditions for two different gaits are shown in Fig. 4. The bones, and cartilage, were considered to be linearly elastic, homogeneous, and isotropic [15] for all simulation. Furthermore, all of the simulations had the same fixed displacement location on the end of the tibia and fibula, as well as the same face load location on the top of the femur. Finally, the models are simulated and subjected to von Mises stress distributions (VMS) and displacement finite element analysis. Fig. 4 shows the boundary condition for all models.

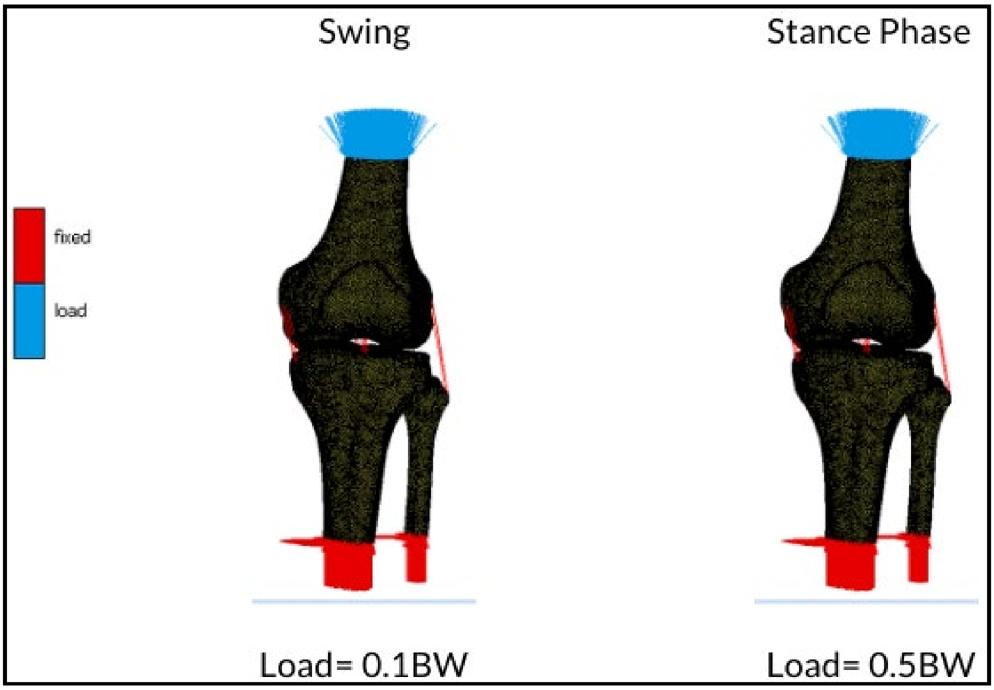


Fig. 4 *Loading condition applied on the knee model*

3. Result and Discussion

3.1 Von Mises Stress

The swing phase and stance phase distributions of von Mises stress were examined. Fig. 5 to Fig. 8 as a contour plot indicate the stress distribution on the femur, tibia, femoral cartilage, and tibial cartilage in these two boundary conditions. Except for the femur bone, the knee OA grade 2 had the greatest levels of stress during the stance phase of all models. Fig. 9 shows a distribution of von Mises stress with different models.

For the femur, the maximum stress during swing phase is (141.035 MPa) whereas it is (147.704 MPa) for stance phase. Moreover, the maximum stress shown on the tibia are (102.617 MPa, and 104.6938 MPa) for swing and stance phase respectively. The soft tissues in our model, femoral and tibial cartilage, show a maximum stress value of (2.6274 MPa, and 2.60278 MPa) for the swing phase. On the other hand, the maximum stress during stance phase is (2.66087 MPa, and 2.73041 MPa).

