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Three Dimensional Finite Element Modelling and Analysis of Human Knee Joint-Model Verification

Three Dimensional Finite Element Modelling and Analysis of Human Knee Joint- Model Verification

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Abstract. Modelling a three dimensional (3D) model of a human knee joint by extracting the region of interest accurately is one of the main constraints. Oversimplified bone models from previous studies that could affect the accuracy of analyses have become current concerns. An approach to minimize the issue consisting of several steps were done. This study aim to analyse a more precise human knee joint model using finite element technique. Reconstruction of 3D knee models were done by segmenting Computed Tomography (CT) data of a healthy male. Cancellous and cortical bones were segmented based on the Hounsfield unit (HU). The model of knee consists of femur and tibia bones, cartilages and ligaments. Construction of cartilages were done by extracting and offsetting bone layers. Linear spring elements were used to model four ligaments at the knee joint. In order to verify the models, finite element analyses were carried out. Forces ranging between 100 until 1000 N were axially applied on the proximal femur. The results in this study were in an agreement with previous literature reports with maximum peak VMS of 2.928 MPa and 3.25 MPa respectively at articular cartilages. It can be concluded that the knee models were verified.

1. Introduction

One of the largest and complex joint in the lower extremities is knee which is an important load-bearing joint [1]. It is located in between distal femur and proximal tibia bones [2]. There are six different motions of the knee joint that include flexion-extension, internal-external, varus-valgus, medial-lateral, proximal-distal and anterior-posterior [3]. Forces measured at the knee joint during walking were approximately 180 % to 280 % of body weight (BW) and the forces peak up to 350 % BW for stair ascending and descending [4]. The knee is vulnerable to injuries that include fracture and tear of ligaments when excessive forces acted on the joint [5, 6].

Finite element models of human bones and joints to be used in biomechanical analysis have been recognized significantly and widely used since they are capable in providing the understanding of biomechanical behaviours under different types of configurations and loadings [7]. This method could predict and measure local parameters such as internal stress, strain and displacement that would be difficult to be obtained from experimental works [8, 9]. Its efficiency in terms of cost, time and availability is supported by its ability to produce reliable insight and prediction of analysis.

Several finite element studies on knee joints have been reported previously and some of them modelled the bones to be one bone layer (cortical bone only) instead of two bone layers (cortical and cancellous) [10, 11]. Due to that, the material properties for both cortical and cancellous bones were assigned to one single value rather than two different values. These oversimplified models as stated from previous literature could affect the accuracy of analysis [12, 13]. Therefore, a step by step method was carried out to design a more reliable and accurate model of knee joint. The method of developing knee models consists of several steps that includes segmentation of cortical and cancellous bone layers and modelling soft tissues such as cartilages and ligaments. Material properties were assigned to the knee models accordingly and simulated with compression loads ranging from 100 to 1000 N. The results were then compared with other previous study.

2. Materials and Methods

2.1. Three-Dimensional Reconstruction of Bones

Computed Tomography data of a healthy male volunteer with no radiological sign of pathology, 27 years old with height of 169 cm and 75 kg in weight were used to reconstruct 3D models of human knee joint. Mimics (software version 10.01, Materialise, Leuven, Belgium) was used in the development of bones at which different HU values were applied to segment the region of interest autonomously. The HU for segmentation is in a range of 200 to 750 HU for cancellous bone and 750 to 3071 HU for cortical bone [14]. In order to obtain an accurate shape of bone, the layers of CT data were then segmented manually using ‘erase and draw’ functions. By using the layers that have been segmented, a 3D knee joint that consists of femur and tibia with cortical and cancellous bones were reconstructed as illustrated in Figure 1.

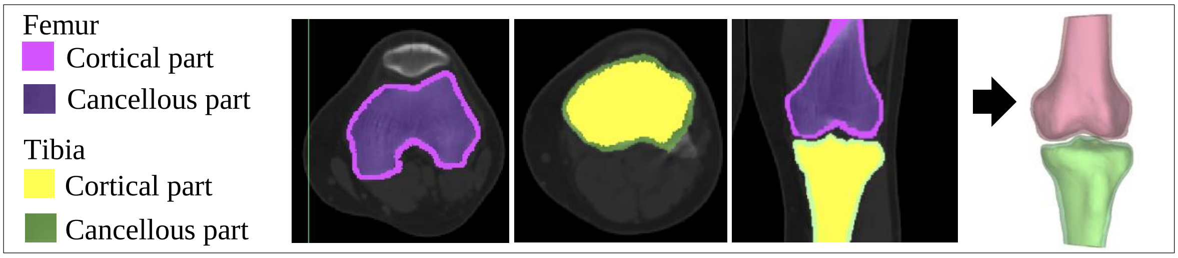


Figure 1. Segmentation of cortical and cancellous bone layers of femur and tibia.

