MULTIBODY DYNAMIC MODELS IN BIOMECHANICS:

MODELLING ISSUES AND A NEW MODEL

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Abstract

In the surgical process of total knee replacement (TKR), it is well known that the three types of failures which are; a) unable to reproduce normal knee function, b) bone-implant interface failure c) wear during use. These failures are certainly due to the motion and the load that influence the prosthesis components. In this study, the modelling questions of the human knee joint will be discussed in relation only to the multibody dynamics models. Firstly, a summary is presented about the relevant literature, where the models with their different features are presented and evaluated. The existing models are mainly focused on the investigation of the ligaments (linear of non-linear properties), the description of the contact path, and contact forces during the motion, kinematics (rotation, abduction and adduction) and even the wear mechanism of the knee joint. The primal advantages of the multibody dynamics models are the easy adaptability in the mechanical parameters to carry out simulations and the connection with CAE programs that helps the design of new prostheses. A new multibody model is also presented by the authors.

Keywords: multibody dynamics, knee, forces, ligaments

1 INTRODUCTION

Computer models proved to be useful tools for predicting human movement in case of 3D scanned bones models or especially TKR models and evaluating the mechanical behaviour of the prosthesis components as a function of their geometry. Besides the investigation, these computer models were developed to offer indications on how prosthesis designs, ligament restraints or surface friction influence the joint motions [1-3].

Even though that more and more new prostheses appear in the medical field some questions related to the design of these implants are still not fully answered; therefore, old problems rise again and again. By narrowing these problems, three kinds of knee prosthesis failure have been so far reported in the literature, namely:

1. Failure to reproduce normal (optimal) knee functionality [4],
2. Failure in bone-implant connection which leads to loosening [5],
3. Wear of the prosthesis itself [5].

To solve these rather fundamental problems, an accurate knowledge of the in vivo kinematics of the human knee is important in order to improve the treatment of knee pathologies. Knee kinematics has been measured extensively with cadavers [6, 7], video techniques [8], radiographic [9] and magnetic resonance imaging techniques [10].

Recent numerical studies have implemented 3D models generated from CT scans to estimate the motion of the contact points of the femur on the tibial plateau using the bony geometry of the femur and tibia [9-10].

However, tibiofemoral contact occurs between the two cartilage layers, which very likely influence the kinematics. Furthermore, it will be shown through other previous studies in the literature that the geometry of the articular cartilage can be taken into account and the difference is quantified by in vivo activities.

Besides the review of these models, the second part of the study describes a new multibody (MB) patellofemoral model, which can estimate – regarding the kinetics – the occurring patellofemoral forces such as the quadriceps forces, patellofemoral compression force, patellar-ligament force and the tibiofemoral force. Regarding the kinematics, the model enables the calculation of the tibial rotation, abduction, adduction, patellar-tilt as a function of flexion angle.

2 MODELLING ISSUES

The knee is statically indeterminate thus the adequate computational system would apply a multibody dynamic model to predict muscle forces with a deformable contact model of the articular surface geometry to predict contact pressures. A review of some published three-dimensional MB knee models are summarized in Table 1 and reveals that only one was dynamic and utilized a full-body rather than knee-only model [18].

Table 1

Authors	**Dynamic or Quasi-Static**	**Model type**	**Contact type**
Wismans et al. [12]	Quasi-static	Knee	Rigid
Blankevoort et al. [13]	Quasi-static	Knee	Deformable
Pandy et al. [14, 15]	Quasi-static	Knee	Deformable
Abdel-Rahmann and Hefzy [16]	Quasi-static	Knee	Rigid
Kwak et al. [17]	Quasi-static	Knee	Deformable
Piazza and Delp [18]	Dynamic	Full-body	Rigid
Cohen et al. [19, 20]	Quasi-static	Knee	Deformable
Dhaher and Kahn [21]	Quasi-static	Knee	Rigid
Chao [22]	Quasi-static	Knee	Deformable

The rigid body models generally include muscle force predictions, but they utilize rigid contact theory, which cannot calculate contact pressures. While most models used deformable contact theory, they were quasi-static and so incapable of predicting muscle forces and joint contact pressures simultaneously during dynamic simulations.

