An in-vivo experimental evaluation of Abaqus Knee Simulator for Total Knee Replacement

[1]Liming Shu, [2]Jiang Yao, [1]Ko Yamamoto, [1]Mamoru Mitsuishi, [1]Naohiko Sugita

1 Department of Mechanical Engineering, School of Engineering, University of Tokyo, Tokyo, Japan

2 Dassault Systemes Simulia Corp., Johnston- RI, USA

***Abstract**: A large number of experimental in-vitro pre-clinical testing devices have been used in the evaluation of new implant designs for total knee replacement, but it is time-consuming and cost prohibitive to evaluate hundreds of design variations during design phase. In-silico knee simulator has provided an efficient way to perform component design evaluations under a variety of dynamic loadings, while it remains a challenge to evaluate the predictive accuracy due to the lack of in-vivo verification. The commercial software Abaqus, Isight and a specialized tool Abaqus Knee Simulator have been widely used to predict implanted knee mechanics. In the paper, a systematical in-vivo experimental evaluation has been carried out in Abaqus Knee Simulator with calibrated soft-tissue properties for gait cycle. A sets of in-vivo measured knee joint loading conditions from grand challenge dataset were used in the study: The quadriceps actuation during gait cycle was calculated by a subject-specific musculoskeletal model. Proportional-Integral-Derivative (PID) controller was concurrent with the knee simulator to track the joint mechanics (flexion angle, compressive joint load, varus-valgus torque, internal–external torque, Anterior–Posterior force) from experiment. The predicted results show a good agreement on tendency and magnitude (Flexion-extension rotation: RMSE=0.56º, R^2=0.99; Internal-external rotation: RMSE=0.68°, R^2=0.93; Anterior-posterior translation: RMSE=1.54mm, R^2=0.89; Medial-lateral translation: RMSE=0.62mm, R^2=0.56) with in-vivo measured fluoroscopic experimental results on tibiofemoral kinematics. This study provides a calibrated total knee replacement workflow in the Abaqus Knee Simulator which can be widely used in preclinical testing and knee prosthesis design evaluation.*

***Keywords**: Abaqus Knee Simulator, Total Knee Replacement; Experimental Validation, Knee kinematics; FEM; Prosthesis*

1. Introduction

The objective of total knee arthroplasty (TKA) is to restore the mechanical function of the knee and enable patients to perform desired daily activities. In order to mimic the biomechanics of a normal knee joint, many kinds of prostheses have been designed and evaluated since the 1950s [1]. However, patients still did not satisfied with the surgical outcomes. It has been reported that 11%–19% of primary TKR patients are unsatisfied with the surgical outcome, while approximately 6% require revision surgery due to operative complications [1–3]. It mainly caused

by long time design cycle of implant and highly standardized geometric structure of the knee implant, while the characteristics of patients differ, resulting in highly variable surgical outcomes. In order to speed up the design efficiency and increase the quality of knee prosthesis, a high fidelity knee simulator, which can reproduce the *in-vivo* knee joint mechanics, is urgent need in design stage. Several studies have developed physical knee simulators for implant evaluation or exploring the knee joint mechanics. The Stanmore knee simulator is a fixed testing rig, which mainly used to test the wear characteristics of the tibiofemoral (TF) joint [4]. The Oxford-type knee rig test, without the hydraulically actuator, has been employed to evaluate six-degree-of-freedom kinematics of the tibiofemoral (TF) and patellofemoral (PF) joints [5]. The Kansas knee simulator included cadaveric knee joint and five actuators, which is capable to simulate the gait cycle and squatting activities. What's more, some *in-vivo* experimental evaluations in patients with equipped knee implant were also performed in recent years [6,7]. However, the number and type of experimental tests is limited by the need of physical parts, obtain cadaveric specimens and substantial time required to carry out each evaluation. Thus, design-phase experimental analysis of component geometry is typically cost and time prohibitive for parametric geometry assessment.
As the limitation of *in-vivo* and *in-vitro* testing, computational knee simulations provide an efficient toolset to perform the evaluation of TKA design under various dynamic loading conditions. Varadarajan et al. [8] used the rigid dynamics method in the commercial LifeMOD KneeSIM software to analyze the kinematic differences among various prostheses. However, it is difficult to simultaneously predict the kinematics and contact mechanics of implant in the model. Explicit dynamic finite element method (FEM) considered a very effective means which is the capable to simultaneously determine the joint kinematics and contact mechanics [9–12]. Godest et al. [13] used FEM to predict the contact mechanics and kinematics during a gait cycle based on the physical setting of the Stanmore knee simulator, while Knight et al. [14] developed a model to simultaneously predict kinematics and wear. Halloran et al. [15] developed FEM of the rigid Kansas knee simulator (KKS), validating both TF and patellofemoral (PF) kinematics from TKR components against analog TKR knee in the experimental simulator. Baldwin et al. [16] developed a series of natural and implanted specimen specific FE models performing a deep knee bend in the KKS, with PF kinematics validated against experimental cadaveric testing. Clare et al. [17] presented a FEM knee model based on KKS interface with PID controller for reproducing repeated *in-vitro* knee joint loading conditions as KKS across different implant design. Recently, Abaqus knee simulator (AKS) (Dassault Systems Simulia Corp., Johnston, USA) highly mimicked the biology construction of normal femorotibial joint (FT) and patellofemoral (PF) joint, which included all ligaments and soft tissues around knee joint. It is capable to analyze kinematics and contact mechanics of TF and PF joint under various loading conditions such as gait, squatting, stairs up and down. In the previous studies using AKS and explicit finite element based knee simulator, there is little information as to whether these simulation studies accurately represent *in-vivo* status, although some models have been evaluated in *in-vitro* testing [16,18–20]. What's more, the quadriceps actuator in the experimental setup and knee simulator, a feedback control system were used to actuate the quadriceps muscle in order to match a hip flexion and knee flexion profile. However, it has been found the quadriceps force is higher than in vivo condition during daily activity [20]. The musculoskeletal model, which is comprised of a skeleton consisting of rigid body segments (bones) connected by joints and muscle-tendon (MT) units, is believed to be an accurate and advance tool that considers the individual characteristics of the patients for muscle force calculation and joint mechanics analysis [21–23]. Hence, the purposes of this study