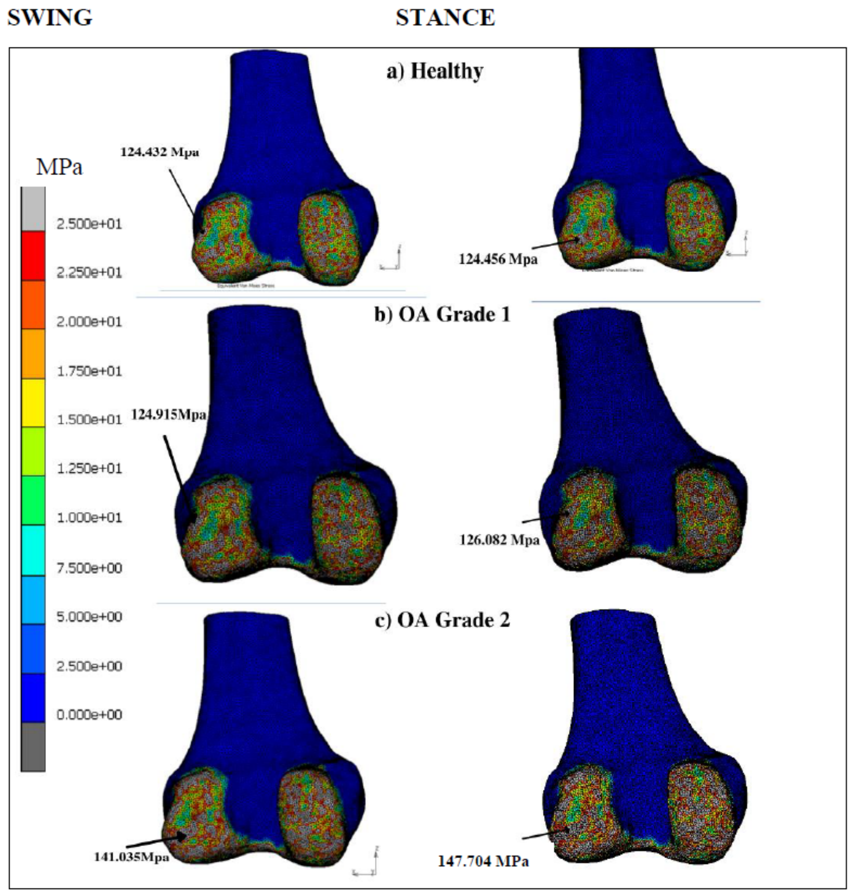


Fig. 5 *Contour band plot of equivalent von mises stress for femur during swing and stance phase*

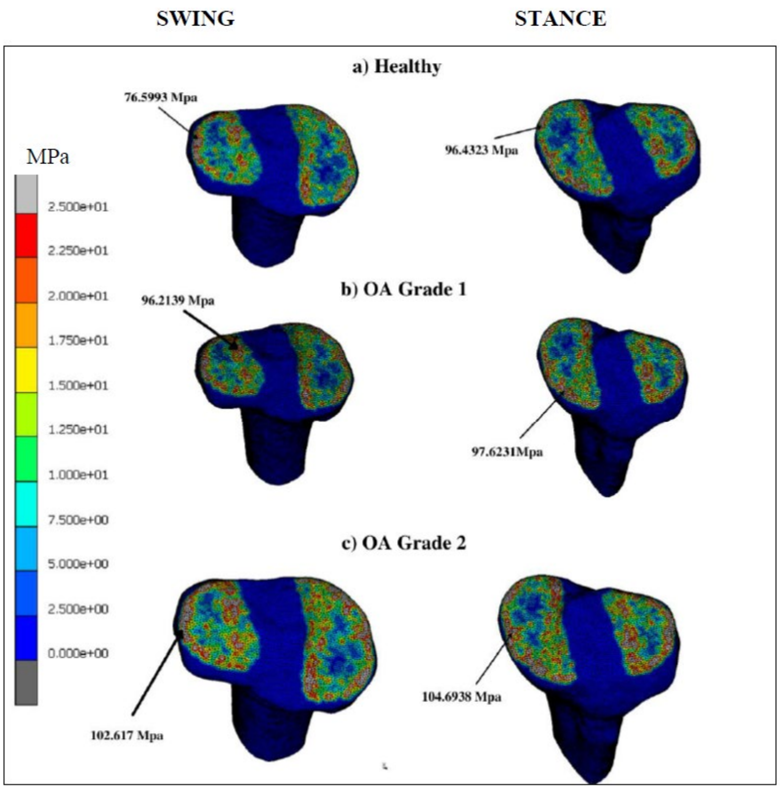


Fig. 6 *Contour band plot of equivalent von mises stress for tibia during swing and stance phase*