The use of rigid body or deformable body, multibody- or finite element models always raises the issue of how to model the knee from the dynamic point of view. Though, it really depends on the actual aim of the research, some thumb-rules for these emerging issues have been gathered and narrowed them into three general questions:

1. Rigid or deformable body should be used in the investigations?
2. Linear or non-linear material laws should be used in the investigations?
3. What should be the contact condition?

2.1 Rigid or deformable body?

Let us start with the first question. It is worth to mention that according to Baldwin et al. [23] explicit Finite Element (FE) solution method is well suited for maintaining computational stability during large displacement and dynamic analyses such as the movement of the patella on the femur under deep squat (see in Figure 1).

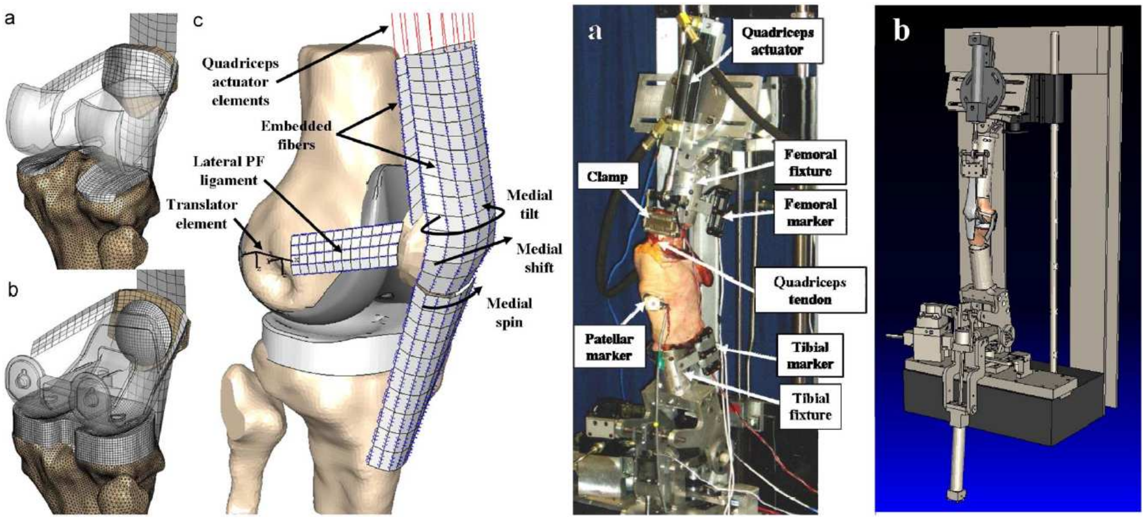

Figure 1. Model by Baldwin et al. [23]

The authors carried out an evaluation between model accuracy and computational time using both deformable and rigid contact formulations for the natural and implanted articular surfaces. The rigid contact analyses showed negligible differences in model-predicted PF kinematics with reductions in run times of two to four times that of the deformable analyses. Halloran et al. [24, 25] predicted kinematics of two types of implant designs under gait loading conditions with both deformable and rigid contact formulations in a similar study and found kinematic predictions to be nearly identical between the two contact formulations for the implants.

Other authors, Trent et al. [26], created both an FE and MB subject specific model to model the role of the menisci under gait (see in Figure 2).