are first, create a musculoskeletal integrate force-driven computational knee methods for joint kinematics and contact mechanics analysis and then, to validate the accuracy of the tibiofemoral kinematics with *in-vivo* measured fluoroscopic experimental results during a treadmill gait cycle.

2. Methods and materials

2.1 Subject data

The patient data used in the research were obtained from the sixth grand challenge dataset from the grand challenge competition to predict *in-vivo* knee loads [6]. The experimental data were obtained from a male subject (DM: age—83, height—172 cm, and body weight—70kg), who took a posterior cruciate-retaining (PCR) total knee replacement (TKR) on his right knees. An instrumented implants have been used on the subject. The 3 DOFs force and 3 DOFs moment can be in vivo measured during the movement of subjects. The 3D geometries of the prostheses were provided in geometric stereo lithography (STL) format.

2.2 Simulator workflow

Abaqus knee simulator (AKS) is an explicit solver based dynamic FE knee simulator (Simulia, Johnston, RI, USA), as shown in Figure 1. The physical component and parameter of AKS is mainly based on previously published researches related to KKS [15,16,20,24]. The femur and tibia bone model were created from CT image and set as rigid triangular shell elements (R3D3. The femoral component and tibial tray were modeled as rigid bodies owing to their Young's modulus, which were meshed using the C3D4 tetrahedron with a 1.5-mm element length. The tibial insert component was modeled as a non-linear elastic–plastic material (UHMWPE) with a modulus of elasticity of 463 MPa and a Poisson's ratio of 0.46 using the experimental stress–strain data. It was meshed using 10-node modified quadratic tetrahedral elements (C3D10M) with a 1.3-mm element length based on the convergence study. The soft tissue related to PF joint structures including the patellar tendon (PT) and quadriceps muscle (divided into rectus femoris (RF), vastus intermedius (VI), vastus lateralis longus (VLL), vastus lateralis obliquus (VLO), vastus medialis longus (VML) and vastus medialis obliquus (VMO) bundles) were modeled as 2D fiber reinforced membrane (M3D4R) and non-linear spring elements (CONN3D2). TF soft tissue included medial, lateral and posterior cruciate ligament (MCL, LCL and PCL), oblique popliteal ligament (OPL), popliteofibular ligament (PFL), medial and lateral posterior capsule ligament (PCAPM and PCAPL), which were also set as 2D fiber reinforced membrane (M3D4R) and non-linear spring elements (CONN3D2). The anterior cruciate ligament (ACL) was sacrificed during the surgery and, therefore, not modeled. The flexion-extension (F-E) rotation were directly controlled by the input motion. There are five loading applied on knee simulator, include quadriceps force, anterior-posterior (A-P) force, inferior-superior (I-S) force, varus-valgus (V-V) torque and internal-external (I-E) torque, as shown in Figure 1 (c). The stiffness-force relationship of the ligaments is modelled as to produce a nonlinear elastic characteristic with a slack region, which has been widely used in previous researches. [19,25,26]. Ligament and tendon material properties were established in separate planar analyses to match published experimental uniaxial force–displacement data [16,20,27,28], as shown in Table 1.

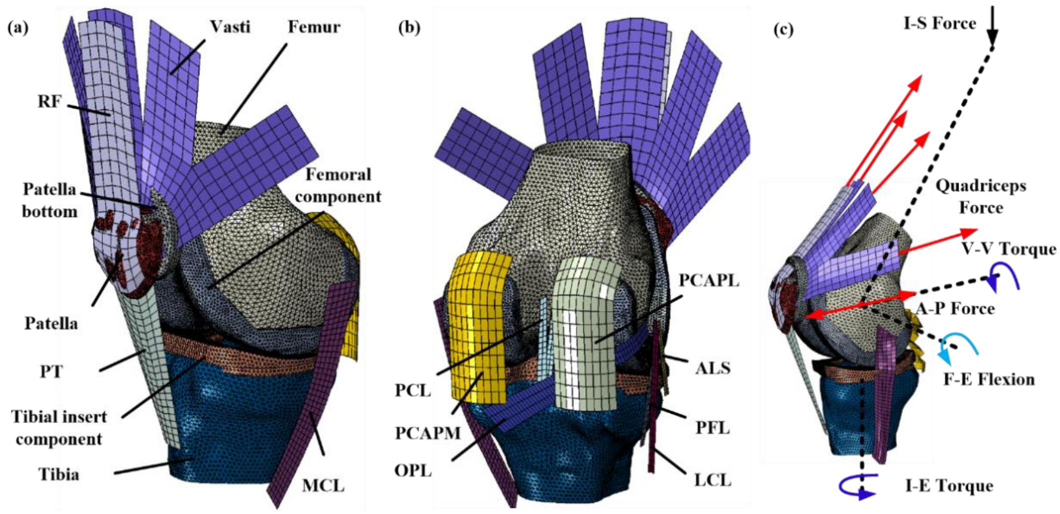

Figure.1. (a) anterior view of 3D finite element model of knee joint, including the bones, muscles, TKR implant, (b) posterior view of knee model with ligaments, and (c) loading conditions during gait inputs applied.

