Ribeiro, A., Rasmussen, J., Flores, P., Silva, L.F., *Modeling of the condyle elements within a biomechanical knee model*. Multibody System Dynamics, Vol. 28, pp. 181-197, 2012

Abstract. *The development of a computational multibody knee model able to capture some of the fundamental properties of the human knee articulation is presented. This desideratum is reached by including the kinetics of the real knee articulation. The research question is whether an accurate modeling of the condyle contact in the knee will lead to reproduction of the complex combination of flexion/extension, abduction/adduction and tibial rotation observed in the real knee? The model is composed by two anatomic segments, the tibia and the femur, whose characteristics are functions of the geometric and anatomic properties of the real bones. The biomechanical model characterization is developed under the framework of multibody systems methodologies using Cartesian coordinates. The type of approach used in the proposed knee model is the joint surface contact conditions between ellipsoids, representing the two femoral condyles, and points, representing the tibial plateau and the menisci. These elements are closely fitted to the actual knee geometry. This task is undertaken by considering a parameter optimization process to replicate experimental data published in the literature, namely that by Lafortune and his co-workers in 1992. Then, kinematic data in the form of flexion/extension patterns are imposed on the model corresponding to the stance phase of the human gait. From the results obtained, by performing several computational simulations, it can be observed that the knee model approximates the average secondary motion patterns observed in the literature. Because the literature reports considerable inter-individual differences in the secondary motion patterns, the knee model presented here is also used to check whether it is possible to reproduce the observed differences with reasonable variations of bone shape parameters. This task is accomplished by a parameter study, in which the main variables that define the geometry of condyles are taken into account. It was observed that the data reveal a difference in secondary kinematics of the knee in flexion versus extension. The likely explanation for this fact is the elastic component of the secondary motions created by the combination of joint forces and soft tissue deformations. The proposed knee model is, therefore, used to investigate whether this observed behavior can be explained by reasonable elastic deformations of the points representing the menisci in the model.*

Keywords: Knee Modeling, Stance Phase, Condyles, Multibody Methodologies

1 INTRODUCTION

The human body has relatively rigid bones, connected by special joints capable of large anatomical articulations. From the mechanical point-of-view, this description of the human body is similar to that of a multibody system. However, the human body system is far more complex than the great majority of the multibody systems. Its components have a complex behavior due to deformations associated with the soft tissues such as the muscles, tendons and ligaments, and due to complexity of the anatomical articulations relative to the standard mechanical joints [1-4].

It is well known that computational simulation of human motion requires the development and implementation of suitable mathematical models that correctly describe the behavior of the human body and its interaction with the surrounding environment. There are two main approaches to modeling the human body as a biomechanical system, finite element analysis [5, 6], and multibody system methodologies [6-8]. The finite element models are applied in cases where localized structural deformations are considered in detail, while multibody models are usually applied in cases where gross-motions are involved and when complex interactions with the surrounding environment are expected. For instance, human gait as a gross-motion simulation is usually described using multibody system methodologies [10, 11].

The study of human body motion as a multibody system is a challenging research field that has undergone enormous developments over the last years. Computer simulation has been useful in many research and development activities, such as: (i) analysis of athletic actions, to improve different sports performances and optimization of the design of sports equipment, (ii) ergonomic studies, to assess operating conditions for comfort and efficiency in different aspects of human body interactions with the environment; (iii) orthopedics, to improve the design and analysis of prostheses; (iv) occupant dynamic analysis for crashworthiness and vehicle safety-related research and design; (v) and gait analysis and subsequent diagnosis of pathologies and disabilities.

Multibody-based methodologies have been developed in such a way that, besides the representation of mechanical systems made only of rigid bodies [12, 13], they also allow the description of deformable bodies [14, 15]. Along with the theoretical developments in recent years, several powerful and reliable multibody commercial codes have been put on the market [16, 17], and the study of human body motion as a multibody system has undergone significant development [18, 19], allowing the geometrical and physical properties of bones, muscles, tendons, etc. that constitute the biomechanical models to be taken into account.

Broadly, much of the research developed with the purpose to simulate daily human tasks is based on the assumption that the joints that constrain the system's components are considered as ideal or perfect joints, such as spherical, revolute and universal joints. Nevertheless, with this approach significant decrease of the kinematic precision compared with the living body can occur because the idealized models fail to capture more complex aspects of joint kinematics [20]. However, some advanced model have presented over the last decades, namely those developed by Stevens and co-workers, who designed several prosthesis knee devices, in particular the crossed four bar mechanism and Stevens six bar linkage, that produces both the gliding motion as well of the rotation of the tibia with respect to the femur [21, 22].

In the field of mechanical system dynamics, computational methods for representation of complex phenomena such as incongruent geometry, contact, friction and lubrication have been developed [23-26]. However, the application of these methods in the field of biomechanical system dynamics lacks somewhat behind. A possible reason is that much biomechanical simulation is based upon inverse dynamics, where movement of all degrees-of-freedom is input to the analysis leading to a presumption of simple joint kinematics. For most applications concerning simple joints, such as the hip joint [27], this is a reasonable assumption, but for detailed investigations of more complex and incongruent joints, such as the knee joint [28], is it not. The purpose of this paper is to present a biomechanics model capturing some of the complexity of the knee joint while retaining the computational efficiency of inverse dynamics analysis. A bi-product of the model development is a new understanding and classification of the mechanisms behind observed knee kinematics.

The knee is one of the most complex synovial joints in the human body. It has two main functions: (i) to permit the movement during locomotion, and (ii) to provide static stability. The mobility associated with the knee joint is indispensable to human locomotion and it helps the correct foot orientation and positioning in order to overcome the possible ground irregularities. The knee is one of the most studied human anatomical articulations due its importance in activities of daily living and due to the high incidence of joint degeneration and injury, which ultimately can lead to serious disability and affect the human locomotion [29]. The history of modeling and analysis of the human knee joint is long and rich [30]. Most of the available models are based on simplified assumptions considering the complexity of in-vivo knee articulations. In this sense, mathematical knee models have been proposed to obtain a better understanding of the complex mechanical behavior of the joint's substructures and its interactions with the human musculoskeletal system.