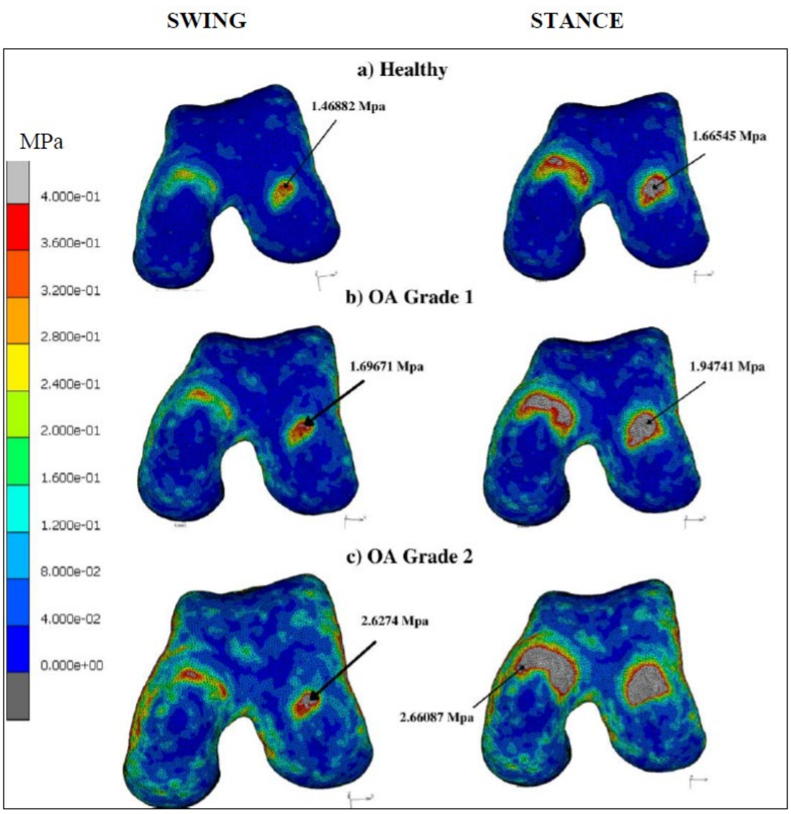


Fig. 7 *Contour band plot of equivalent von mises stress for femoral cartilage during swing and stance phase*

