Systematic review of computational modelling for biomechanics analysis of total knee replacement

Liming Shu, Shihao Li, Naohiko Sugita

Department of Mechanical Engineering, School of Engineering, The University of Tokyo, 7-3-1 Hongo, Bunkyo-ku, Tokyo 113-8656, Japan
E-mail: l.shu@mfg.t.u-tokyo.ac.jp

Abstract: In vitro and in vivo testing can provide insight into knee joint mechanics and implant performance. However, these methods are costly and time-consuming, which always limits their widespread use during the design stage of the implant. This review presents a critical analysis of computational modelling (in-silicon) techniques including (i) development of a generic model of total knee replacement (TKR) and application of material properties, loading, and boundary conditions; (ii) design and execution of computational experiments; and (iii) practical applications and significant findings. The results show that the generic model and techniques provide significant insight into the general performance of TKR but have limited explicit validation. The introduction of design-of-experiments, probabilistic, and neural network methodologies in computational modelling has enabled simulation at the population level. Further advances in subjective modelling appear to be limited, mainly because of the lack of subjective materials and boundary conditions. Computational modelling will increasingly be used in the preclinical testing and design of TKR. This modelling should include subjective, multi-scale, and closely corroborated analyses to account for the variability of TKR.

1 Introduction

Osteoarthritis and rheumatoid arthritis cause ∼44.7% of the pain experienced by humans, and hinder daily activities such as walking and stair climbing. [1]. Total knee replacement (TKR) surgery was first performed in 1968. In this surgery, part of the damaged bone was excised and replaced with a metal joint implant [2]. The use of TKR to address knee disease has grown exponentially in the past 60 years and is expected to continue increasing by 601% from 2005 to 2030 in the USA [3]. Improvements in surgical materials and techniques have significantly enhanced the effectiveness of this procedure. However, 11–19% of primary TKR patients reportedly remain dissatisfied with the surgical outcome, and ∼6% require revision surgery owing to operative complications and failure of the implant [4, 5].

Although in vivo clinical testing provides the most realistic information regarding the biomechanics of the TKR knee, certain types of experiments or data collection methods are difficult or unethical to complete on human subjects. In addition, applying clinical testing for prosthesis evaluation is difficult during the design stage. Therefore, various types of in vitro knee simulators have been developed to predict the joint kinematics during the design of the knee prosthesis [6–11]. However, the number and type of experimental tests are limited because they require physical parts and cadaveric specimens and are very time-consuming. Therefore, experimental analysis of component geometry during prosthesis design is in general difficult in terms of cost and time. Computational knee simulations provide an efficient toolset to overcome the limitations of in vivo and in vitro testing, to accelerate the design, and to improve the quality of the knee prosthesis. These simulations allow the evaluation of a TKR design under various dynamic loading conditions to address clinical design and to evaluate ligament performance, laxity, kinematics, wear, contact mechanics, cement fatigue, bone remodelling-induced stress shielding, muscle force, and joint loading. The simulations also allow comparative analyses of TKR designs [10].

Computational modelling of TKR can be approximately divided into two types: macro-mechanics studies by musculoskeletal modelling (MSM) (multi-body dynamics-based modelling) and micro-mechanics studies by finite element modelling (FEM). MSM encompasses a skeleton consisting of rigid body segments (bones) connected by joints and muscles/tendons and describes bodies through kinematic relations. This has a high potential for body-level mechanics analysis as joint loading and muscle activation analysis [12–17]. FEM is based on the discretisation of complex geometries and boundary conditions into a simple collection of subdomains (elements). Partial differential equations based on the principle of virtual work describe the mechanics of each element [18, 19]. FEM has provided an efficient way to understand the time-dependent tissue-level and implant-level mechanics of TKR [20–24]. Both methods have been developed simultaneously during the last 50 years. Recently, concurrent analysis has attracted interest in multi-scale mechanics analysis of patients with orthopaedic devices [25–28]. Moreover, computational modelling not only provides significant insight into the mechanics of TKR but also offers a platform to implement parametric analysis and optimisation [29–32].