One of the first published works on this issue is due to Strasser [31], who in 1917 developed a simple knee model based on a two-dimensional four-bar mechanism, in which two bars represent the cruciate ligaments, and the remaining bars denote the femur and tibia bones. This planar model was subsequently improved by Menschik [32] by including two curves representing the femur and tibia articular surfaces. In this model, the location of the insertion areas of the collateral ligaments was studied. Crowninshield and his co-workers [33] presented an analytical model to study the biomechanics of the knee joint. This method is the so-called inverse method, in which the ligament forces caused by a set of translations and rotations in specific directions are determined by comparing the geometries of the initial and displaced configurations of the knee joint. Wismans et al. [34] developed one of the first three-dimensional analytical knee models. This model considers not only the knee geometric properties, but also the static equilibrium of the system. They included a three-dimensional curved geometry of the tibia and femur surfaces, as well as nonlinear elastic spring to model ligaments. Moeinzadeh et al. [35] developed a two-dimensional dynamic model of the knee including ligament resistance, and specified a force and moment on the femur. A similar model was developed by Abdel-Rahman and Hefzy [36], which was later extended to three-dimensions [37]. Piazza and Delp [38] presented a rigid body dynamic model of a total knee replacement performing a step-up task. Patterns of muscle activity and kinematics of the hip were measured experimentally and used as inputs to the simulation. The model included both tibiofemoral and patello-femoral interactions and predicted the flexion-extension pattern of the step-up activity. The contact problem between the knee joint elements was based on the linear complementarity problem [39]. More recently, Guess [40] presented a quite complete three-dimensional knee model developed within the framework of multibody systems. This study combines a cadaver-based validated natural knee model with a muscle driven forward dynamics simulation from a subject of similar height and weight for prediction of joint contact mechanics. This knee model also includes the menisci. The problem of modeling and analyzing the response of the patella-femoral joint was also performed by some authors [41-43]. In particular, the model developed by Essinger et al. [43] is characterized to be a quasi-static approach, which incorporates both the tibiofemoral and the patella-femoral joints. This model presents a limitation related to the geometry of the patella, which was simplified.

One of the notions that are frequently repeated in the scientific literature on knee models is that the knee is a six degree-of-freedom joint [44]. The meaning of this statement is likely that movement in all the translations and rotations defined in an anatomical coordinate system can be observed. However, from a multibody dynamics point-of-view, a six degree-of-freedom joint is not a useful classification because it constrains none of the mutual degree-of-freedom between the segments it connects. Since the human knee physically connects the femur and the tibia, it does obviously provide some amount of mechanical constraint between the two.

In this work we therefore present the development of a three-dimensional multibody knee model, which realizes movement with respect to all anatomical axes with physiologically motivated mechanical constraints. This biomechanical knee model is composed by two anatomic segments, namely the femur and the tibia, whose geometric and inertia properties are based on the real bones and segments. The surfaces of the condyles of the femur are represented by two ellipsoids, which impose kinematic conditions between the femur-tibia pair. The geometric elements of the condyles are closely fitted to the actual knee data found in the literature, in particular that published by Lafortune and his co-workers in 1992 [45]. This task is performed by means of a parameter optimization procedure to replicate experimental information. The resulting knee model captures the fundamental properties of the real knee articulation, thus providing a tool to better understanding the components of in-vivo knee kinematics. Then, kinematic data in the form of flexion/extension patterns are imposed on the model corresponding to the stance phase of the human gait, and the smaller articulations, more precisely abduction/adduction and tibial rotation follow from the joint constraints.

The model proposed here shows that the combination of flexion/extension, internal/external rotation and abduction/adduction can be explained by the shapes of condyles in combination with elastic deformations of tissues, leading to an understanding of the knee as a single degree-of-freedom elastic joint with a complex behavior.

2 MULTIBODY DYNAMICS

The dynamic analysis of multibody systems, made of interconnected bodies that undergo large displacements and rotations, is a research area with applications in a broad variety of engineering fields that has been deserved significant attention over the last few decades [1, 7, 8]. The various formulations available for modeling and analysis of multibody systems differ in the principle used, types of coordinates adopted and the method selected for handling constraints in systems characterized by closed loop topology [12-14]. The solution of constrained multibody systems can be obtained using the Lagrange's multipliers technique which leads to a set of differential and algebraic equations, in which the coordinates and the Lagrange multipliers are unknown quantities.

In summary, it can be said that a general multibody system embraces two main characteristics, namely: (i) mechanical components that describe large translational and rotational displacements and (ii) kinematic joints that impose some constraints or restrictions on the relative motion of the bodies. Thus, a multibody system encompasses a collection of rigid and/or flexible bodies interconnected by kinematic joints and possibly some force elements [12]. Driving elements and prescribed trajectories for given points of the system components, can also be represented under this general concept of multibody system. Figure 1 depicts an abstract representation of a multibody system [10]. In the present work, the generalized Cartesian coordinates are selected as the variables to describe the bodies' degrees-of-freedom. The formulation of the equations of motion uses the Newton-Euler approach, augmented with the constraint equations that lead to a set of differential algebraic equations.