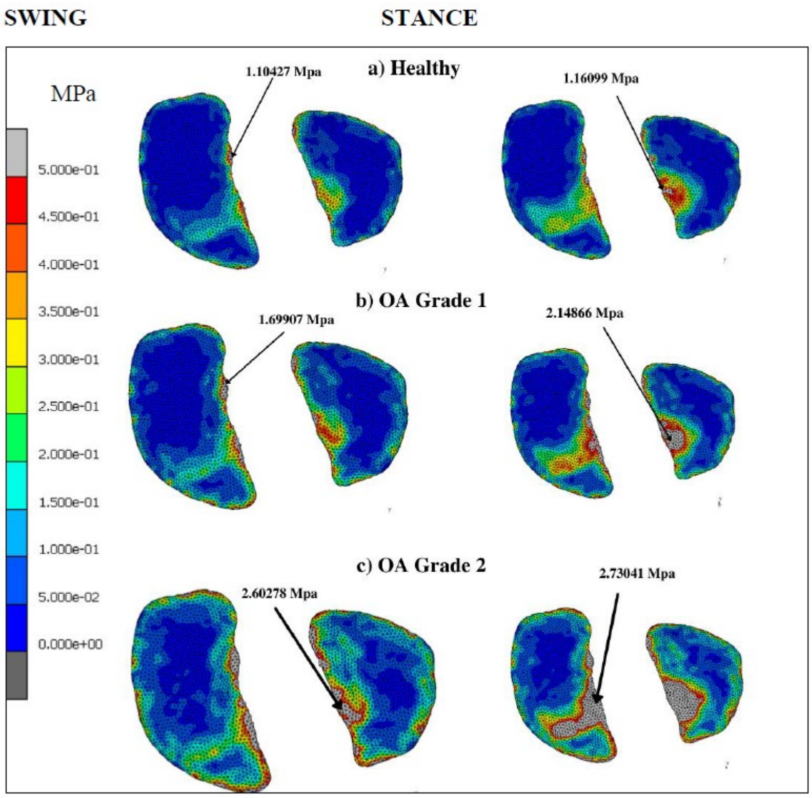
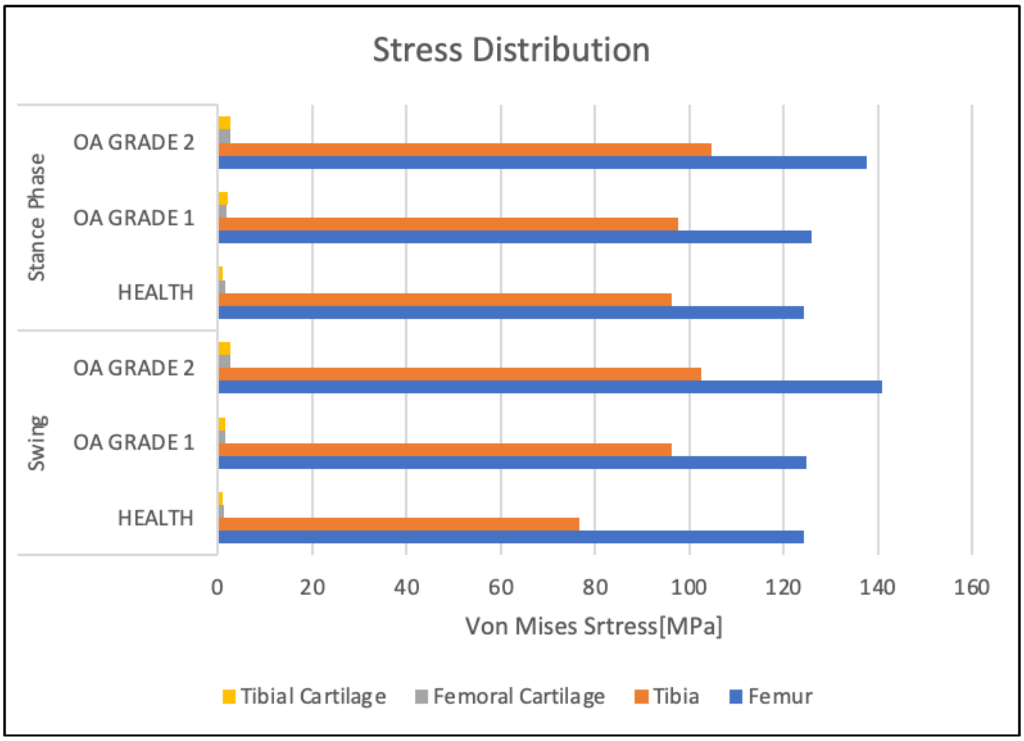


Fig. 8 *Contour band plot of equivalent von mises stress for tibial cartilage during swing and stance phase*