2 Literature search approach

A great deal of research has been focused on computational mechanics analysis of TKR to accelerate TKR design and obtain insight into the biomechanics of clinical problems. Fig. 1 shows the trends in the number of publications per year from 1968 to 2018 related to computational analysis of TKR. The keywords ‘finite element (FE)’, ‘musculoskeletal (MS) model’, and ‘computational’ within the ‘TKR’ or ‘total knee arthroplasty (TKA)’ field were searched in article titles, abstracts, and keywords in the Web of Science database [33]. The publications resulting from this search over the previously mentioned 60-year period were then refined manually to define a more specific subject area. The first article that applied a computational method to TKR was published in 1977 by Chao et al. [34]. The number of
Publications on FEM-based TKR analysis surged from the year 2000 onwards, whereas MSM-based work started in 2005. Computational modelling also expanded the analysis of TKR from static analysis to full dynamics analysis with patient-specific loading conditions. The following review is intended for readers already familiar with the detailed methodology of FEM and MSM, although only modelling methodology, application, components, and validation will be summarised. On account of a large number of published studies, research studies on a natural knee joint will be excluded from this review.

3 Critical analysis of computational modelling of TKR

3.1 Development of mimicking computational model of TKR

Based on the complexity of knee joint models in FEM and MSM, we divided the research on computational modelling of TKR into three periods: 1977–1993, static analysis for bone and implant stress distribution; 1994–2009, dynamic analysis for kinematics, contact mechanics, and wear of the tibiofemoral (TF) joint under a simplified generic condition; and 2009–2019, dynamic analysis with complex knee joint constraints under in vivo joint loading conditions. Fig. 2 provides a historical perspective on the progression of FEM and MSM for TKR. The typical computational models focused on TKR mechanics are listed in Table 1 in chronological order.

3.1.1 FEM-based computational modelling of TKR: With the advent of FEM-based methods and increased computational power, early studies on TKR—until 2000—focused mainly on the stress distributions of the tibial insert (TI) and bone under static loading conditions with a small number of elements. The first application of three-dimensional (3D) FEM for TKR was reported by Chao et al. [34]. The goal was to understand the stress distribution of approximate models of TI under certain axial static loadings. Other studies by Bartel et al. [70] and Lewis et al. [71] followed. Cheal et al. [42] used FEM to analyse the influence of cement injection depth of TI on bone stress distribution in the surrounding trabecular and cortical bone. Tissakht et al. [72] and Van Lenthe et al. [73] analysed stress shielding in the distal femur in TKR. Aitih et al. [74] created a FEM-based testing method for fatigue testing of the tibial tray (TT). Perillo-Marcone et al. [75] analysed the influence of tibial alignment in TKR by FEM with static loading and found that valgus tilt of TT reduces the stress on cancellous bone.