When the configuration of the multibody system is described by *n* Cartesian coordinates $\mathbf{q}$, then a set of *m* algebraic kinematic independent holonomic constraints $\mathbf{\Phi}$ can be written in a compact form as

$$\begin{bmatrix} \mathbf{M} & \mathbf{\Phi}_{\mathbf{q}}^{T} \\ \mathbf{\Phi}_{\mathbf{q}} & \mathbf{0} \end{bmatrix} \begin{Bmatrix} \ddot{\mathbf{q}} \\ \boldsymbol{\lambda} \end{Bmatrix} = \begin{Bmatrix} \mathbf{g} \\ \boldsymbol{\gamma} \end{Bmatrix} \tag{1}$$

with the reference frame placed at the center of mass for each body, $\mathbf{M}$ is the system mass matrix, $\mathbf{\Phi}_{\mathbf{q}}$ is the Jacobian matrix of constraint equations, the vector $\ddot{\mathbf{q}}$ contains the generalized state accelerations, $\boldsymbol{\lambda}$ is the vector that contains the Lagrange multipliers, $\mathbf{g}$ is the vector of generalized forces and $\boldsymbol{\gamma}$ is the vector of quadratic velocity terms. For details, the interested reader is referred to references [12-14].

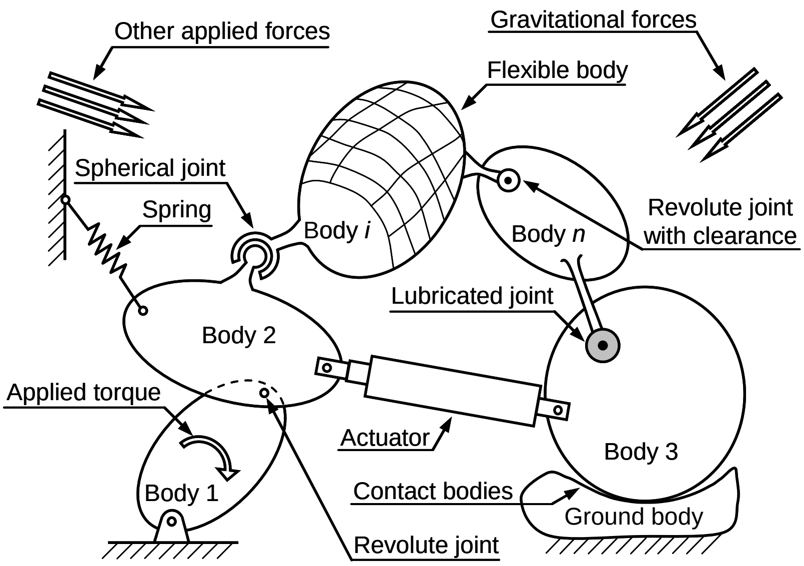


Figure 1: Generic representation of a multibody system with its most significant components: bodies, joints and forces elements

3 KINEMATICS OF THE KNEE DURING STANCE PHASE

The stance phase of gait comprises about 60% of the gait cycle for each leg and is responsible for most of the force generation in the knee. The stance phase is initiated with heel strike followed by a double support phase in which both feet are in contact with the ground. From toe-off of the contralateral leg follows the single support phase. The stance phase ends with toe-off from which the swing phase begins [10].

During stance phase, the ground reaction forces cause compression in the knee, which is carried by contact forces between the femoral condyles and the tibial plateau. These bony surfaces are incongruent, but the soft tissues forming the menisci and articular cartilage offer some amount of congruency against the two bony surfaces.

Detailed observations of knee kinematics reveal that the femur and tibia articulate with respect each other about all of the three anatomical axes depicted in Figure 2. This figure shows the main possible motions within the knee articulation, namely the primary flexion/extension and secondary motions of abduction/adduction and internal/external tibial rotation. While flexion and extension are obviously necessary for ambulation, the secondary motions are considered important for the knee's ability to accept the applied loads [46].

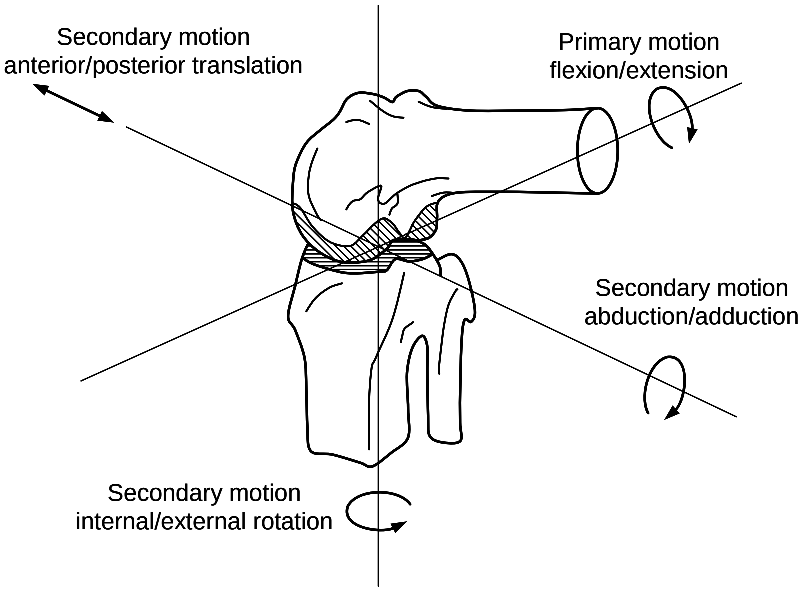


Figure 2: Primary and secondary motions present the knee articulation

Detailed knowledge of the *in vivo* knee movement is crucial for a better understanding of the normal knee function, as well as for addressing clinical injuries. The most frequently used method for measuring knee joint motion is to track the motion of optical markers attached to the skin, the so-called skin markers, of the shank and thigh. The motion of the markers is typically used to recover the underlying relative movement between adjacent segments, which in turn define the movement of the knee joint. However, the marker configurations and primarily the error due to skin movement artifacts limit the accuracy of the motion recovery [47, 48]. This limitation particularly has effect on the measurement of more subtle secondary movements.