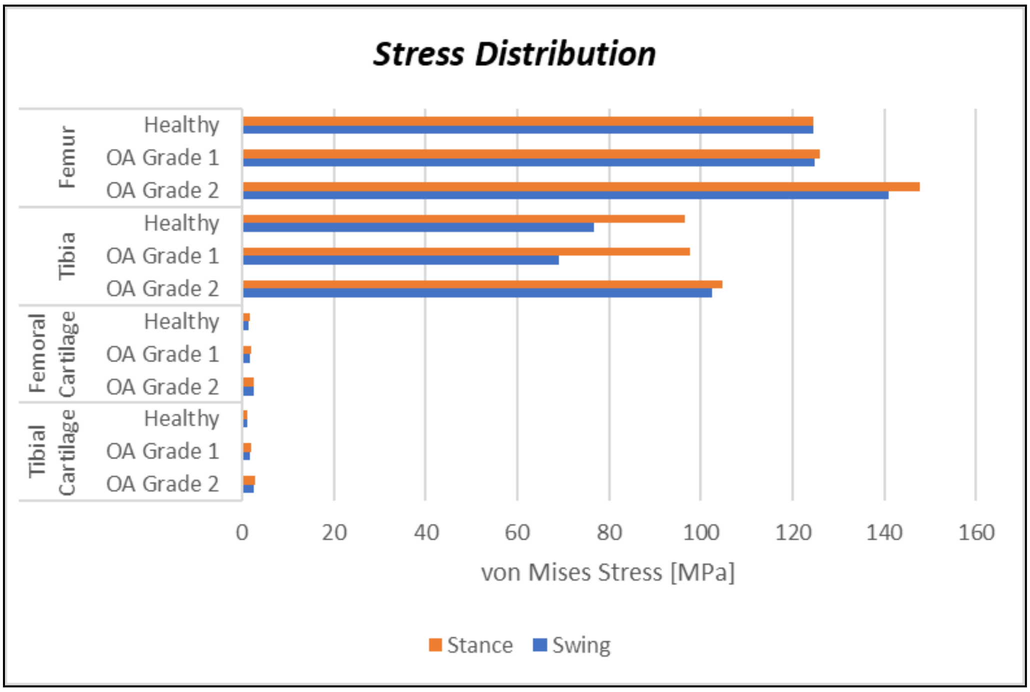


Fig. 9 *Graph of the stress distribution*

3.2 Displacements

The distributions of displacements in the swing and stance phases were Fig. 10 to Fig. 13 show the displacement distribution on the femur, tibia, femoral cartilage, and tibial cartilage under these two boundary conditions. During the stance phase, the knees with OA grade 2 had the most displacements of all the models.

The greatest displacements for the femur during the stance phase are (8.77645 mm), while the maximum displacements for the swing phase are (2.7055 mm). Furthermore, the maximum displacements shown on the tibia for the swing and stance phases are (0.056795 mm and 0.0151765 mm, respectively). For the swing phase, the soft tissues in our model, femoral and tibial cartilage, show maximum displacements of (1.65971 mm and 2. 0.757734 mm). The maximum displacements during the stance phase, on the other hand, are (4.25937 mm, and 0.834481 mm).

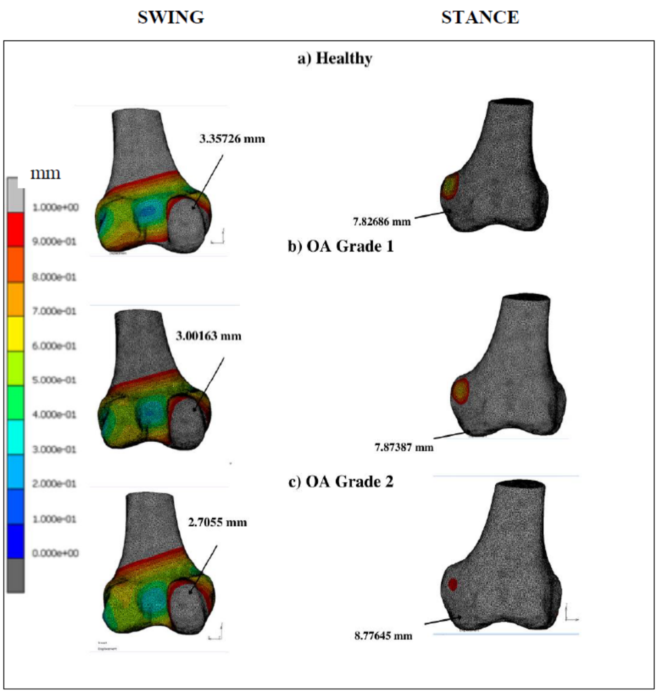


Fig. 10 *Contour band plot of displacement for femur during swing and stance phase*