In 1994, Sathasivam and Walker [44] first used FEM for the optimisation of bearing surface geometry of TKR considering both the contact on TF and patellofemoral (PF) joints and the dynamic motion. However, the analysis was still performed with constant loading (1000 N). After this study, the same authors [45] developed the first knee simulator of the TF joint with multi-directional force and torque during a gait cycle for subsurface damage analysis of the TI. The conflicting requirements of laxity and conformity in TKR were also highlighted in their FEM study [35]. The loading and boundary conditions defined in these studies were further extended as an ISO standard (ISO 14243-1) in preclinical wear testing of TKR. The soft tissue around the knee joint was simplified to four parallel springs to mimic the natural laxity of the knee joint. Meanwhile, Walker et al. [36] also developed a physical knee simulator (the so-called Stanmore knee simulator) that is widely used nowadays in wear testing. A large number of FE models based on the Stanmore knee simulator have been developed for kinematics, contact mechanics, and wear analysis of TKR. Godest et al. [22] used FEM to predict the contact mechanics and kinematics during a gait cycle based on the physical setting of the Stanmore knee simulator. The computational biomechanics group at the University of Denver has...
| Authors | Physical prototype | Method | Loading | Focus |
|---------|-------------------|--------|---------|-------|
| Chao et al. [34] | TI model | FEM | axial loading | stress distribution of TI component |
| Cheal et al. [42] | TI with the tibia | FEM | axial loading | stress shielding of the tibia |
| Essinger et al. [43] | cadaver-based TKR knee | MSM | flexion-extension (F-E) rotation with constant loading on TF joint | kinetics, contact pressure, and ligament force of PF and TF joints |
| Garg and Walker [38] | cadaver-based TKR knee | MSM | F-E rotation with constant loading on TF joint | kinetics analysis of TF joint with various conformity of TI |
| Sathasivam and Walker [44] | TF joint of TKR | FEM | F-E rotation with constant loading on TF joint | optimisation of sagittal and coronal conformity of TF joint |
| Essinger et al. [43] | TF joint of TKR | FEM | loading during the gait cycle (ISO 14243-1) | subsurface damage, laxity, TI conformity optimisation |
| Piazza and Delp [46] | lower limber | MSM | EMG, marker position | the knee joint contact forces and translation of TKR |
| Godest et al. [22] | Stanmore knee simulator | FEM | loading during the gait cycle (ISO 14243-1) | kinematics analysis of TF joint |
| Garg and Walker [38] | cadaver-based TI knee | MSM | axial force and motion from in-vivo fluoroscopic | kinematics, contact mechanics and ligaments forces |
| Sathasivam and Walker [44, 45] | TF joint of TKR | FEM | loading during the gait cycle (ISO 14243-1) | deformation contact algorithm for efficient MSM calculation |
| Piazza and Delp [46] | lower limber | MSM | EMG, marker position | rigid-body and full-deformable analysis |
| Valerie and Halloran et al. [49] | Stanmore knee simulator | FEM | loading during the gait cycle (ISO 14243-1) | probabilistic finite element prediction of kinematics and contact mechanics |
| Lin et al. [54] | lower limb | MSM | motion data, ground reaction force | understanding the moment arm and sensitivity of each muscle on implant position |
| Lin et al. [51] | lower limb | MSM | motion data, ground reaction force | rigid-body and full-deformable analysis |
| Willing and Kim [29, 56-58] | Stanmore knee simulator | FEM | F-E rotation and fixed axial loading | probabilistic finite element prediction of wear |
| Zhang et al. [128] | lower limb | MSM | motion data and ground reaction force | surrogate contact modelling for improving the computational speed of dynamic contact simulations |
| Gerus et al. [14] | full body | MSM | motion data and ground reaction force | evaluation of joint contact force and muscles forces |
| Dasault Systems {Simulia Corp. [37]} | KKS | FEM | F-E rotation and fixed axial loading | kinetics analysis on TF and PF joint |