Table 1. Material constants for ligaments and tendons in AKS model

Ligaments	aLCL	mLCL	pLCL	aMCL	mMCL	pMCL	aPCL	mPCL
Prestrain	0.989	1.026	1.024	1.039	1.029	1.03	0.8	1.0
Stiffness (N/mm)	20.8	20.8	20.8	58.4	58.4	58.4	120	57.0
Ligament	ALS	PFL	OPL	PCAPM	PCAPL	PT	RF	VASI
Prestrain	1.038	0.975	0.93	1.039	1.039	1	1	1
Stiffness (N/mm)	38.3	9.4	123.1	45.0	45.0	5000.0	5000.0	5000.0

The whole workflow were presented in Figure 2. The quadriceps force were calculated by a FE based lower limb MS model. Description of specifics of FE based MS model has been published previously [23,29], and for brevity it is repeated here. The center of mass, muscle parameters, and muscle origin, insertion and via-point coordinates were obtained from Biomechanical Data Resource (http://isbweb.org/data/) [30]. Subject-specific MS model was developed by scaling and positioning the generic model using the first frame of the motion capture data. The objective function was to minimize the least square error of the marker coordinates on each segment with the joint constraints. The Flexion-extension rotation of knee joint were calculated by inverse kinematics which was executed by a combination of shell elements and the distributed coupling feature with joint constraints to remove the skin motion component from the motion data and was then solved using the implicit quasi-static solver in Abaqus [31]. The quadriceps force was calculated by muscle force optimization which was based on minimize the metabolic energy of the muscles subject to the constraints, including the positive muscle force and moment equilibrium equations for the flexion axis of each joint.

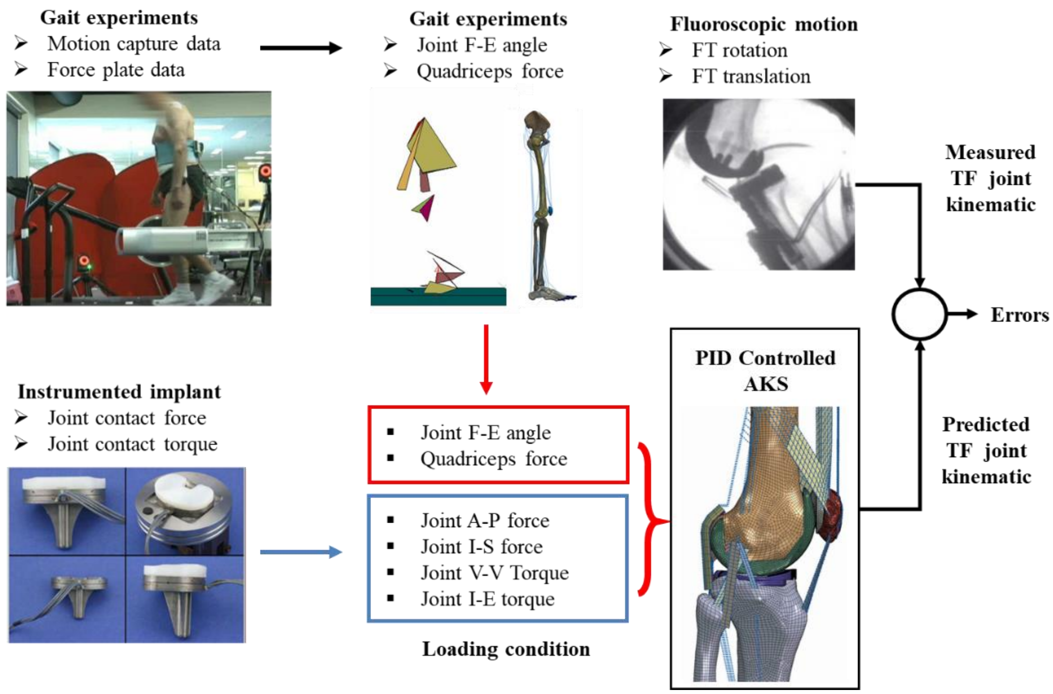

Figure.2. Diagram showing how modeling and experiments were integrated to determine knee joint kinematics during treadmill gait. Motion capture and force plate data were input into a FE-based MS model to calculate quadriceps force and flexion-extension (F-E) rotations. The anterior-posterior (A-P) force, inferior-superior (I-S) force, varus-valgus (V-V) torque and internal-external (I-E) torque were in-vivo measured by instrumented implant. The measured joint A-P force, IS force, V-V torque, I-E torque, and calculated quadriceps force, joint F-E rotation were then input into a PID controlled AKS model to determine the TF and PF joint kinematics. However, due to lack of PF joint fluoroscopic data, only the TF joint kinematics were evaluated.

