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Computational Modelling of Patella femoral kinematics during gait cycle and experimental validation

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Abstract

The effect of loading and boundary conditions on patellar mechanics have been greatly important due to the complications arising in patella femoral joints during total knee replacements. To understand the patellar mechanics with respect to loading and motion, a computational model representing the patella femoral joint was developed and validated against experimental results. The computational model was created in IDEAS NX and simulated in MSC Adams/View. The results obtained in the form of internal external rotations and anterior posterior displacements for a new and experimentally simulated specimen for patella femoral joint under standard gait condition were compared with experimental measurements performed on the Leeds ProSim knee simulator.

A good overall agreement between the computational prediction and the experimental data was obtained for patella femoral kinematics. Good relation between the model and the past studies were observed when the ligament load was removed and the medial lateral displacement was constrained. The model was sensitive to ±5% change in kinematics, frictional, force and stiffness coefficients and insensitive to time step.

Keywords: Patella Femoral Joint, Kinematics, Knee Simulator, Validation

Introduction

The complications arising from the patella femoral joint (PFJ) leading to total knee replacement (TKR) revisions is a concern around the world [1-9]. Hence, the effect of loading and boundary conditions on patellar mechanics is important. In the past, the patellar mechanics have been investigated *in vivo* using magnetic devices [10], motion analyses [11-12] and photographic devices [13-14]. All these investigations were performed in knee joint for flexion angle above 90^o. The effect of various translations and rotations were recorded and compared at uncontrolled or constrained tibial rotation. However, the accuracy and repeatability of the PFJ kinematics has been complex and harder to quantify *in vivo* due to the smaller surface area of the patellae and hence, the positioning of the pins for determination of kinematics becomes challenging [14].

The *in vitro* model is another way to validate and improve the accuracy and repeatability. The model assess different factors related to design and contact mechanics which affects the kinematics resulting in maltracking. However, the cost associated with manufacturing and time for testing patient's stratification is huge and in many cases, impossible to meet. Computational modelling is an inexpensive alternative way to analyse these features. However, initial validation against the experiment is crucial. Verified computational models create the opportunity to further understand the mechanics and motion tracking, which can be difficult to obtain experimentally. Computational models in addition are helpful at the design stage in determining the possible failure and arriving at proper design without the need to repeat the manufacturing process and conducting difficult experiments. The verified kinematic model also acts as the first step in the prediction of the wear rate when the experimental wear simulations are costly and time consuming.

The explicit finite element models of the Kansas knee simulator (KKS) have been developed in the past [15-18]. The KKS model predicted the kinematics of knee implants due to the variations in load and ligament tensions. The resulted kinematics were verified with experimental KKS simulation [19]. The Leeds Knee Simulator is another platform which can be employed for computational and experimental wear simulations. However, the first step is the active comparison between the kinematics predicted by the computational model and experimental simulation.

The aims of this study were to evaluate the explicit kinematics of the artificial PFJ and hence, validate the results with the experimental model. The objectives were to create and develop a PFJ model for both new patellae button and patellae button that have undergone experimental wear simulation. The internal external rotations and anterior posterior displacements were predicted from the computational model and verified against experimental observations. The model was also tested for sensitivity analysis at various input parameters including friction, percentage change in input kinematics, material stiffness and number of steps. The clinical relevance of the model was to give an overall understanding of the effects of various parameters on the PFJ biomechanics.

Materials and Methods

The components used for the wear simulation test were commercially available; Co-Cr PFC Sigma® right femur (size 3) and 38mm UHMWPE round dome patella supplied by DePuy Synthes Joint Reconstruction (Leeds, UK). The average surface roughness (Ra) for the dome patella buttons and femur components were 1.0 ±0.23 µm and 0.03 ±0.01 µm respectively.

The kinematics profile was chosen to represent the physiological behaviour of the patella during the complete gait cycle. Data from past investigations on total knee replacements was not sufficient to decide on the control strategy or kinematics profile. So, data was obtained from a combination of anatomical and post-replacement investigations (Figure 1). The parameters acting on the femur were Flexion Extension (FE) and axial load, which passed through the centre of the patella as shown in Figure 2. The patella was acted on by Abduction Adduction (AA) rotation, Internal External (IE) rotation (also, known as patella tilt), Medial Lateral (ML) and Superior Inferior (SI) displacements. The maximum flexion angle acting on the femur was 22 degrees and the total SI displacement was 22mm. The AA rotation (1^{o} maximum) was based on data from Ellison et al. [20], Lafortune and Cavanagh [21] and Halloran et al. [15-16]. The axial load was taken from Gill and O'Connor [22], with a maximum load applied through the central patellar axis of 1200 N.