| Thelen et al. [63] | lower limb | MSM | motion data, ground reaction force | muscle and contact forces on TF and PF joints during high knee flexion |
| Marra et al. [64] | full body | MSM | motion data and ground reaction force | laxity, wear, and fatigue-based geometrical optimisation of TKR with experimental evaluation |
| Okamoto et al. [85] | KKS (KneeSim) | MSM | F-E rotation and fixed axial loading | kinematics, contact mechanics, and ligament force on TF and PF joint of TKR |
| Chen et al. [86] | full body | MSM | F-E rotation and fixed axial loading | PID control of knee simulator for predicting joint loading |
| Fitzpatrick et al. [90] | full body | MSM | F-E rotation and fixed axial loading | sensitivity analysis of the role of design, surgical, and patient-specific parameters on joint |
| Fitzpatrick et al. [90] | full body | MSM | F-E rotation and fixed axial loading | kinematics during deep squatting |
| Liu et al. [62] | lower limb | MSM | motion data and ground reaction force | understanding the differences of joint kinematics, joint moments, and muscle contributions to |
| Gerus et al. [14] | full body | MSM | motion data and ground reaction force | determining the influence of knee joint geometry on the prediction accuracy of joint contact force |
| Dasault Systems {Simulia Corp. [37]} | KKS | FEM | F-E rotation and fixed axial loading | kinematics, contact mechanics, and ligament force on TF and PF joint of TKR |
| Thelen et al. [63] | lower limb | MSM | motion data, ground reaction force | concurrent prediction of kinematics, contact mechanics, and muscle force on TF and PF joints |
| Marra et al. [64] | full body | MSM | motion data and ground reaction force | concurrent prediction of kinematics, contact mechanics, and ligaments force on TF and PF joints |
| Okamoto et al. [85] | KKS (KneeSim) | MSM | F-E rotation and fixed axial loading | effect of tibial posterior slope on knee kinematics, quadriceps force, and PF contact force |
| Chen et al. [86] | full body | MSM | F-E rotation and fixed axial loading | understanding the effect of alignment error on joint loading during the gait cycle |
| Fitzpatrick et al. [90] | full body | MSM | F-E rotation and fixed axial loading | high mimic 6D knee simulator |
| Zhang et al. [128] | lower limb | MSM | motion data and ground reaction force | patient-specific wear prediction of TKR during the gait cycle |
| Shu et al. [28] | lower limb | MSM | motion data and ground reaction force | concurrent predicting the kinematics, contact mechanics, and muscle force on TF joint during the gait cycle |
| Navacchia et al. [41] | lower limb | MSM | motion data, ground reaction force | developing a computationally efficient strategy to estimate muscle forces in FE MS model |
| Hume et al. [89] | lower limb | MSM | motion data, ground reaction force | a muscle-driven explicit FE-MS model for prediction of muscle force, ligaments force and |
| KKS FEM | F-E rotation and fixed axial loading | concurrent predicting the kinematics, contact mechanics, and muscle force on TF joint during the gait cycle and chair rising. | kinematics during the gait cycle and chair rising. |
presented a number of articles related to a FEM-based knee simulator. Halloran et al. [49] developed rigid-body analyses to reproduce the kinematics and contact mechanics in a fully-deformable-body analysis to study the dynamic behavior of the knee. Fitzpatrick et al. [67] developed a FEM-based six-degrees-of-freedom (6-DOF) knee simulator during various daily activities, applying the Stanmore knee simulator restrictions to the DOFs. The Stanmore knee FE simulator presented in [76] aimed to predict the kinematics and contact mechanics of different implant designs and various daily activities. However, the FEM based on the Stanmore and 6-DOF knee neglects the muscle force and interaction between femur and patella in the model. This can influence the kinematics and contact mechanics of the prosthesis [77, 78]. A higher-fidelity knee simulator including the PF joint, muscle loading, and high-fidelity ligaments constraint is preferred for the sake of a more comprehensive analysis of TKA knee joints.