Reliable observations of detailed knee kinematics are therefore difficult to obtain, making it necessary for the present work to build upon experimental data obtained from *in vivo* bone pin studies performed by Lafortune et al. [45]. Intra-cortical pins directly inserted into both bones and close-range photogrammetry were used to provide a 3-D reconstruction of the kinematics free from the influence of soft tissue artifacts. Five healthy male subjects (mean age, 27.2 years old; mean height, 180.6 cm and mean body mass, 75.2 kg) with no history of knee injury or previous surgical treatment of lower extremity participated in the study. Their knee joints were assessed clinically to be within normal limits. The reader is referred to references [30, 45] for a comprehensive description of the underlying experiment.

Figure 3 depicts the angular motions of the tibiofemoral articulation during the stance phase, i.e. the flexion/extension motion, the abduction/adduction motion and internal/external rotation. The plots represent the average of the five measurements obtained by Lafortune et al. [45] described above. The patterns begin at heel strike and end with toe-off [49]. The flexion/extension motion comprises two flexion events and one extension event during the stance phase. From the initial contact, when the knee is near fully extended, the knee flexes approximately 13.8 degrees. At opposite toe off, the first period of single support, the knee is continuing to flex to reach the peak stance phase flexion at about 19.5 degrees. After this period of knee flexion the knee starts to extend again, peaking at 0.6 degrees close to instant of heel rise. From heel rise until toe-off the knee flexes about 30.0 degrees. Contrary to the complex flexion/extension pattern, abduction/adduction of the tibiofemoral joint is practically uniphasic and is limited to about 1.8 degrees. From heel strike until shortly before toe-off (0.6 seconds, approximately) no abduction/adduction motion took place and the tibiofemoral joint remains abducted approximately 0.6 degrees [30]. Finally, the internal/external rotation follows a pattern somewhat similar to the flexion/extension, albeit with less than 10 degrees total angle variation over the stance phase.

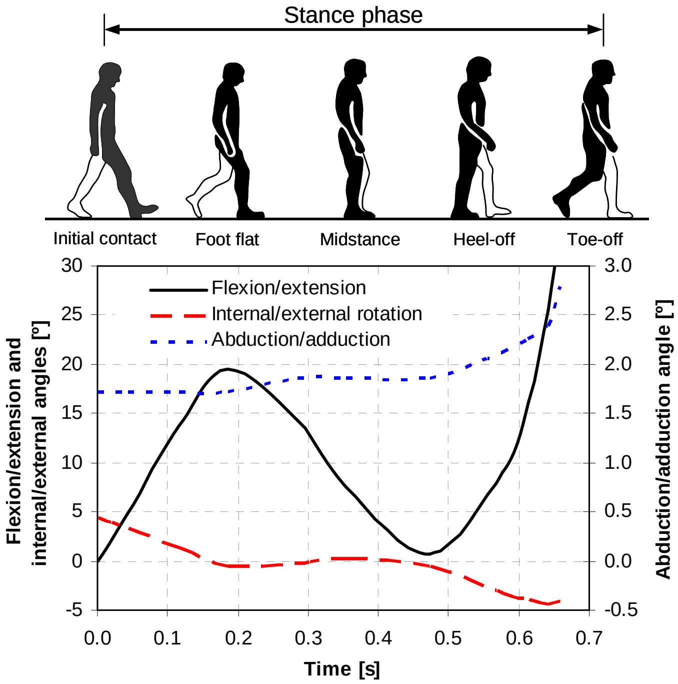


Figure 3: Angular motions of the tibiofemoral joint for the stance phase {adapted from [45]}

A better understanding of the correlation between flexion/extension and tibial internal/external rotation can be obtained by charting one as a function of the other. Figure 4 shows the tibial rotation as a function of the flexion angle for five subjects of the study by Lafortune et al. [45].

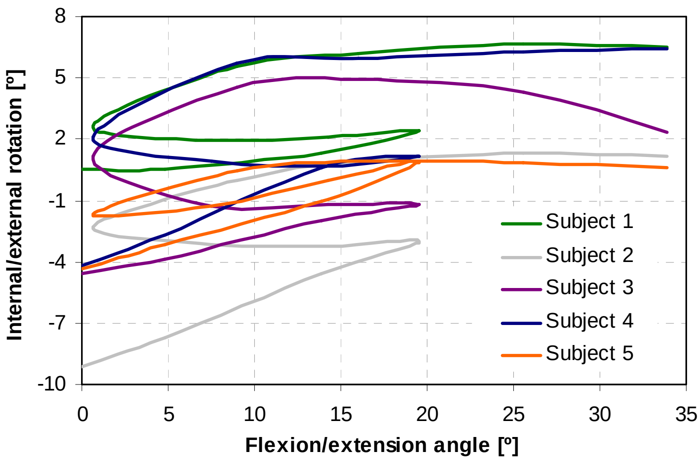


Figure 4: Angular pattern of the internal/external rotation versus knee flexion/extension angle during stance phase for the five subjects

The curves reveal significant similarities between the subjects, although the magnitudes of tibial rotation vary. The maximum ranges of the internal/external rotation for each subject extracted from Figure 4 are listed in Table 1.

	Maximum range of the internal/external rotation [º]
Subject 1	6.1
Subject 2	10.3
Subject 3	6.9
Subject 4	10.6
Subject 5	4.9

Table 1: Inter-individual differences in the internal/external rotation

Please observe from Figure 4 that each pattern of tibial rotation is a non-unique function of the flexion angle in the sense that, for the same knee flexion angle, there can be as much as three different angles of the internal/external rotation of the tibiofemoral joint. This behavior is consistent with the tibial rotation being the result of a kinematic constraint in the joint augmented by an elastic component associated with soft tissue deformations. This observation is the inspiration for the knee model presented in the next section.