The knee was represented as a hinge joint in the MS model, the moments and force in coronal and transverse plane could not be accurately predicted. Hence, anterior-posterior (A-P) force, inferior-superior (I-S) force, varus-valgus (V-V) torque and internal-external (I-E) torque were *in-vivo* measured by instrumented implant and input into the AKS simulator. What more, it has been know that changing the geometry of the TKR components, implant positioning or limb alignment will influence the forces, torques and kinematics during the activity. The ligament force will also change the loading condition of the implant. For instance, the specific *in-vivo* measured V-V torque was added as the loading condition in TF joint. However, the V-V moment on tibia insert is still be changed with surrounding ligament force during the simulations. Hence, the actual joint loading condition is different from *in-vivo* loading. In order to reproduce *in-vivo* knee joint loading conditions in current AKS models, a flexible proportional–integral–derivative (PID) control system was developed and interfaced with the FE model through an Abaqus/Explicit user subroutine (called VUAMP), which is coded in the Fortran language. There are four sensors were settled in AKS model to monitor each load output (A-P force, I-S Force, I-E torque and V-V torque). In the subroutine, the instantaneous loading outputs are compared to a loading profile which the FE model is trying to match. The output calculated by the controller is subsequently fed back and applied to the FE model (Figure 3). A similar method could be found in previous publication and found that the derivative terms had minimal influence on the performance of the controller [17]. Hence, in order to provide stable performance of the control system, the *Kp* and *Ki* was automated tuned through Isight, commercially available software for the automation of simulation process flows (Simulia, Johnston, RI, USA), which interfaces seamlessly with Abaqus and provided easy comparison of Abaqus outputs (sensors) and target profiles. A Latin Hypercube algorithm, demonstrated as an efficient sampling method, was used to sample the PID parameter values (McKay et al. 1979), with the optimal criteria based on minimizing the root-mean-square error (RMSE) between target profiles and measurements from the sensors during the simulation.

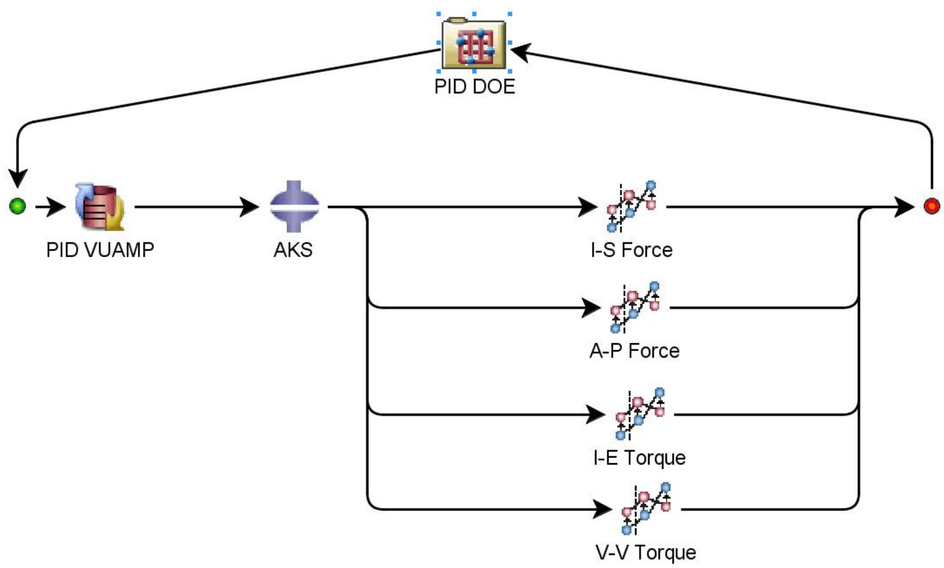

Figure.3. Schematic of workflow implanted in Isight™ to optimize the tune parameters in PID controller. A DOE Latin Hypercube technique is implemented to tune the control parameters to match the target data.

2.3 Model evaluation

In this paper, a slow walking speed (0.5m/s) treadmill gait of fluoroscopy trial from the dataset was used to obtain an estimate of TF joint kinematics, namely F-E rotation, I-E rotation, Ab-Ad Rotation, A-P translation, M-L translation, and I-S translation. The simulation results were sampled on a 0–100% gait cycle from heel strike to the subsequent heel strike. The predicted quadriceps muscle activation levels were compared with the EMG results. In order to verify any differences between the model prediction and experimental measurements of joint kinematics, the measurement indexes of the root-mean-square error (*RMSE*) and squared Pearson correlation coefficient (r^2).

3. Results and discussion

The predicted activation patterns for quadriceps muscle, which calculated by MS model, during treadmill gait trial are shown with comparison to experimental EMG results in Figure 4. Good temporal agreement can be observed on the muscles, although the vastus lateralis activation is predicted to be higher and more prolonged during the stance phase and swing phase, which may be explained by the inaccuracy of the PF kinematics.