Understanding the limitation of previous TF knee simulators, Halloran et al. [7] developed a FEM of the rigid Kansas knee simulator (KKS), validating both TF and PF kinematics from TKR components against an analogue TKR knee in the experimental simulator. Baldwin et al. [59] 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. Fitzpatrick et al. [60] presented a FEM knee based on a KKS interface with a proportional–integral–derivative (PID) controller for reproducing repeated in vitro knee joint loading conditions as KKS across different implant designs. Recently, the Abaque knee simulator (AKS) [37] (Dassault Systemes Simulia Corp., Johnston, USA) faithfully mimicked the biological construction of the normal TF and PF joints, including all ligaments and soft tissues around the knee joint. In [79], the authors developed a PID-controlled MSM-integrated AKS model for predicting the patient-specific kinematics and contact mechanics of the TF and PF joints.

3.1.2 MSM-based computational modelling in TKR: With the application of FEM in the TKR field since 1977, Van Eijden et al. [80] first developed an MSM-based mathematical model to analyse the PF joint on the sagittal plane. This model considers the dynamic movements and forces on the PF joint. Essinger et al. [43] presented an intact TKR model with PF and TF joints on the basis of a total energy minimisation principle. They also reported a spring-based contact algorithm, which was verified by in vitro experiments. It was the first time the contact mechanics and kinematics on an intact TKR joint were predicted, thereby providing a theoretical basis for the current MSM. Garg and Walker [38] developed a computer-graphic model for kinematics and ligament analysis of TKR. However, the absence of dynamic loading and muscle force during a specific daily activity remains a limitation of the model. In addition, the dynamic analysis remains focused mainly on the joint level with rigid contact. The body-level MSM has been gradually used in the orthopaedic field after Delp et al. [13] developed the interactive graphics-based lower limb model. However, it is applied mainly to the analysis of natural knee mechanics [39, 81]. It was not until 2001 that Piazza and Delp [46] developed a 3D dynamic simulation for TKR during a step-up task to predict the joint load, muscle force, and kinematics of TKR. The contact algorithm inside a TKR knee was mostly the basis of the study by Essinger et al. [43], which was for a rigid contact. However, this algorithm was not capable of predicting the contact pressure through the rigid contact theory. Ball and Fregly, therefore, developed a novel algorithm for combining MSM with a deformable contact knee model. This methodology increased the prediction accuracy and efficiency. It was followed by the rigid-body-based contact algorithm in the FEM presented by Halloran et al. [49], which is very similar to the contact algorithm. The mirror dynamic has been widely used in traditional MSM, providing good accuracy on muscle force prediction. However, a forward dynamic simulation potentially offers a powerful approach for investigating the connection between muscle and joint movement. Hence, many studies have focussed on this topic. One of the proposed methods called computed muscle control (CMC) has been highlighted and widely used in MSM [40, 82]. CMC employs feedforward and feedback control to closely track experimental kinematics using only a single integration of the model state equations. High accuracy was achieved in comparison with electromyography (EMG) data. Furthermore, the commercial MSM-based knee simulator LifeMOD/KneeSim (Biomechanics Research Group Inc., San Clemente, USA) [47], which can predict the kinematics, contact mechanics, and ligament tension on both TF and PF joints has been widely used in the TKR field. In this period, MSM-based models were focused mainly on knee-level models at the beginning. Dynamic full-body models were gradually developed for concurrent prediction of the muscle force and knee joint level kinematics. However, an explicit evaluation to verify the model is still lacking; verification is performed only by using EMG or fluoroscopy measurements. In addition, the application of design and parametric analysis of implants on body-level kinematics and muscle force has not yet been conducted.

With the development of commercial and open-access MS models (such as Anybody, OpenSim, and LifeMOD) and algorithms for contact and dynamic modelling, the application of MSM has significantly increased in the TKR field since 2010. Previous research studies have reported different approaches for improving those commercial or open-access models. Furthermore, this has provided an efficient way to verify MS models with the development of instrumented knee implants through the grand challenge dataset (https://simtk.org/projects/kneeloads) [24, 83, 84]. Lin et al. [15] developed a method for simultaneously predicting the contact and muscle forces on a patient by forward dynamics with an instrumented knee replacement during a gait cycle. However, a large error occurred in the prediction of the lateral contact force, mainly because of the omission of neighbouring joints in the model. Trepczynski et al. [55] used a traditional inverse dynamic method to predict the contact force on TF and PF joints. High accuracy was achieved when compared with in vivo experiments performed during various activities. Thelen et al. [63] developed a concurrent MS model for predicting the knee and neuromuscular mechanics during a gait cycle by using the CMC algorithm in OpenSim and by applying the quasi-static knee mechanics model by Garg and Walker [38]. They also investigated the influence of valgus alignment on the contact pressure of TF. This was the first time, to the best of our knowledge, that the full-body MS model was applied for the in vivo clinical analysis of knee prosthetics. Marra et al. [64] developed a patient-specific MSM framework to investigate the kinematics, contact mechanics, and muscle force during a gait cycle by using inverse dynamics under the platform Anybody (AnyBody Technology A/S, Aalborg, Denmark). A much higher predicted accuracy was achieved in this work compared with previous studies.