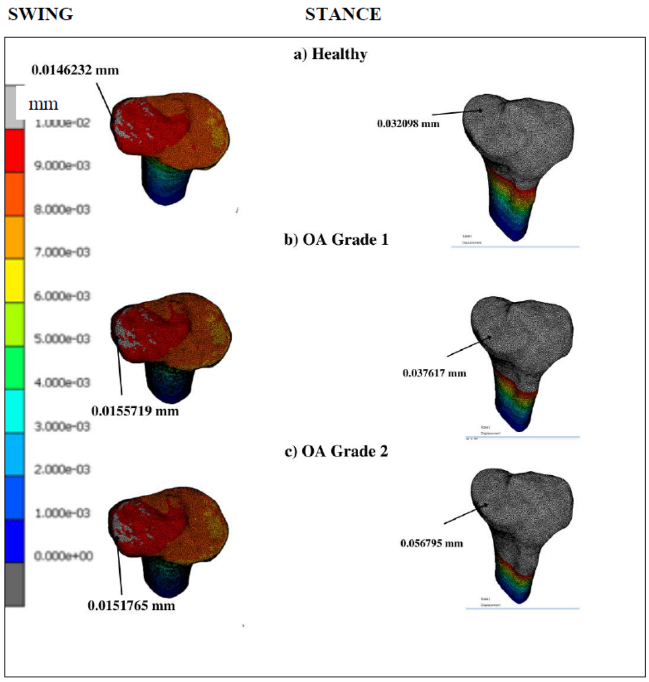


Fig. 11 *Contour band plot of displacement for tibia during swing and stance phase*

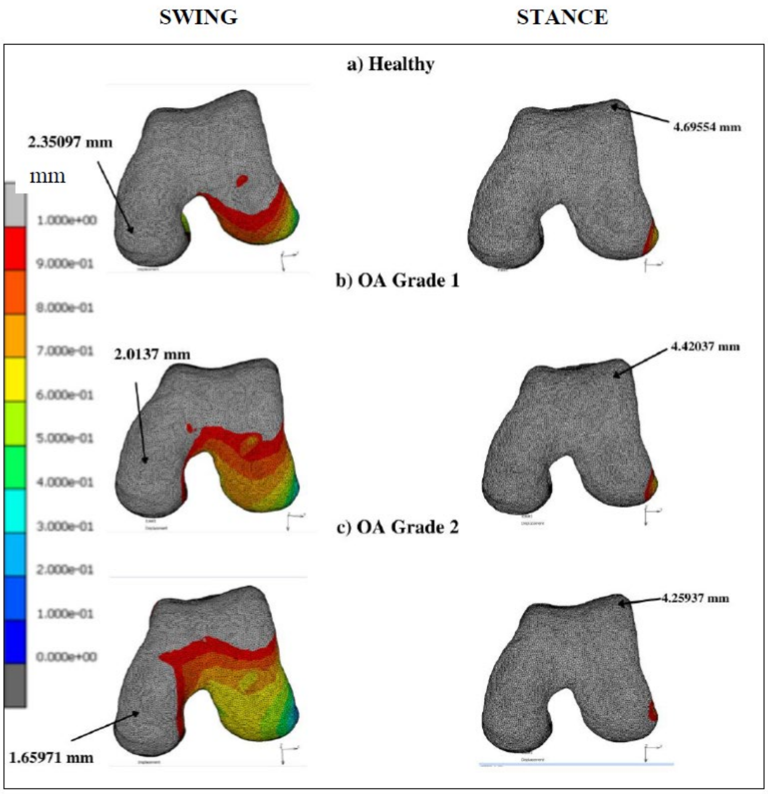


Fig. 12 *Contour band plot of displacement for femoral cartilage during swing and stance phase*