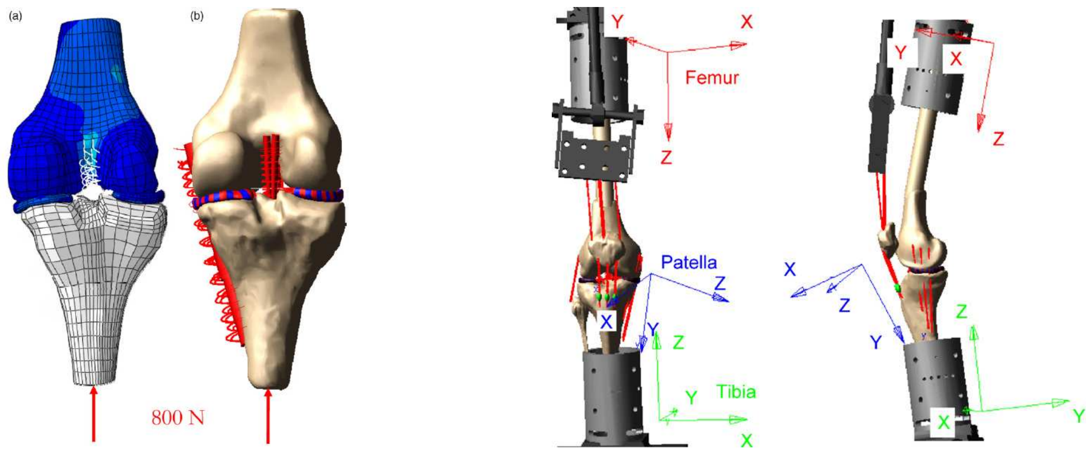

Figure 2. Multi body and Finite Element model by Trent et al. [26]

The results of the models showed that the both of the MB and FE calculation of the kinematic part corresponded very well with one another, although in some parts of the contact forces there was significant difference.

Thus, the rigid body approach combined with the absence of menisci has lesser effect on the kinematics of the knee joint, while the kinetics is more affected. It has to be mentioned that the effect is due to the absence of menisci.

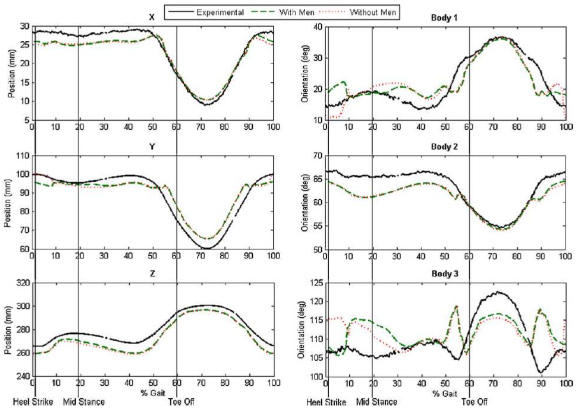

Figure 3. Patellar kinematics by Trent et al. [26]

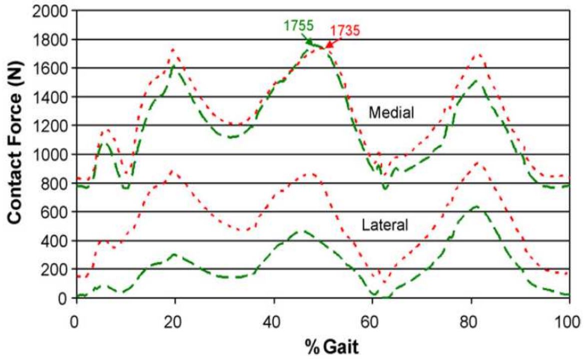

Figure 4. Contact forces by Trent et al. [26]

This comprehensive study showed as well, that the development time of the MB menisci models was much faster than development and meshing time of the FE menisci models, which makes the MB models more useful in this research area.

Consequently, the first question could be answered by suggesting rigid body models in case of kinematic investigation since no significant difference has been reported yet, but appropriate results. Related to the kinetics, it is not proven directly with these studies whether the forces are significantly influenced by MB or FE approaches.

2.2 Linear or non-linear material law?

Obviously the behavior of ligaments, muscles and the condyles alongside with the menisci are non-linear due to their biological nature, though the linear approximation can be proper fitting and convenient to use.

Bei and Fregly [27] formulated a modular modelling approach that permits incorporation of a deformable knee model into any multibody dynamic simulation environment on the valuable foundation provided by earlier studies (see in Figure 4 and Figure 5).