3.1.3 FEM + MSM computational modelling in TKR: MSM is known to efficiently predict the patient-specific body-level biomechanics by including muscle force and joint loading, but cannot accurately predict tissue-level mechanics such as contact pressure and motions. The capability of FEM is just the opposite. Therefore, concurrent FEM with MSM has become a new topic in computational simulation of TKR. Zhang et al. [28] presented a novel approach to predict the patient-specific wear of the TKR by concurrent MSM and FEM. In [26], the authors developed an FE–MS model that could simultaneously predict tissue-level kinematics, contact mechanics, and body-level mechanics based on inverse dynamics. The authors also reported a higher predicted accuracy for this FE–MS model than for models of previous publications when compared with in vivo experimental data. In addition, Navacchia et al. [41] and Hume et al. [69] presented a forward-dynamic-based FE–MS model, although the contact analysis between TF and PF joint was not included. However, this model remains attractive for future applications of preclinical evaluation and design of TKR.
3.2 Design of computational-model-based experiments

The outcome of computational simulation is highly dependent on the approach followed concerning boundary and loading conditions set in the model. In addition, the large variety of patient-specific and surgical parameters will significantly affect the outcome of the simulation. Clinical results show that the property of ligaments, implant alignment, walking habits, and anatomical structure of bones affect the service life of TKR. The design of computer-simulation-based experiments, therefore, provides an efficient way to perform sensitivity analyses or even optimisation to address specific clinical problems.

The most widely used mathematical approaches for the design of computational model-based experiments are the design of experiments (DOE), sensitivity analysis, probabilistic analysis, and optimisation. DOE allows a reduced number of experiments while still extracting the required information [85, 86]. Sensitivity analysis is used to determine the significance of each design variable on each selected response. Such an analysis is normally applied after DOE in computational simulation to distinguish the significant parameters [87, 88]. However, both DOE and sensitivity analysis assume that the possibilities for each experimental condition are the same, which is not realistic. Probabilistic analysis is a powerful method to simultaneously explore the distribution of each parameter and the probability of a particular outcome. Monte–Carlo analysis is the most widely used method in this field, featuring randomly distributed parameters [52, 61].

For example, Pal et al. [52] developed a probabilistic wear prediction model to consider the uncertainties in component alignment, constraints, and loading conditions. Fitzpatrick et al. [61] explored the effect of the patient, surgical, and implant design variations in TKR performance through probabilistic analysis. However, most of these experimental design methods were used with FEM, with the exception of Pal et al. [51], who developed an MS model concurrent with a probabilistic model to understand the sensitivity of kinematics and alignment of TKR to the muscle activation. To the best of our knowledge, it was also the first time a probabilistic method was used in body-level mechanics analysis.

The optimisation of implant design and preclinical planning is the ultimate goal of this research, but very limited research focuses on it. Only Willing and Kim [29, 56, 57] developed several serial parametric studies on geometrical optimisation with consideration of wear, stress, pressure, and laxity of TKR based on MSM. A large design space (14 design parameters) was explored in this optimisation. Nonetheless, the knee simulator was significantly simplified to reduce the cost of the calculation. The performance of the TKR is known to be influenced significantly by the design of the simulator. It would be questionable whether the optimised version would also be applicable under actual in vivo conditions. The challenge of applying optimisation technology to computational modelling of TKR consists in reducing the computational cost, increasing the validity of the knee model, automating modelling of the TKR design, and validating the optimised result. The sensitivity of the computational model is also an important factor that affects the predicted accuracy and reliability. A comprehensive optimisation model is expected to be associated with the development of fully automated modelling [91], high-fidelity FE-MS modelling [26], supercomputers, and artificial intelligence in the next decade. In addition, a supplementary method to consider patient variability or optimised design consists of developing a statistical model that represents the relationship between design and performance of TKR. An advantage of active appearance models is that they can be used to generate hundreds to thousands of synthetic instances based on a smaller representative TKR design or joint training set.