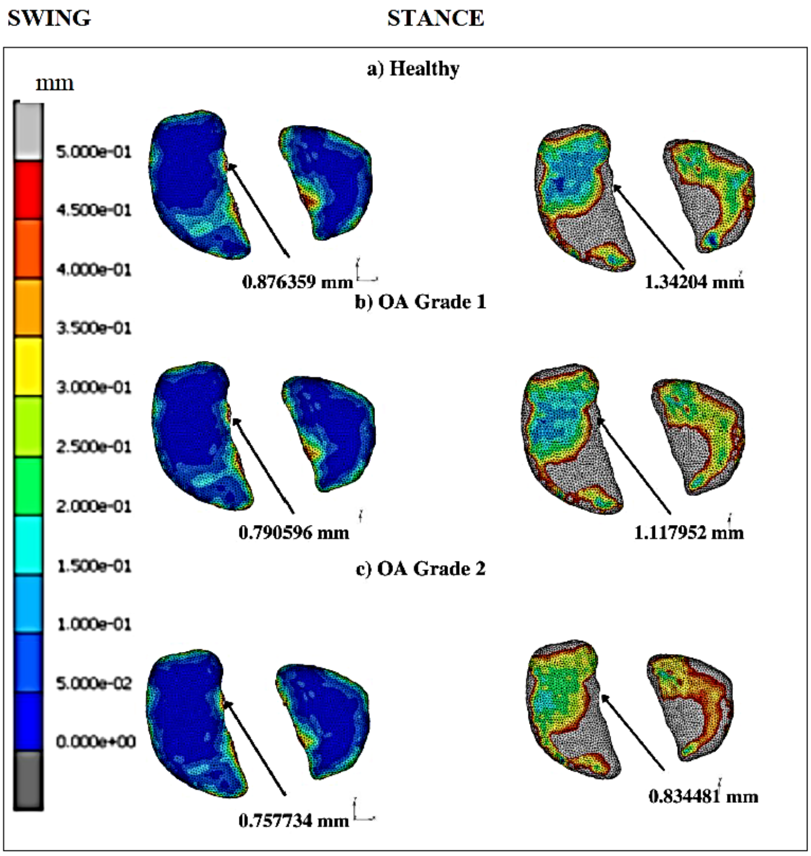


Fig. 13 *Contour band plot of displacement for tibial cartilage during swing and stance phase*

3.3 Discussion

The distribution of stresses and displacements in the health and osteoarthritis knee joint under varied axial loading is determined using FE analysis. In general, the findings show that with increasing body weight load, stresses and displacements in articular cartilage and bones increased.

For the stress results, the maximum stress on the tibia bone during stance phase (104.6938 MPa) is higher than the maximum stress in the tibia bone during swing phase (102.617 MPa). The same situation has been observed for the articular cartilage (femoral and tibial) which has been increased from (2.6274 MPa) to (2.66087 MPa) for the femoral and from (2.60278 MPa) to (2.73041 MPa) for the tibial for stance and swing phase, respectively. Based on previous research, it is known that ultimate strength of the human articular cartilage is (2.18MPa) [16]. Comparing with the results obtained, it clearly see that the maximum von Mises stress during the swing phase for the femoral cartilage and tibial cartilage in the healthy state are 1.46882 MPa, 1.10427MPa, respectively. Moreover, the maximum von Mises stress during stance phase for the femoral cartilage and tibial cartilage in the healthy state are 1.66545 MPa, 1.16099 MPa, respectively. These two values are less than the ultimate strength of the cartilage. From this comparison, findings shows that there is no degeneration in the articular cartilage during healthy state for both phases which is expected.

For grade 1 Knee osteoarthritis, the maximum value for the von Mises stress is 1.69671MPa for femoral cartilage and 1.69907 MPa for tibial cartilage during the swing phase. On the other hand, the maximum values for the von Mises stress is 1.94741 MPa for femoral cartilage and 2.14866 MPa for tibial cartilage during the stance phase. These two values actually are still less than the ultimate strength of the cartilage even though it is already exposed to the osteoarthritis. This result is accepted since grade 1 is an early grade of the osteoarthritis and there is an increase of the stress in this grade comparing to grade 0 (healthy knee). However, the stress on the tibial cartilage during stance phase is almost near to the ultimate strength where it showed the degenerated happen faster when the weight is high.

By comparing the stress value in grade 2 with the ultimate strength, we can find out that the stress values during swing and stance phase in articular cartilage are higher than the ultimate one. From this finding we can conclude that the femoral cartilage and the tibial cartilage are damage in this level of osteoarthritis and has been started to degenerate.