In vitro testing

The recently described Leeds Patella simulator (Simulator Solutions, UK) was used for this study [23]. The uncontrolled ML and AP displacements were measured using a LVDT transducer (RDP Group CE S7M Transducer, Wolverhampton, UK) and recorded with an oscilloscope (Tektronix TDS 210, Florida, USA). The ML displacement introduced cross shear at the articulating surfaces and hence, is an important factor for influencing wear of conventional polyethylene [24-25]. The IE rotation was measured using a potentiometer (ASM Gmbh, Germany) and was recorded with an oscilloscope. The resistance by medial retinaculum equivalent to 10 N [26] was applied by introducing a load of 0.2 kg on the lateral side of the PFJ. This load induced a resistance; similar to medial retinaculum resistance to medial translation in an anatomical state. In addition, the 0.2 kg load assisted in avoiding patella slip at higher IE rotations. Three readings for each output were recorded for accuracy and repeatability and mean for five specimens with 95% confidence limits are presented.

Multi body solid dynamics (MBSD) model

A three dimensional model of the Leeds Patella simulator was created in I-Deas v 11 NX (Siemens, Texas, US). The CAD drawing of the femur and new patella specimen model were obtained from DePuy (DePuy International, Leeds, UK) and the models were imported to I-Deas v 11 NX for assembly. The model of the patella button that went through experimental wear simulation for 6 million standard gait cycles was initially scanned in MicroCT 80 (Scanco Medical, Busserdorf, Switzerland) in form of slices, followed by reconstruction in SCANIP software (Simpleware software, IN) and exported in the I-Deas v 11 NX for assembly. The procedures followed for construction to execution of the model are briefly highlighted below.

1. Export component and simulator design from I-Deas v11 NX in parasolid format.
2. Import the parasolid files to MSC Adams/VIEW R3 (MSC Software Corporation, CA, USA).
3. Apply constraint, inertia, friction and material properties.
4. Force and displacement feedbacks for each actuator (i.e. experimental kinematic outputs) were used as the actual input profiles for the computational model.
5. Initially, the model was tuned to the experimental model at 4 degrees of freedom with constrained ML displacement and IE rotation.
6. Following tuning, the model was executed under simulator conditions i.e. active six degrees of freedom with uncontrolled IE rotation ($<5.2^{o}$).
7. The results from model were compared to the experimental observations at different conditions.

All connecting fixing links between the fixtures/parts were modelled as perfect unions. The revolute links and translational links were frictionless. The station centre of gravity and the moment of inertia were measured from the fixtures in the Leeds knee simulator using weighing balance (KERN FTB 35K1, Eyholz, Switzerland) and applied to the model. Contact parameters were obtained from previous investigation by Ellison [27]. Loading-unloading tests were carried out in a servo-hydraulic universal testing machine with a maximum force of 1.2kN. Parameters for the model were stiffness coefficient = 5702 N/mm, force coefficient = 1.9, damping coefficient = 35.4 N/mm and displacement 0.4333 mm at 1.2 kN [27]. Tri-pin on disc tests have shown independency of friction on sliding within velocity range 35 to 240 mm/s [28].

The stiction velocity and dynamic velocity were fixed to 35 mm/s and friction coefficient to 0.04 between patella and femur surface [15-16, 29]. The model sensitivity for different frictional coefficients (0.01-0.1) and the number of steps (100-1000) were obtained. The sensitivity to a 5% change in input kinematics was also performed to investigate the influence of the model to different input kinematics.

The analyses are based on the following simplifying assumptions.

1. All bodies were considered as rigid
2. Patellofemoral contact was represented as spring damping element based on simple elastic impact algorithm.
3. All joints were considered to have zero friction except the patella femoral contact joint.
4. All fixtures were manufactured without consideration of tolerance.
5. There was no material loss due to surface wear.
6. All materials were considered as homogeneous.

Results and Discussion

The medial displacement was mainly due to the curvature of femoral groove and increase in SI translation. The direction of displacement was dependent on the direction of patella articulating groove which in the current study was medial. The maximum medial displacement was 3.5 and 4.5 mm for experimentally simulated and new specimens respectively at highest flexion and SI displacement. Chew and Co-authors [10] also reported that majority of the displacement in their artificial implants were medial. However, their PFC sigma control specimen showed lateral displacements, completely opposite to the current study. This may be due to the soft tissue constraint influencing the patella movements.

The kinematic profiles predicted by the computational model for new and experimentally simulated patella specimens (at uncontrolled ML displacement) for AP displacement and IE rotations are shown in Figure 3 and Figure 4 respectively along with the experimental results. The AP displacement for new and experimentally simulated patellae (Figure 3) follow similar trend to the FE rotation (Figure 1) with maximum anterior displacement at highest flexion rotation. The maximum AP displacement (Figures 3) for an experimentally simulated and new patella specimens was 3.0 and 4.3 mm respectively. The difference in the maximum AP displacement is attributed to the wear of material during experimental simulation. AP translation is in phase with FE rotation and increases with the curvature of the femoral component.