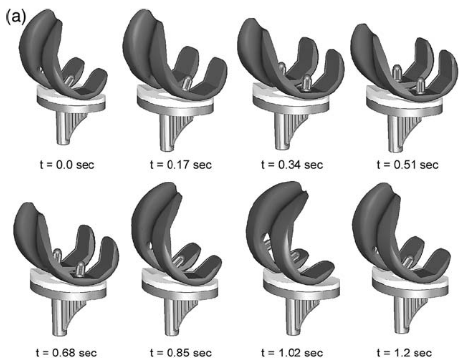

Figure 5. Motion simulation by Bei and Fregly [27]

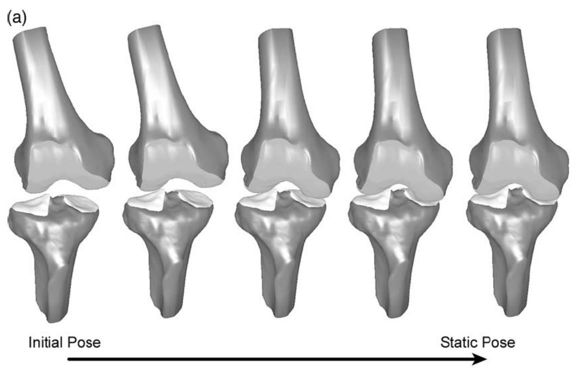

Figure 6. Contact simulation by Bei and Fregly [27]

The natural knee contact model was created from MRI and CT data and the artificial knee contact model produced from manufacturer's CAD data. This approach is an important step toward the realization of dynamic patellofemoral and musculoskeletal modelling, which can provide in vivo kinematic prediction under realistic loading simulation. By taking a look at Table 2 from Bei and Fregly [27] we can see the difference in numbers between the linear and non-linear approximation.

Table 2.

Quantity	**Linear material**		**Non-linear material**	
	Medial	**Lateral**	**Medial**	**Lateral**
Contact force (N)	1467	623.30	1417	613.1
Contact torque (Nm)	35.82	16.67	34.23	15.95
Peak pressure (MPa)	28.09	17.07	15.58	11.34
Average pressure (MPa)	16.60	9.03	12.50	8.01
Contact area (mm^2)	88.37	69.02	113.4	76.47

2.3 Contact conditions

Many research use computational support where the mechanical contact boundary conditions have to be set. Depending on the contacting bodies, the contact process can be divided into two categories: **impact** and **continuous** or **non-colliding** contact. The non-colliding term is preferred here because it encompasses the continuous contact, and also allows for loss of contact.

Impact is characterized by the impulsive change in the momentum of each of the contacting bodies, while non-colliding contact features a continuous evolution of the momentum.

The development of non-colliding contact calculation drew less attention compared to the large amount of other, mostly impact contact routines, though its relevance is unquestionable. The existing approaches to model non-colliding contact dynamics include the so-called **compliant** approaches (also called **penalty** methods) and complementarily formulations. Both of these have been employed in the context of multibody systems and can model contact occurring at multiple points.

Sharf and Zhang [28] introduced a new contact model related to the non-colliding contact and to earn credit to their method, they compared the obtained results with commercial software MSC.ADAMS, which employs a compliant contact force model. After they determined the contact modelling setting parameters, the results have been compared with the MSC.ADAMS result in case of a simple model (see in Figure 7).

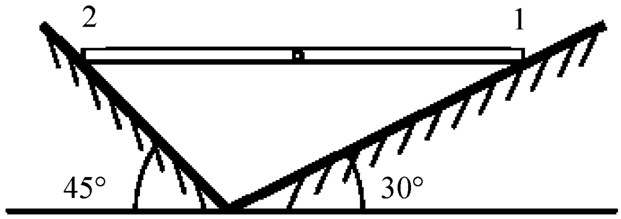

Figure 7. Test example by Sharf and Zhang [28]

The computation was carried out in frictionless case and with friction as well. In case of the frictionless case difference between the two predictions could not be distinguished, while in case of friction the results were nearly identical and the small deviation was originated to the fundamental difference between the two models (see in Figure 8 and Figure 9).