2.2. Development of Cartilages

Cartilages at the knee joint cover the end of distal femur and proximal tibia that function as cushioning effects for shock absorption and minimize the frictions between the bones during knee movement [15]. The cartilages are considered to be linear elastic and isotropic since the loading time for time constant of viscoelastic properties of cartilages approaching 1500 seconds are far more than the loading time of interest that corresponded to single leg stance [16]. In order to predict the behaviour of cartilages under short-term condition, the properties is applicable and accurate enough since there are no significant changes in the contact behaviour of cartilages shortly after loading as demonstrated and proven by Donzelli et al. [17]. Surface layers at femur and tibia were extracted by following the anatomical shapes of cartilages at knee joint. The thickness of the region of interest from the bones that represents the cartilages were developed according to previous literature that ranges from 1.54 to 2.98 mm [18]. Figure 2 shows the modelling process of articular cartilages for femur and tibia at the knee.

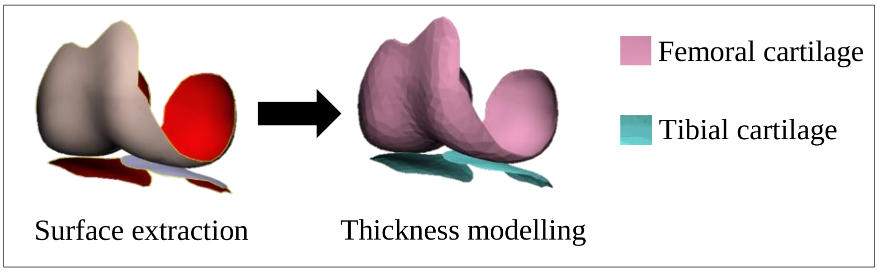


Figure 2. Modelling of articular cartilages at knee joint using surface extraction and extrude function.

2.3. Modelling of Ligaments

Ligaments are soft collagenous tissue bundles that connect one bone to another bone to form joints and enable movements [19]. Linear spring elements were used to model four ligaments at knee joint. Anterior Cruciate ligament (ACL), Posterior Cruciate ligament (PCL), Medial Collateral ligament (MCL) and Lateral Collateral ligament (LCL) were developed accordingly between femur and tibia bones by referring to anatomical attachment sites [20].

2.4. Finite Element Modelling

Finite element models of bones and cartilages were imported into 3-Matic (software version 6.1, Materialise, Leuven, Belgium) for mesh editing. Triangular surface meshes were generated on the surface layer of bones and cartilages. The bone geometry of the knee joint was checked to ensure that it is in a proper anatomical size and shape. Boolean operation was performed to ensure that there are no intersection bodies exist in the finite element models. The surface meshes for all models were converted into volume meshes to generate 3D solid models. The knee joint were set into two different degrees of knee flexion which are at 0° and 30° of flexion as illustrated in Figure 3. Both of the knee joint models with different knee flexion were exported into Patran files for biomechanical analysis in the simulation software.

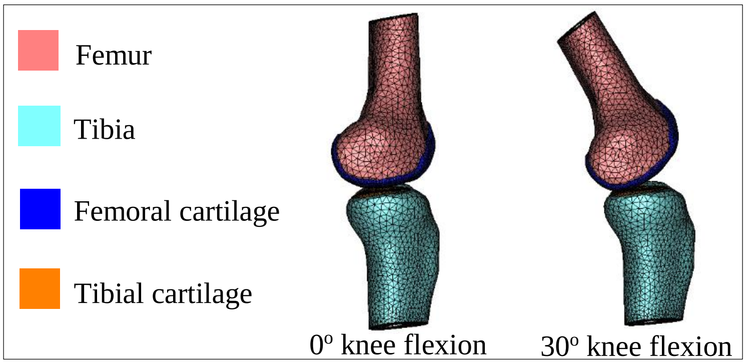


Figure 3. Finite element models of knee joint with 0° and 30° knee flexion.

2.5. Biomechanical Analysis

The finite element models were imported into MSC.Marc Mentat (software version 2005r2, MSC.Software Corporation, Germany) to be assigned with respective geometrical and mechanical properties [10]. Table 1 below shows the mechanical properties of bones, cartilages and ligaments that are used in the simulation. Young modulus (E) and Poisson’s ratio (v) of the tissues are set based on previous literatures [20, 21]. Linear link elements that represents ACL, PCL, MCL and LCL were created by using spring elements between femur and tibia. The stiffness coefficient (K) value of each ligament bundles ranging from 20 to 75 N/mm as shown in the table [22]. Next, contact bodies were created between cortical bones and cancellous bone of femur and tibia together with the articular cartilages [24]. Compression loads that vary from 100 to 1000 N were applied axially on femur for full knee extension. Meanwhile, 35 N of compression load was assigned for both knee joint models with different knee flexion as shown in Figure 4.