3.3 Applications and findings through computational simulation of TKR

Long-term satisfaction of the patients and survival of the TKR are highly dependent on the restoration of sensation, stability, range of motion, and functions by the knee prosthesis. Computational modelling has provided an efficient tool to obtain insight into the detailed performance of TKR in those aspects. The top-down research approach of computational modelling in TKR is shown in Fig. 3.

The main applications of computational simulation in TKR can be approximately divided into implant design, implant alignment, ligament strategy, and rehabilitation strategy on both TF and PF joints. For example, Morra [92] used KneeSIM to predict the contact stress on patellar components during daily activities. They also compared the kinematic pattern of various TKR designs with

Fig. 3 Research lines by using computational modelling for improving the performance.
that of a healthy knee and found that none of the designs reproduced the normal kinematics [93]. Fitzpatrick et al. [61] analysed the effect of the patient, alignment, and design on the outcome of TKR. They concluded that both the design and the patient presented the largest effects on the contact area of the TF and PF joint, respectively. In addition, they investigated the influence of the sagittal radius on translation and found that a gradually reducing sagittal radius followed by an increasing brake radius would improve the kinematic performance of TKR [10]. For the implant alignment application, Okamoto et al. [65] studied the effect of tibial posterior slope on TF joint kinematics, quadriceps force, and PF contact force by using KneeSIM. Increased tibial slope was found to improve the exercise efficiency of the quadriceps muscle and could reduce the contact force of the PF joint. The anterior translation was found when the slope angle was greater than 5°. Fitzpatrick et al. [94] analysed the patellar bone strain in natural and implanted knees during simulated deep flexion through FEM. They found that the volume of bone experiencing strains was >0.5% with the implanted condition, whereas it was ~200% larger than with the natural condition. A related comprehensive analysis of TKR alignment was conducted by Chen et al. [66] with full-body MSM. These authors found that an internal rotation angle >3° could lead to unsatisfactory pain levels. Likewise, varus rotation >3° could lead to medial bone collapse. The activation of ligaments and muscle force will influence sensation and pain after TKR; Lee et al. [95] reported that the stress on the medial collateral ligament (MCL) and lateral collateral ligament (LCL) decreased with increased tibial posterior slope. Kang et al. [96] found that a patient-specific TKR design could significantly reduce ligament tension during deep squatting and the gait cycle compared with the standard TKR design. Rehabilitation planning after TKR is another application of computational modelling, although only a reduced number of research studies can be found in the literature. Li et al. [62] used MSM to understand the body-level difference between healthy and TKR knee joints. Their results show that TKR patients presented a ‘quadriceps avoidance’ gait pattern, with little contribution by the vasti muscle to the extension moment during the stance phase of the gait cycle.

Ardestani et al. [97] created a wavelet time-delay neural network with a database from MSM to reduce the medial maximum contact pressure on TI, which could keep the balance of wear on the medial and lateral sides of the TI. With the surge of machine learning technology in human movement biomechanics [98], we strongly believe that computational-modelling-based rehabilitation is essential for minimising the postoperative recovery time of TKR.

4 Discussion and conclusion

Computational modelling for preclinical testing and implant design of TKR has attracted great interest (Fig. 1) owing to characteristics such as high precision, high efficiency, low cost, and configurability. It promises advances in biomechanics analysis, better decision-making in surgery, and acceleration of the design updating of the implant. As a conclusion of the present review, the development of FEM-based computational modelling technology in TKR appears to have reached a plateau, whereas body-level MSM-based methods remain at an early stage of development. Moreover, the concurrent study of FEM and MSM is a growing trend in computational modelling of TKR, as indicated in Fig. 2. The development of forward dynamic algorithms and explicit FEM in computational modelling would also broaden the application of the model. However, the practical and widespread use of those methods will remain limited until at least the following challenges are overcome:

(a) **Material property**: It is still difficult to specify the detailed material properties of bone, implant, soft tissue, and muscle, which significantly affect the accuracy of prediction. For example, in both MSM and FEM, kinematics and ligament forces depend on the force–strain property of ligaments, but little research has been performed on precisely the material property of ligaments.