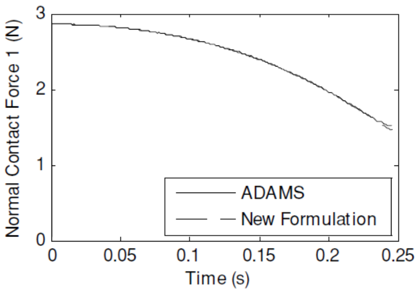

Figure 8. Without friction by Sharf and Zhang [28]

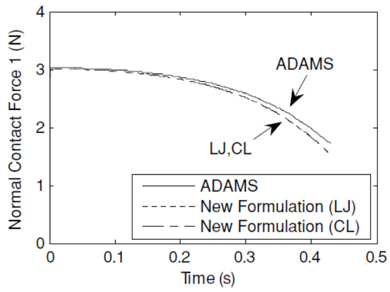

Figure 9. With friction by Sharf and Zhang [28]

As a consequence, beside the impact contact, the non-colliding contact or in the MSC.ADAMS the compliant approach are very advisable to use due to the fact that the contact might become loss in certain moment.

3 NEW MULTIBODY MODEL

By taking into account the earlier discussed studies and concerns, a model was built in the MSC.ADAMS program system (see in Figure 10). The bones, such as the tibia, patella and femur were assumed as rigid bodies, since the influence of deformation in this study is neglected due to the above mentioned reasons. The geometry of the femur and tibia is based on a prosthesis prototype, which is under tests and developments.

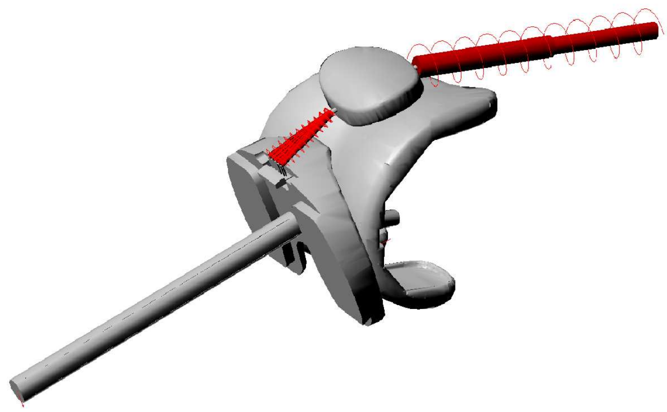

Figure 10. Multibody dynamic model

The patellar ligament and the quadriceps muscle were considered in the numerical model. The quadriceps muscle and the patellar ligament were modeled as simple linear springs. The stiffness coefficient was set to 130 N/mm and the damping coefficient to 0.15 Ns/mm in case of both springs which corresponds with the measured values in the literature [29, 30]

The 10 N as a magnitude of load was applied in case of the numerical model. The point of application was the end point of tibia. The femur was constrained by FIXED JOINT, where all the degrees of freedom were considered.

Between the femur, tibia and patella, CONTACT constraints were set according to Coulomb's law with respect to the very low friction (μ_s = 0.0003 μ_d = 0.0001) similarly to real joints [31, 32]. The relationship between the contact forces ($\boldsymbol{F_n}$, $\boldsymbol{F_s}$) and the patellofemoral forces was not analyzed in this study. No specific kinematic constrain was applied on the tibia, it could freely roll and glide on the condyle of the femur.

The simulation time was set to 0.045 sec in 200 time steps due to the fact that the motion was simulated in a short time interval and the applied number of time steps provided proper accuracy.

GSTIFF type integrator [33] was used for solving the ODE and DAE of the motion. The solver routine was set to work maximum 10^{-3} tolerance of error, while the maximum order of the polynomial was defined as 12. The solution converged very quickly with these parameters.

The post-procession was carried out in the MSC.ADAMS and partly in the Excel. The MSC.ADAMS can compute directly forces, velocities and accelerations, but not rotations.