4 MODELING OF THE KNEE JOINT

The biomechanical multibody knee model proposed in this work has been developed in the AnyBody Modeling System version 4.2 (AnyBody Technology, Aalborg, Denmark) [50]. The AnyBody Modeling System uses a multibody system formulation based on a Cartesian approach that enables the handling of closed and open kinematic chains. The knee model is composed of two rigid bodies, the tibia and the femur, whose characteristics are based on the geometric, anatomic and physiologic properties of the actual bones. The segment masses are estimated according to Winter [49] with the mass of the soft tissues surrounding each bone attributed to the bone. From the point-of-view of the multibody dynamics modeling, segments are effectively moving local reference frames equipped with mass properties, nodes and surfaces representing the actual bones and in particular the joint surfaces.

In standard multibody human models, the knee joint is typically modeled by a classical hinge joint, which only allows a single degree-of-freedom and does not reflect the complex nature of the knee motion found in the preceding section. Thus, in the present work, in order to get a realistic picture of the knee articulation, the tibiofemoral system is constrained by two ellipsoids that represent the femoral condyles and contact points that allow for the simulation of the menisci. Please notice that the contact of the two condyles with the tibial plateau effectively forms a closed kinematic chain, so the ability for the multibody dynamics system to handle closed chains is essential for knee modeling. The modeling of the ellipsoid joint is initiated with the identification of ellipses that best represent actual bone shapes. This idea has been corroborated by Li et al. [51] who used a similar approach for automatic analysis of the distal femur articular geometry. Thus, with the purpose of identifying the ellipses that represent actual condyle shapes, several points located on the two parasagittal planes through the condyles are considered, as Figure 5 illustrates. These two parasagittal planes are defined where the maximum deformation between the condyles and tibial plateau is taking place.

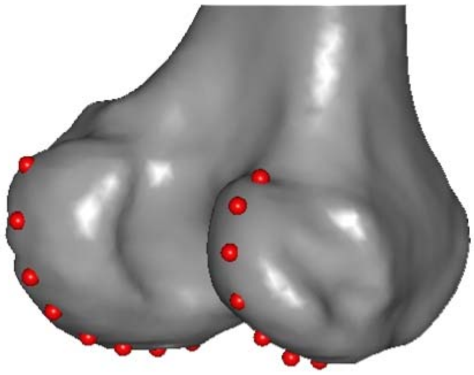


Figure 5: Points located on the articular condyle surfaces used to define the femur condyle ellipses.

The coordinates of the points located on the articular condyle surfaces were obtained from parasagittal planes intersecting the condyles of femur bone available in the AnyBody repository was considered. This bone was obtained from CT scans and represents an actual bone geometry. Figure 6 shows the local Cartesian coordinates of the potential contact points situated on the condyle surfaces. An optimization procedure was implemented using the classical equation of an ellipse to identify the semi-axes *a* and *b*, the local coordinates of the center, x_c and y_c, and the angular orientation, θ, of an optimally fitted ellipse. This equation can be written as

$$\left\{\frac{\left[(x-x_c)+(y-y_c)\right]^2}{a^2}+\frac{\left[(x-x_c)+(y-y_c)\right]^2}{b^2}-1\right\}^2=0 \qquad (2)$$

Equation (2) is squared to ensure that the left-hand side is always positive. Any point, (x, y), on the ellipse equation leads to a zero residual from Eq. (2), and any point outside the ellipse returns a non-zero residual. In the interest of simplicity of the equations, the ellipse equation (2) is aligned with the coordinate system and the sample points are rotated by $-\theta$ before insertion into the equation. By minimizing the sum of residuals of the sampled points of Figure 6, two sets of optimal parameters are obtained for the desired ellipses. Given the explicit nature of the problem, it can easily be solved using any readily available mathematical tool [30]. The two sets of five parameters that fit the medial and lateral condyles are listed in Table 2. In the next section, the influence of these five parameters on the behavior of the proposed knee model will be presented and analyzed.

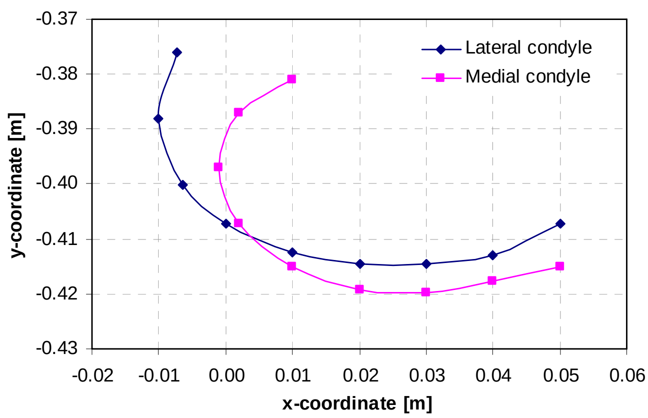


Figure 6: Cartesian coordinates of the potential contact points located on the condyle surfaces

Parameter	Medial condyle	Lateral condyle
a [m]	0.031279015	0.036536206
b [m]	-0.022139603	0.028515785
x_c [m]	0.029910996	0.026312422
y_c [m]	-0.397632647	-0.386432911
θ [º]	0.665029667	4.321339379

Table 2: Optimal parameters obtained for the medial and lateral condyle ellipses

Once the optimal ellipse parameters are obtained using the above procedure, they are used to define the ellipsoids that represent the femoral condyles. The AnyBody software and its modeling language, AnyScript, allow for definition of kinematic constraints between ellipsoid surfaces and points. Using this ability, four contact points representing the anterior and posterior horns of the lateral and medial menisci are defined on the tibia segment (Figure 7) [52]. A simple constraint preventing lateral displacement of the tibia with respect to the femur is also defined. The model is now constrained to a single degree-of-freedom, which can be perceived as flexion/extension. Please notice that these constraints are hard and do not take the elastic nature of menisci or other soft tissues into account. However, the constraints create a complex coupling between the anatomical articulations, i.e. flexion/extension, abduction/adduction and internal/external rotation within this single degree-of-freedom. In the following we shall first assess the resulting nominal motion and subsequently investigate whether deviations from it can be explained from elastic deformation of the soft tissues.