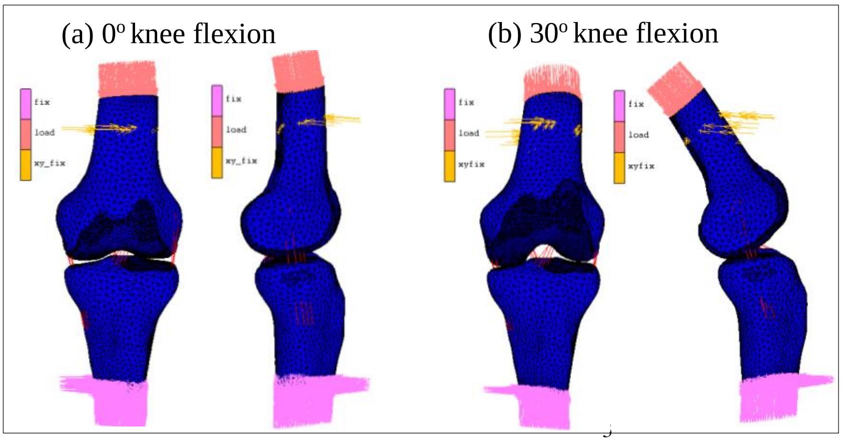


Figure 4. Finite element analysis of knee joint with boundary conditions of 35 N compression load and fixed displacement at two different knee flexion; (a) flexion of 0°, (b) flexion of 30°.

Table 1. Mechanical properties of bones, cartilages and stiffness coefficient value for ligaments.

Tissue	K (N/mm)	E (MPa)	ν
Cortical bone	-	16200	0.36
Cancellous bone	-	389	0.3
Articular cartilages	-	10	0.4
ACL	75	-	-
PCL	75	-	-
LCL	20	-	-
MCL	70	-	-

3. Result and Discussion

3.1. Finite Element Analysis

Results from the analysis are presented in the following figures. Figure 5 shows the contour plot of peak Von Mises Stress (VMS) at articular cartilages of knee joint and the variation of the predicted peak VMS results using finite element analysis are explained using graph in Figure 6. From the presented figures, the value of peak VMS at knee joint increases with increment of compression loading force. Minimum value of peak VMS for both analyses were during 100 N loading, with 1.69 MPa from literature and 1.012 MPa from simulation. Meanwhile, the maximum value of peak VMS were at 1000 N compression load, with 3.25 MPa and 2.928 MPa respectively [21].

Based on Figure 6, the results from the literature were observed to have higher magnitude as compared to the simulation results obtained in this study. This might be due to the variables exist in the finite element models that include value of mechanical properties assigned to the models and their geometrical sizes. Generally, a larger Young modulus would increase VMS while increase in Poisson’s ratio would result in lower VMS [21]. Nonetheless, the mechanical behaviours of the cartilages at the knee joint in terms of peak VMS between the literature and the finite element analysis from this study exhibit similar patterns.

On the other hand, Figure 7 illustrated the peak VMS for knee joint with two different flexion. It could be observed that the value of peak VMS was higher in 30^{o} knee flexion as compared to knee with 0^{o} knee flexion. The contact area between the cartilages at knee joint were noticed to be reduced in value when the knee flexion increases from 0^{o} to 30^{o}. In general, the value of pressure increases when the surface area decreases. This is also proven by previous literature that reported increases in knee flexion will decrease the contact area and increase the contact pressure at knee joint [23]. Thus, results from this simulation that proved the contact pressure increases when the knee flexion angle increases were in coincident agreement with the experimental works.