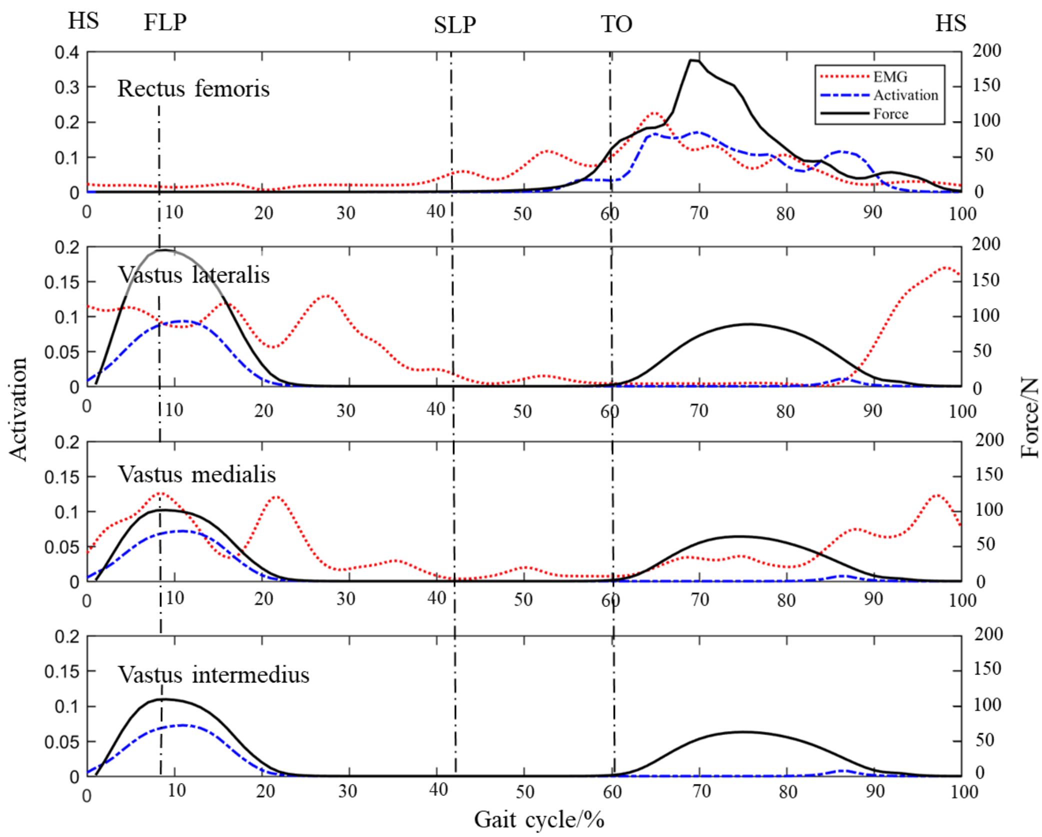

Figure 4. Comparison between quadriceps muscle activation patterns with EMG-measured muscle activation pattern during single gait cycle. The EMG results of vastus intermedius muscle is not available in the experiment, though the vastus intermedius muscle was not evaluated here. (**Note: heel strike (HS); first loading peak (FLP); second loading peak (SLP); toe off (TO)**)

A 20-trial Latin Hypercube technique, with three variables (K_p and K_i terms) for each actuator, provided adequate tuning of the PID control parameters. Good tracking accuracy between AKS simulation and experimental results could be found in Figure 5. RMSE between simulation and target profiles were 28 N, 19 N, 0.3 Nm and 0.9Nm for I-S force, A-P force, I-E torque and V-V torque, respectively, throughout a slow speed treadmill gait cycle. All of the squared Pearson correlation coefficient (r^2) are larger than 0.90, except in V-V torque (0.52) which PID gain should be further optimized.

Figure 6 provides the model predicted and measured TF joint kinematics for the low speed treadmill gait trials. The TF F-E rotation, which predicted by inverse dynamics of MS model, had an average RMSE from experimental results of 0.56^{o}. The A-P, I-S, M-L translation were also well matched with an average RMSE of 1.54 mm, 0.68 mm and 0.62 mm, respectively, while the I-E and V-V rotations matched experimental results with an average RMSE of 0.68^{o} and 2.7^{o}. The simulated PF joint kinematics were presented in Figure 7. However, due to the lack of experimental results, the PF joint kinematics were not verified here. In general, good agreement was found between the experimental and model-predicted kinematics for both rigid body and deformable representations. Translational degrees of freedom averaged approximately 1 mm of the RMSE. RMSE for rotational degrees of freedom averaged near 2 deg. All experiment translational and rotational trends were captured with the model.