Having completed the new knee joint model, a gait pattern from the standard gait model of the AnyScript Managed Model Repository is imposed on the musculoskeletal model, and the results can be investigated and compared with the empirical data of Lafortune et al. [45].

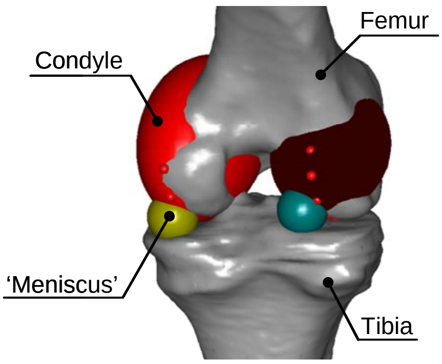


Figure 7: Representation of the proposed knee model with two contact points defined on the tibial plateau. The big spheres represent the posterior horns of the medial and lateral menisci.

5 RESULTS AND DISCUSSION

In this section, the main results obtained from computational simulations are used to discuss the main assumptions and procedures used throughout this work. For this purpose, the kinematic data in the form of flexion/extension patterns are imposed on the model described in the previous section that correspond to motion-captured data based on bone-mounted markers describing the stance phase of non-pathological human gait. The *in vivo* data of Figure 3 reveals that the abduction/adduction is quite small and we consequently focus on the internal/external rotation. Figure 8(a) shows that the imposed gait pattern in terms of knee flexion/extension is similar to the data of Lafortune et al. [45]. Figure 8(b) compares the realized tibial rotation with Lafortune's data over the stance phase. Finally, Figure 8(c) shows the tibial rotation as a function of the flexion angle with the empirical data.

In general, the results obtained with the proposed knee model have very similar trends as the measured data. Figure 8(c) shows, as expected, that the multibody model contrary to the empirical observations provides a unique relationship between flexion/extension and tibial rotation, but it closely reproduces the average trend. A possible explanation is that the observed non-unique behavior is the result of deformation of the soft tissues in the knee joint, thus deviating the rigid-body kinematics that depends on the forces acting over the knee. This observation allows for a new understanding of the knee as a single degree-of-freedom joint with a complex kinematic behavior given by the shapes of the contacting surfaces augmented by elastic deformations.

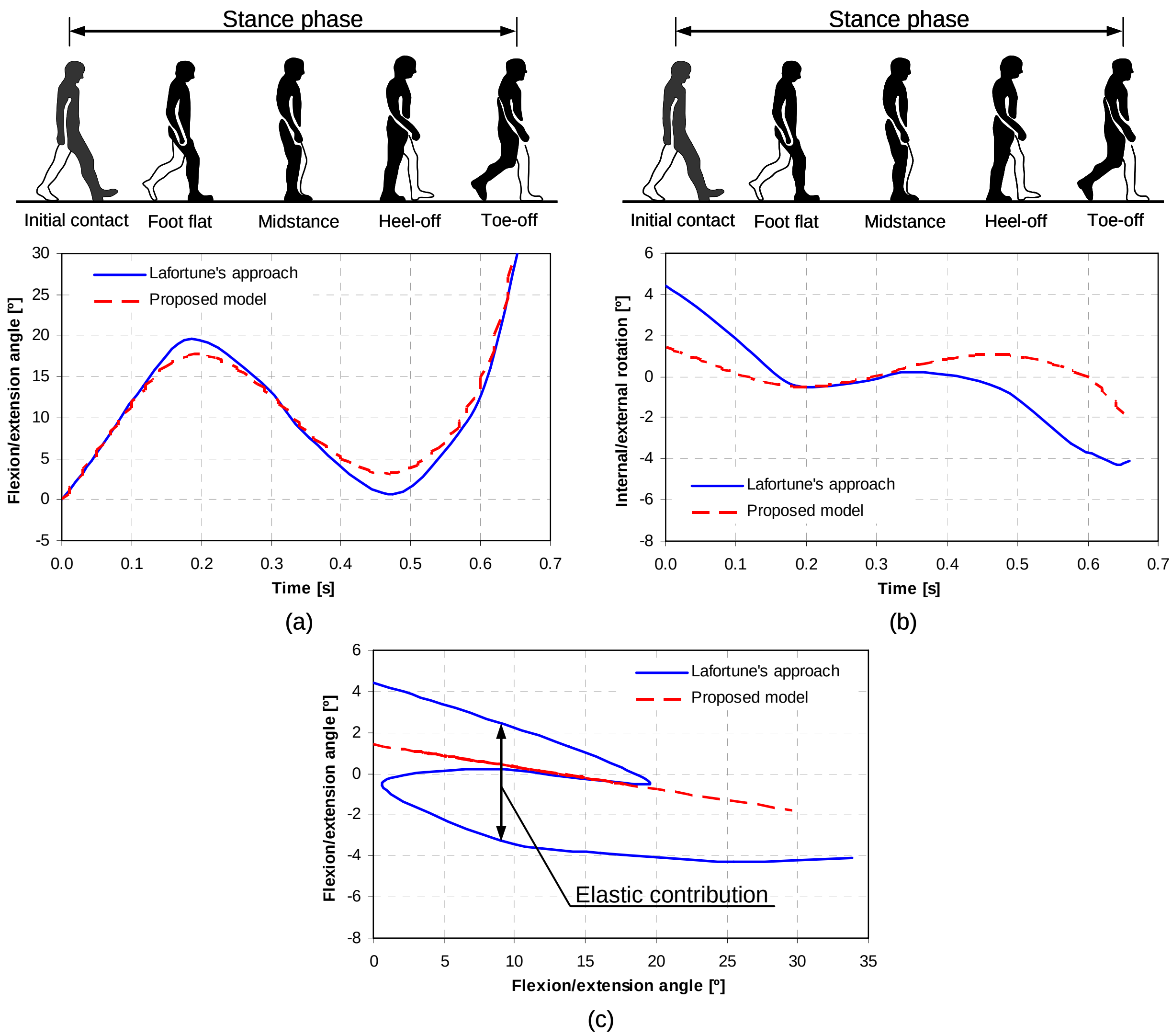
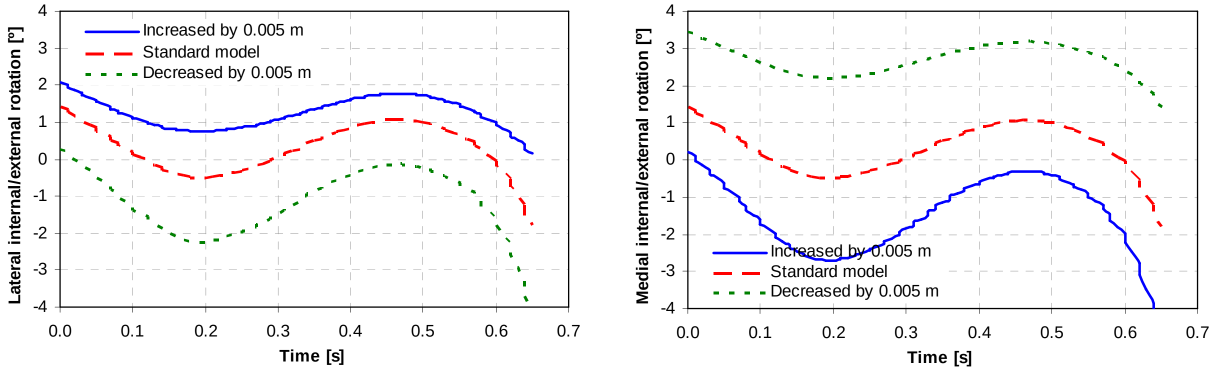