(b) **Subject-specific**: Joint performance significantly influences the performance of TKR and varies among subjects. The patient-specific differences in TKR must be considered, e.g.

![Workflow and processing of patient-specific knee prosthesis design with computational preclinical testing](image-url)
muscle insertion position, body mass index, personal habits, and ligament properties. The inputs of the computational model are also critical aspects that affect prediction accuracy. Most studies of FEM and MSM focused on static or quasi-static conditions, not the standardised loading conditions. However, analysing performance at a subjective level with consideration of the patient-specific loading condition remains challenging, although the FEM + MSM combination is a potential approach.

(c) Evaluation: Only the validated models can be safely applied to actual clinical use or design of TKR. However, in some cases, the computational model is not sufficiently verified, for instance in terms of ligament force, muscle activation, in vivo wear of TI and stress distribution in the bone. A comprehensive evaluation for a large number of subjects would be valuable for biomechanics, but such an evaluation is expensive and time-consuming. In addition, in most studies, the computational models are validated by adjusting parameters in the model to fit the experiment. Whether these validated models can be applied under other conditions remains questionable.

(d) Experimental design: Population-based modelling is preferred in the design and evaluation of standard implants. However, most of the studies are focused on very limited conditions and activities. Moreover, an effective computational approach to accelerate the implant design is urgently needed.

(e) Sensitivity: This is a decisive factor for the capability of the computational model. For example, should the model catch only first-order effects, or should it be capable of distinguishing second-order effects? It is difficult to answer every question with a single model, and nonetheless, the model must be capable of discriminating against the effects of the two orders.

(f) Comprehensive: The performance of TKR must be considered from a wide perspective in computational simulations. Current studies focus only on limited aspects. For instance, wear and laxity are important for the service life and sensation of TKR, respectively. Most wear simulations have highlighted that an enhancement in the conformity of TI could decrease the wear and therefore increase the service life of the implant. However, greater conformity would decrease the joint laxity, which would worsen the user experience. A comprehensive model that enables investigation of the related performance changes is therefore preferred in practical clinical applications.

The significant interest in patient-specific knee prosthesis design and preclinical planning makes computational modelling an effective platform to accelerate the design and optimisation. Challenges remain in applying those models to commercial uses, but technological advances in patient-specific property identification, model generation, and implant manufacturing may lead to the emergence of this technology in the coming decade. Fig. 4 provides a workflow of a patient-specific knee prosthesis design with preclinical testing on the basis of patient-specific FE-MS models from the authors’ group. Overall, computational modelling has already provided great insight into the mechanics of TKR. The patient-specific knee prosthesis is designed on the basis of computed tomography (CT) data from an osteoarthritis patient. First, the 3D model of the knee joint and bone resection contour was obtained based on the CT data. An anatomical analysis of the healthy knee based on CT data was conducted to understand the anatomical characteristics. Next, the sagittal and coronal functional curves of the femoral component were calculated based on the in vivo femoral condylar with the mapping of patient data. Finally, patient-specific knee prosthesis was designed based on the patient data and the concept of surface-guided TKR for deep flexion and natural kinematics. To be specific, the medial pivot motion was generated by the ball-in-socket on the medial side of the TF joint. The enhanced rollback was generated by special shapes of the articulating surfaces on the TF joint. The in-vivo experimentally evaluated and the patient-specific knee model was implemented in the preclinical testing, which included the kinematics, contact mechanics, wear, and soft tissue balance.

With the development of modelling techniques, computational modelling would further assist biomechanics surgeons, implant manufacturers, and surgeons that implement in more practical and comprehensive issues.

5 References

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