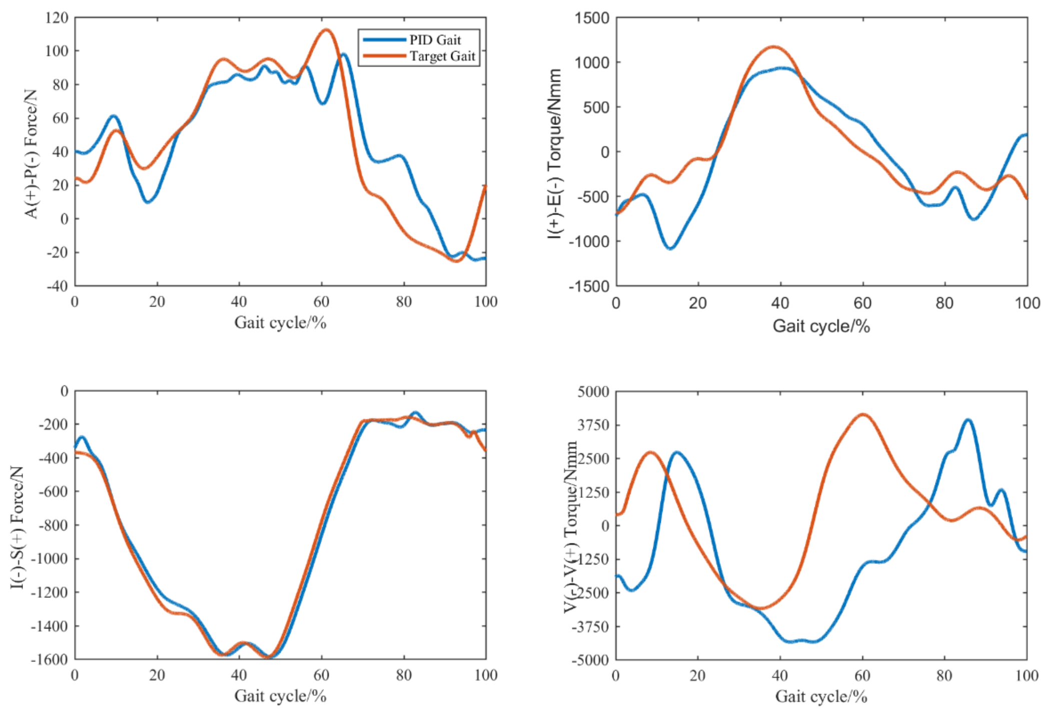

Figure.5. Comparison of target profiles with PID-controlled joint load during a low speed treadmill gait cycle.

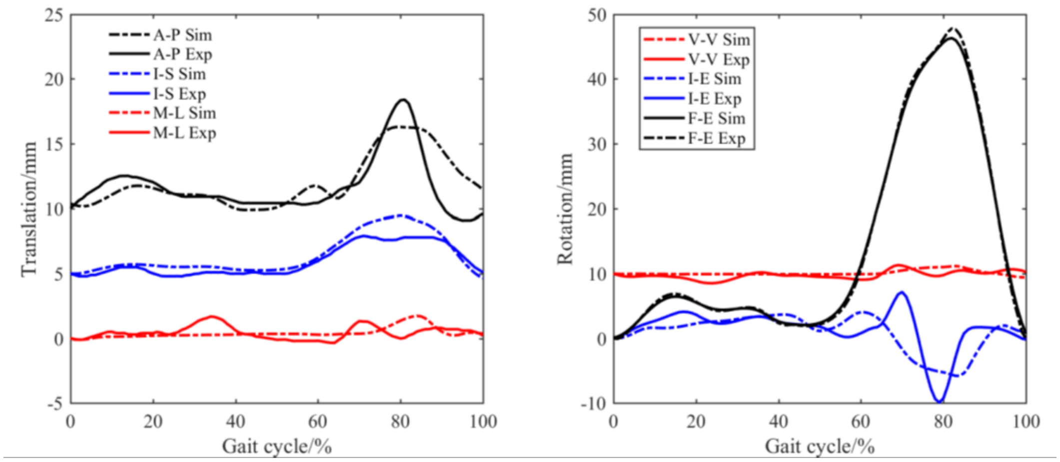

Figure.6. Comparison of simulated experimental results of TF joint during low speed treadmill gait cycle.

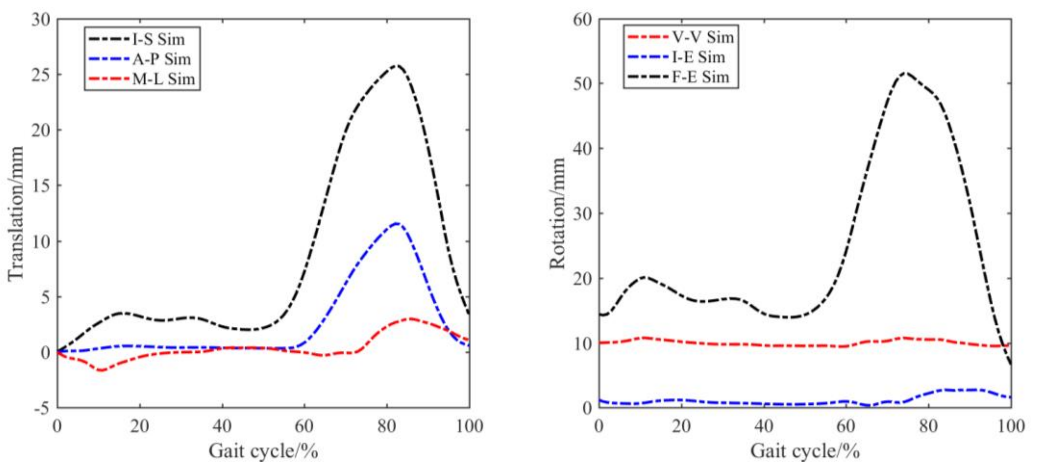

Figure.7. Simulated results of PF joint during low speed treadmill gait cycle.