Figure 8: (a) Flexion/extension knee angle; (b) Internal/external knee rotation; (c) Angular pattern of the internal/external rotation versus knee flexion/extension angle during stance phase

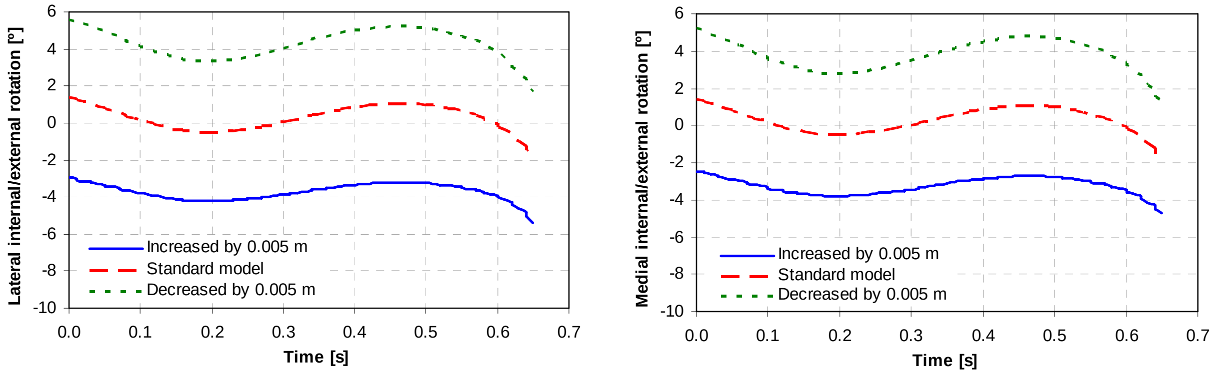
At this stage it is important to recover the fundamental research question of the present research work, which is whether an accurate modeling of the condyle contact in the knee will lead to reproduction of the complex combination of flexion/extension, abduction/adduction and tibial rotation observed in the real knee? The interesting finding is that the proposed knee with kinematics driven by realistic condyle shapes replicates many of the features of *in vivo* 3-

D complex knee motion. Over the last few years the femoral condyle shape has been investigated in order to understand whether and how the bone geometry influences the knee kinematics. This understanding is important because it quantifies to which extent each subject is unique and to which extent medical technology has to take the inter-individual variations into account. The present model allows for a study of the influence of reasonable parameter variations on knee kinematics.



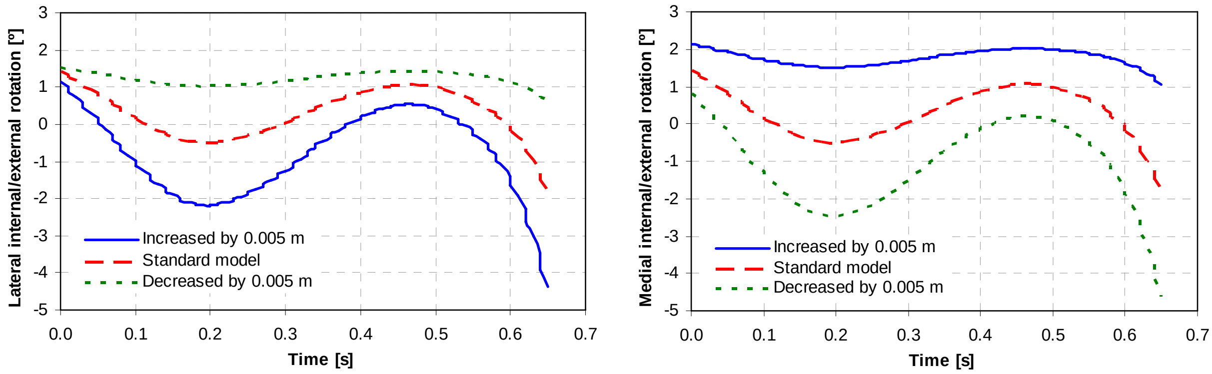
(a) (b)