Furthermore, same comparison procedures have been done to the bones in evaluating the level of degeneration that they exposed to. The ultimate strength for the bone is 133 MPa [17]. By comparing this value with the results, stress for the femur is 124.432 MPa and tibia is 76.5993 MPa during the swing phase for a healthy status and for both of them during grade 1 is 124.915 MPa and 96.2139 MPa, respectively. From the stress value as compared with the ultimate strength, it is predicted that the bones in these two grades have not been damaged or degenerated, even though there is an increase of the stress value in grade 1 knee OA.

On the other hand, the femur bone during grade 2 of the osteoarthritis has a value of 141.035 MPa during the swing phase and a value of 147.704 MPa during the stance phase, in which these values exceeded the ultimate strength of the bones. From here, it is predicted that the articulating cartilage of OA knee start to degenerate and got damaged. However, the tibia bone during the swing in this grade has a value of 102.617 MPa and a value of 147.704 MPa during the stance phase, which is still less than the ultimate value. In this point, it is predicted that the tibia has not been degenerated yet for this grade.

As a result, the load imposed to the knee joint is proportional to the maximal displacement and von Mises stress. It is possible that the primary causes of osteoarthritis of the knee, such as obesity, regular exercise, and joint trauma, put stress on the articular cartilage and bones in the knee that is higher than usual. This damages the articular cartilage and, as a result, the bones in that knee. This indicate that being overweight or obese may be a risk factor for joint articular cartilage wear and the development of osteoarthritis of the knee.

In this study, finite element model has been verified and validated early prior to the analysis on different OA conditions [18-19]. For verification, we compared our 3D model with the one that have been used by Wang et al. [20]. A total of 10 simulations have been conducted in our previous study [18] where the forces were ranging from 100N to 1000N. The results of von Mises stress were compared with [20] and found that the trend of stresses was similar. For a validation work, our model has been validated through an experimental work using synthetic bone by comparing the result of vertical displacement of the knee [19]. A total of six experiments have been established with the compression load of 10N until 50N have been conducted for the comparison. Based on the results, the finite element model has been validated with similar results obtained.

Some simplifications have been introduced in the finite element analysis in this study. Firstly, the use of isotropic, linear and homogeneous properties of bone and cartilage model. It is well known that the bone and cartilage are not behaved in isotropic and homogeneous properties. However, due to the constraints on the computer resources to simulate such a complex property of bone and cartilage, thus, this simplification could not be avoided as demonstrated by other literature. The other limitation of study was regarding the use of a single CT dataset of human. As a recommendation for the future study, more data is needed to be simulated so as to provide a convince predictions in the future.

4. Conclusion

Consequently, For the clinical diagnosis of disorders affecting the knee joint, observations from biomechanics and finite element (FE) analysis of the knee joint are helpful. To record tissue reactions to external loads, the FE method is utilized, which is well-established in the biomechanics field. The findings from this study could help to better understand how the joint behaves mechanically during various gait phases. This indicates that being overweight or obese may be a risk factor for joint articular cartilage degeneration and the initiation of osteoarthritis of the knee. High loads increase von Mises stress and displacements, according to the presented data. Considering subject-specific loading conditions, it addresses the necessity for research into displacement and stress distribution in knee joints.

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Conflict of Interest

Authors declare that there is no conflict of interests regarding the publication of the paper.

Author Contribution

The authors confirm contribution to the paper as follows: ***study conception and design:*** *Aceel Ahmed Hasan Alsaqqaf, Muhammad Hanif Ramlee;* ***data collection:*** *Muhammad Hanif Ramlee, Mohammed Rafiq Abdul Kadir;* ***analysis and interpretation of results:*** *Aceel Ahmed Hasan Alsaqqaf, Muhammad Hanif Ramlee, Gan Hong Seng, Ahmad Kafrawi Nasution;* ***draft manuscript preparation:*** *Aceel Ahmed Hasan Alsaqqaf, Muhammad Hanif Ramlee,* *Gan Hong Seng, Ahmad Kafrawi Nasution, Mohammed Rafiq Abdul Kadir. All authors reviewed the results and approved the final version of the manuscript.*

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