4. Conclusion

We have introduced a framework based on concurrent Abaqus knee simulator and patient-specific musculoskeletal (MS) technology for in vivo predicting of kinematics and contact mechanics within TKR knee joint and verified by in vivo experiment. The specific knee joint was interacted with muscle, ligament, and joint contact forces under the context of dynamic multi joint movement. It is initially shown that the concurrent framework can be used to predict knee kinematics that is well consistent with those measured directly by measured fluoroscopic experimental results during a treadmill gait cycle from sixth "grand challenge competition to predict in vivo knee loads" database. Proportional-Integral-Derivative (PID) controller was concurrent with the knee simulator to track the joint mechanics (flexion angle, compressive joint load, varus-valgus torque, internal–external torque, Anterior–Posterior force) from experiments. The predicted results showed a good agreement on tendency and magnitude (Flexion-extension rotation: *RMSE*=0.56º, R^2=0.99; Internal-external rotation: *RMSE*=0.68°, R^2=0.93; Anterior-posterior translation: RMSE=1.54mm, R^2=0.89; Medial-lateral translation: *RMSE*=0.62mm, R^2=0.56) with in-vivo measured fluoroscopic experimental results on tibiofemoral kinematics. There are still a number of limitations to be considered in our low limb model. Firstly, it has been known that there are large intra- and inter- subject variations for properties of ligaments, so the variability should also be considered. Secondary, the kinematics on PF joint was not capable to be verified in the study due to lack of experimental data. A further systematically experimental verification will be conducted in the next research.

This study provides a calibrated total knee replacement workflow in the Abaqus Knee Simulator which can be widely used in preclinical testing and knee prosthesis design evaluation. What's more, it potentially provided a way to create subject specific loading condition that can be used for Abaqus Knee simulator. In the next step, we will concurrent the wear analysis with the Abaqus Knee Simulator, which will make it capable to predict the service life of implant design.

5. Acknowledgment

The authors thank the knee load grand challenge team, led by B. J. Fregly, Ph.D. and Darryl D'Lima, M.D., Ph.D. (https://simtk.org/home/kneeloads), for providing the images, biomechanical measures, and in vivo knee contact forces used in this study. This research is supported by JSPS KAKENHI Grant Number 16H05874.

6. References

[1] Bourne RB, Chesworth BM, Davis AM, Mahomed NN, Charron KDJ. Patient satisfaction after total knee arthroplasty: Who is satisfied and who is not? Clin Orthop Relat Res 2010;468:57–63. doi:10.1007/s11999-009-1119-9.

[2] Pabinger C, Berghold A, Boehler N, Labek G. Revision rates after knee replacement: Cumulative results from worldwide clinical studies versus joint registers. Osteoarthr Cartil 2013;21:263–8. doi:10.1016/j.joca.2012.11.014.

[3] Von Keudell A, Sodha S, Collins J, Minas T, Fitz W, Gomoll AH. Patient satisfaction after

primary total and unicompartmental knee arthroplasty: An age-dependent analysis. Knee 2014;21:180–4. doi:10.1016/j.knee.2013.08.004.

[4] DesJardins JD, Banks SA, Benson LC, Pace T, LaBerge M. A direct comparison of patient and force-controlled simulator total knee replacement kinematics. J Biomech 2007;40:3458–66. doi:10.1016/j.jbiomech.2007.05.022.

[5] Elias JJ, Wilson DR, Adamson R, Cosgarea AJ. Evaluation of a computational model used to predict the patellofemoral contact pressure distribution. J Biomech 2004;37:295–302. doi:10.1016/S0021-9290(03)00306-3.

[6] Fregly BJ, Besier TF, Lloyd DG, Delp SL, Banks SA, Pandy MG, et al. Grand challenge competition to predict in vivo knee loads. J Orthop Res 2012;30:503–13. doi:10.1002/jor.22023.

[7] Sandholm A, Schwartz C, Pronost N, De Zee M, Voigt M, Thalmann D. Evaluation of a geometry-based knee joint compared to a planar knee joint. Vis Comput 2011;27:161–71. doi:10.1007/s00371-010-0538-7.

[8] Varadarajan KMM, Zumbrunn T, Rubash HE, Malchau H, Li G, Muratoglu OK. Cruciate Retaining Implant With Biomimetic Articular Surface to Reproduce Activity Dependent Kinematics of the Normal Knee. J Arthroplasty 2015;30:2149–2153.e2. doi:10.1016/j.arth.2015.06.018.

[9] Kang KT, Park JH, Lee K Il, Shim YB, Jang JW, Chun HJ. Gait cycle comparions of cruciate sacrifice for total knee design.-explicit finite element. Int J Precis Eng Manuf 2012;13:2043–9. doi:10.1007/s12541-012-0269-y.

[10] Naghibi Beidokhti, Hamid; Khoshgoftar, Mehdi; Sprengers, André; Van den Boogaard, Ton Janssen D, Verdonschot NV. A Comparison Between Dynamic Implicit And Explicit Finite Element Simulations Of Native Knee Joint. Med Eng Phys 2016;38:1123–30.

[11] Pal S. Explicit Finite Element Modeling. 2008.

[12] Qian Z, Ren L, Ding Y, Hutchinson JR, Ren L. A dynamic finite element analysis of human foot complex in the sagittal plane during level walking. PLoS One 2013;8:1–10. doi:10.1371/journal.pone.0079424.

[13] Godest AC, Beaugonin M, Haug E, Taylor M, Gregson PJ. Simulation of a knee joint replacement during a gait cycle using explicit finite element analysis. J Biomech 2002;35:267–75. doi:10.1016/S0021-9290(01)00179-8.