The IE rotation (Figure 4) plot from the computational model for new patellae varies from -4^{o} externally to 1^{o} internally. However, the same plot for experimentally simulated patellae was constant at 1^{o} external rotation. The difference can explained by the presence of conforming contact between experimental simulated patella and the femoral component. The new patella buttons has higher tilt due to non-conforming nature. The maximum external rotation of 5.2^{o} was observed in computational and experimental studies for these buttons. This was the maximum IE rotation obtained in this study. Further rotation (IE rotation > 5.2^{o}) led to patella slip. Hence, the IE rotation was restricted to 5.2^{o}.

IE rotation is highly dependent on the ML displacement. Higher medial displacement resulted in higher medial torque (external rotation) as shown in Figure 5.

The change in ML displacement from centre to medial resulted in an external torque which led to increase in external rotation starting at 60% gait cycle. The external rotation is maintained till the end of the gait cycle (Figure 4) i.e. until the patella has medial displacement. Henceforth, the pull (dead weight placed on lateral side) due to the resistance from medial retinaculum at the beginning of the corresponding gait cycle influencing internal rotation till 60% gait cycle. Disturbances due to restriction on the movement of the PFJ fixtures were noticed between 70% and 80% of gait cycle. The IE rotation plot for the experimentally simulated specimens was constant external rotation of 1^{o} due to conformity of the patella specimen to the femoral counterpart as a result of wear simulation.

Kinematic comparison with literature at different boundary conditions.

The AP displacement was found to be approximately 7mm in the literature [12] as compared to 5 mm displacement in the current study (Figure 3). Ostermeier and co-authors [12] worked on the difference between AP displacements on Interax ISA prosthesis with resurfaced or non-resurfaced patella. The higher depth of femoral groove gave the additional AP translation at highest flexion angle._The presence of either resurfacing or non-resurfacing did not affect the AP displacement at 20^{o} flexion.

IE rotation has been measured by few authors in the past [10-11, 30]. They have reported a variation of internal rotation varying from 0 to -4^{o} at initial knee flexion angles. However, change in flexion resulted in increase of external rotation in current study. The absence of knee ligaments in the simulator as compared to natural knee could create this difference.

A comparison of current gait cycle study with constrained ML displacement defined as condition 1 was made with previous investigations of Halloran et al. [15-16] and Ellison [27] as shown in Figure 6. The tilt at condition 1 varied from 0.5 to -4.5^{o} as compared to average tilt by previous investigators varying from 1 to -4^{o}. The tilt obtained from new PFC sigma round dome patella was not different from the value obtained by Halloran *et al.,* [15-16] and Ellison [27].

For the conditions when ligament force and uncontrolled ML displacement were included, comparison with the literature was based on natural knee [21]. High internal rotation for PFC sigma round dome patella (conditions 3) was observed in comparison to kinematics of natural knee. As compared to PFC sigma, the tilt in natural knee was -5 to -8^{o} laterally. There was no similarity between the trends. The presence of other knee ligaments may have led to difference in the tilt. In addition, the kinematics reported in Lafortune and Cavanagh [21] was only limited to one volunteer. Hence, more investigations must be performed for a valid comparison. The removal of ligament force (condition2) did not affect the tilt. The variation of tilt was from -4 to 2^{o}.

Sensitivity analysis

AP translation increased and tilt decreased as the input parameters were changed from actual to ideal conditions for uncontrolled and controlled ML displacement. The simulator followed the actual kinematic due to presence of pneumatic motors. As the actual kinematics was less than the ideal kinematic, the value of AP translations was 96% lower as compared to ideal scenario. FE and superior inferior displacement when lower in actual kinematics led to a decrease in AP translation.

With increase in 5% of the input kinematics, the AP and tilt doubled and decreased by 60% respectively. Conversely, AP and tilt decreased and increased with a decrease in input kinematics by 5%.

Frictional coefficient had an adverse effect on tilt; with an increase (0.1) or decrease in friction (0.01) lead to a stiffer joint bearing and hence, a minimum 200% change in tilt were observed. AP displacement did not vary with change in frictional coefficient. The time step had no effect on the tilt nor the AP displacement. The frictional contact was effective when conformity of the joint in any motion was higher. In AP displacement, there was point/line or lower surface contact. However, the tilt had high surface contact and hence, tilt was affected due to change in the frictional coefficient.

The increase of force coefficient led to decrease in deformation and conformity increases. Hence, tilt was found inversely proportional to the force coefficient. However the change in kinematics was lower than 19%. With increase in stiffness, the deformity is lower, hence conformity decreases and a higher tilt by 20% was observed. The tilt were found inversely proportional to stiffness coefficient.

Conclusion

A good overall agreement between the computational prediction and the experimental data was obtained for patella femoral kinematics. The ML displacement was dependent largely on articular geometry, flexion angle and axial load. AP displacement and tilt were dependent on shift in medial direction and axial load.

Good relation between the model and the past studies were observed when the ligament load was absent and the ML displacement was controlled. The model was however sensitive to ±5% change in kinematics, frictional, force and stiffness coefficients and insensitive to time step.

Conflict of interest statement

The authors have no conflicts to report.

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