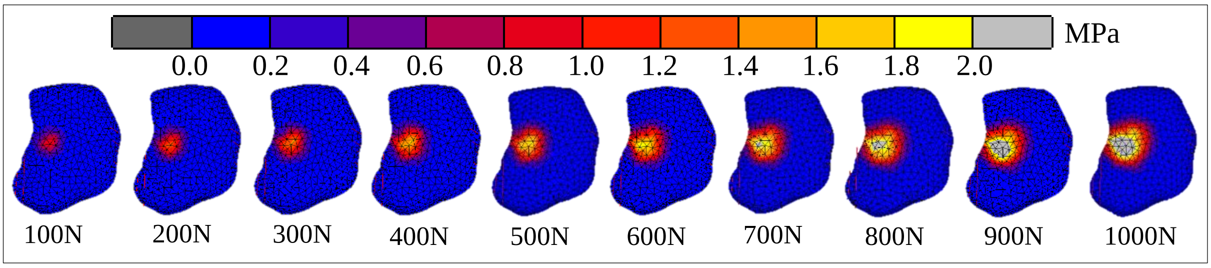
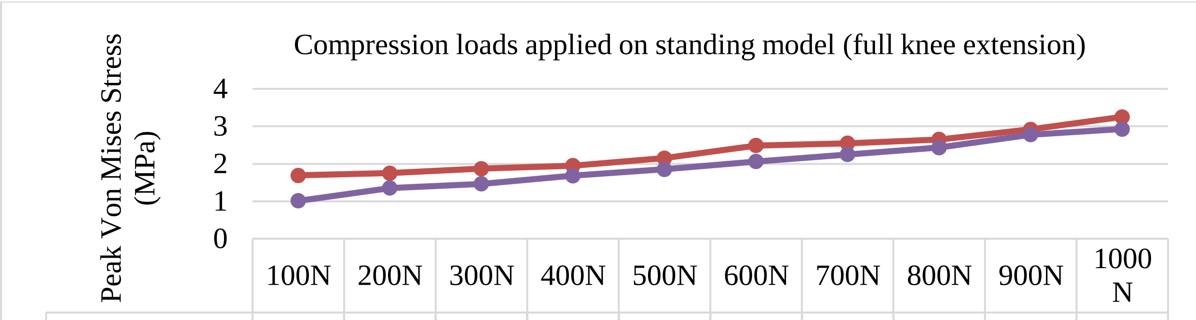
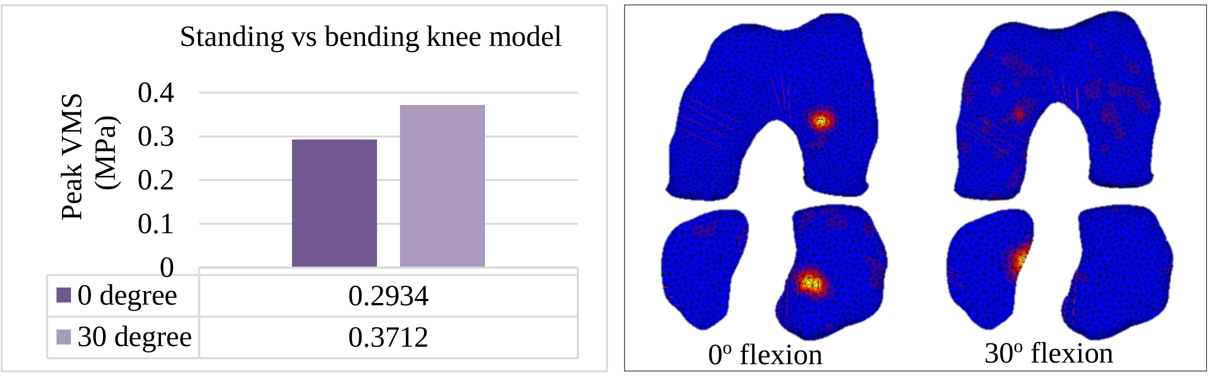


Figure 5. Contour plot of Peak Von Mises Stress at articular cartilages from FEA.



	100N	200N	300N	400N	500N	600N	700N	800N	900N	1000 N
Wang et. al	1.69	1.75	1.87	1.95	2.15	2.49	2.55	2.65	2.92	3.25
Simulation	1.012	1.356	1.466	1.681	1.854	2.062	2.251	2.434	2.778	2.928

Figure 6. Peak Von Mises Stress at articular cartilages on knee joint from previous literature work and finite element simulation.



0 degree	0.2934
30 degree	0.3712

(a) (b)

Figure 7. Peak Von Mises Stress at articular cartilages on knee joint at 0° and 30° knee flexion; (a) graph with table, (b) contour plot.

4. Conclusion

The 3D finite element models of human knee joint based on CT data were developed and analysed successfully by using Mimics, 3-Matic and MSC.Marc Mentat softwares. The cancellous bones, cortical bones, cartilages and ligaments were modelled to complete the knee joint. From the results and discussions on the models and simulation, it can be concluded that the finite element analyses of the 3D knee joint were in an agreement with previous literature works in terms of maximum value of peak VMS at contact point on articular cartilages with value of 3.25 MPa and 2.928 MPa respectively. In addition to that, the increase in contact pressure when the degree angle of knee flexion increase that was observed from the simulation was in concurrent with reported experimental investigation.

Acknowledgments

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