The flexion angle was derived by integrating the angular velocities of the femur and tibia about the *x* axis, taking into account that the model was in an initial 15 degrees of squat in the beginning of the motion. In order to deal with the fact that the motion is three-dimensional, the angles are decomposed to three separate angles. These angles are the so-called Euler angles and the (313) convention was set to calculate them in a Cartesian coordinate system during the simulation. After integrating the appropriate angular velocities, the results were summarized in charts by using the Excel program.

As a validation, the model's results were compared to experimental result. Balassa et al. [34] designed and built an experimental set up, which is capable to measure continuously the quadriceps force, the rotation about the tibial axis and the flexion angle (see in Figure 11).

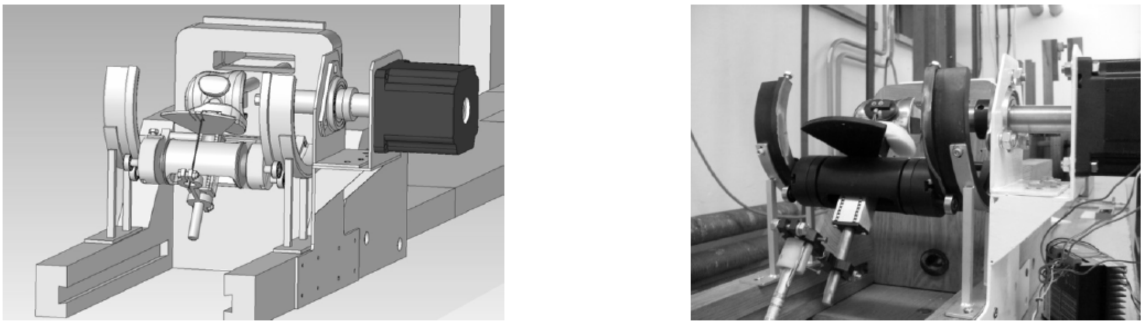

Figure 11. Test set up by Balassa et al. [34]

4 RESULTS

Since this multibody dynamic model is rather simple, only kinematic investigations were done, which was limited to the calculation of the rotation as a function of flexion angle. In Figure 12 the experimental and numerical results are shown. The obtained results are in good agreement, which gives ground for further investigations and development in the multibody model.

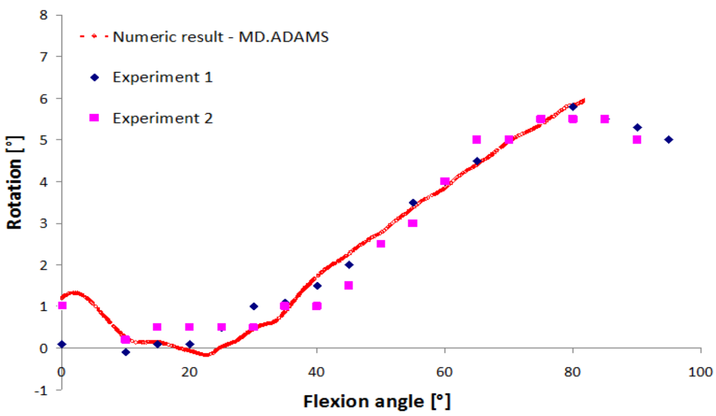

Figure 12. Comparison of the results

As for further plans; a new squat MB model is under development, which includes the movement of the torso as a new parameter. Since other authors could already implement the menisci into MB models, we have this aim as well. The kinetics (patellofemoral forces) will be thoroughly investigated in a future study.

5 CONCLUSIONS

A review has been given over a few important modelling questions and issues related to multibody dynamics models. General directions have been presented concerning the modelling in case of kinematic investigation and the differences were quantified through other published studies.

In the second part of the study a new multibody model has been introduced, which enables the calculation of the tibial rotation as a function of flexion angle. The multibody model's results have been validated by the results of an independent experimental study, and showed good agreement.

Further improvement of the model is planned, which involves the squat modelling with moving centre of gravity and menisci. These features are currently under development.

6 ACKNOWLEDGEMENTS

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