[14] Knight LA, Pal S, Coleman JC, Bronson F, Haider H, Levine DL, et al. Comparison of long-term numerical and experimental total knee replacement wear during simulated gait loading. J Biomech 2007;40:1550–8. doi:10.1016/j.jbiomech.2006.07.027.

[15] Halloran JP, Clary CW, Maletsky LP, Taylor M, Petrella AJ, Rullkoetter PJ. Verification of predicted knee replacement kinematics during simulated gait in the Kansas knee simulator. J Biomech Eng 2010;132:81010. doi:10.1115/1.4001678.

[16] Baldwin MA, Clary C, Maletsky LP, Rullkoetter PJ. Verification of predicted specimen-specific natural and implanted patellofemoral kinematics during simulated deep knee bend. J Biomech 2009;42:2341–8. doi:10.1016/j.jbiomech.2009.06.028.

[17] Fitzpatrick CK, Baldwin M a, Clary CW, Maletsky LP, Rullkoetter PJ. Evaluating knee replacement mechanics during ADL with PID-controlled dynamic finite element analysis. Comput Methods Biomech Biomed Engin 2012;5842:3. doi:10.1080/10255842.2012.684242.

[18] Fitzpatrick CK, Komistek RD, Rullkoetter PJ. Developing simulations to reproduce in vivo fluoroscopy kinematics in total knee replacement patients. J Biomech 2014;47:2398–405. doi:10.1016/j.jbiomech.2014.04.024.

[19] Marra MA, Vanheule V, Rasmussen J, Verdonschot NJJ, Andersen MS, Fluit R, et al. A subject-specific musculoskeletal modeling framework to predict in vivo mechanics of total knee arthroplasty. J Biomech Eng 2014;137:20904. doi:10.1115/1.4029258.

[20] Baldwin MA, Clary CW, Fitzpatrick CK, Deacy JS, Maletsky LP, Rullkoetter PJ. Dynamic finite element knee simulation for evaluation of knee replacement mechanics. J Biomech 2012;45:474–83. doi:10.1016/j.jbiomech.2011.11.052.

[21] Delp S, Anderson FC, Arnold AS, Loan P, Habib A, John CT, et al. OpenSim: Open-source software to create and analyze dynamic simulations of movement. IEEE Trans Biomed Eng 2007;54:1940–50. doi:10.1109/TBME.2007.901024.

[22] Nakamura S, Tian Y, Tanaka Y, Kuriyama S, Ito H, Furu M, et al. The effects of kinematically aligned total knee arthroplasty on stress at the medial tibia: A case study for varus knee. Bone Joint Res 2017;6:43–51. doi:10.1302/2046-3758.61.BJR-2016-0090.R1.

[23] Shu L, Sugita N, Saraswat P, Yao J, Yoshizaki R, Yamamoto K, et al. A concurrent patient-specific musculoskeletal and finite element modeling framework for predicting in vivo kinematics and contact mechanics of total knee replacement. Sci. Age Exp., 2017, p. 658–71.

[24] Ali AA, Shalhoub SS, Cyr AJ, Fitzpatrick CK, Maletsky LP, Rullkoetter PJ, et al. Validation of predicted patellofemoral mechanics in a fi nite element model of the healthy and cruciate-de fi cient knee. J Biomech 2015;49:1–8. doi:10.1016/j.jbiomech.2015.12.020.

[25] Kang K-T, Koh Y-G, Jung M, Nam J-H, Son J, Lee YH, et al. The effects of posterior cruciate ligament deficiency on posterolateral corner structures under gait- and squat-loading conditions. Bone Jt Res 2017;6:31–42. doi:10.1302/2046-3758.61.BJR-2016-0184.R1.

[26] Blankevoort L, Kuiper JH, Huiskes R, Grootenboer HJ. Articular contact in a three-dimensional model of the knee. J Biomech 1991;24:1019–31. doi:10.1016/0021-9290(91)90019-J.

[27] Griffith CJ, Wijdicks CA, LaPrade RF, Armitage BM, Johansen S, Engebretsen L. Force measurements on the posterior oblique ligament and superficial medial collateral ligament proximal and distal divisions to applied loads. Am J Sports Med 2009;37:140–8. doi:10.1177/0363546508322890.

[28] LaPrade RF, Tso A, Wentorf FA. Force measurements on the fibular collateral ligament, popliteofibular ligament, and popliteus tendon to applied loads. Am J Sports Med 2004;32:1695–701. doi:10.1177/0363546503262694.

[29] Yao J, Saraswat P, Chinnakonda M, Bertucci R, Oancea V, Hurtado JA, et al. Using Abaqus and Isight to simultaneously determine muscle forces , joint kinematics and strain / stresses in soft tissues during walking. Proc. 7th World Congr. Biomech., 2014, p. 1–2.

[30] Delp SL, Loan JP, Hoy MG, Zajac FE, Topp EL, Rosen JM. An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. IEEE Trans Biomed Eng 1990;37:757–67. doi:10.1109/10.102791.

[31] Chinnakonda M, Saraswat P, Yao J, Oancea V, Hurtado J, Sett S. Generating kinematically consistent data for musculoskeletal inverse dynamic analyses using finite element tools. Am. Socieity Biomech., 2013, p. 367–8.