Figure 9: Influence of the value of the semi-axis *a* on the internal/external rotation on (a) Lateral condyle; (b) Medial condyle



(a) (b)

Figure 10: Influence of the value of the semi-axis *b* on the internal/external rotation on (a) Lateral condyle; (b) Medial condyle



(a) (b)

Figure 11: Influence of the value of the x_c coordinate on the internal/external rotation on (a) Lateral condyle; (b) Medial condyle

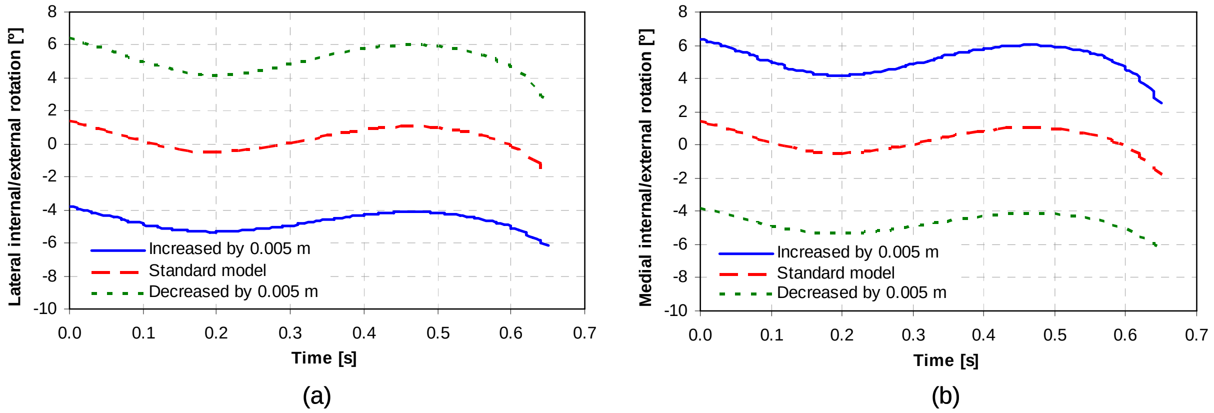


Figure 12: Influence of the value of the y_c coordinate on the internal/external rotation on (a) Lateral condyle; (b) Medial condyle

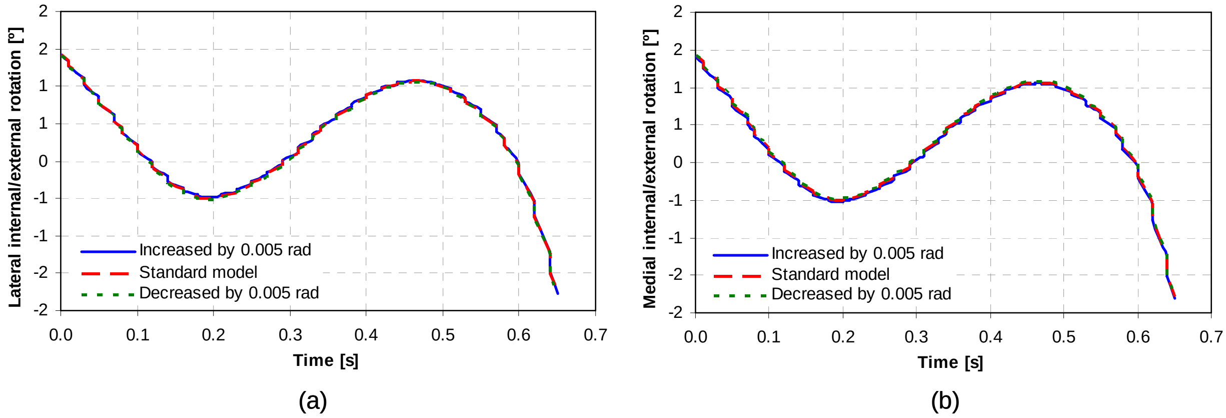


Figure 13: Influence of the value of the θ angle on the internal/external rotation on (a) Lateral condyle; (b) Medial condyle

Figures 9 through 13 present the influence of the variation of the five parameters that define the ellipses used to describe the condyles on the behavior of the proposed knee model. The standard values used in this study are those listed in Table 2. The response of the knee model is characterized by the plots of the internal/external rotation for both lateral and medial condyles. The variation of the parameters a, b, x_c, and y_c clearly affect the system response, as it can be observed in Figures 9-12. This fact, can explain the inter-individual differences of the internal/external rotation observed experimentally be Lafortune et al. [45]. The higher difference observed in the internal/external rotation is for the case in which the parameter is x_c. Finally, it appears that the parameter θ does not significantly affect the response of the knee model, as it can be observed in Figure 13.

The experimental data also reveal a difference in secondary kinematics of the knee in flexion versus extension. The likely explanation for this is an elastic component of the secondary motions created by the combination of joint forces and soft tissue deformation, as it can be observed in Figure 8(c). The menisci play important roles in friction reduction, load bearing and shock absorption within the knee. In this work they are furthermore presumed to play a role in centering each femoral condyle on its equivalent location on the tibial plateau, and that they are also secondary stabilizers of the joint. The menisci are dynamic structures, and to effectively keep an optimum load-bearing function over a moving, incongruent joint surface, they have the ability to deform, as the femur and tibia move, to keep maximum congruency. The model was therefore used to investigate whether the observed behavior could be explained by reasonable elastic deformations of the points representing the menisci in the model. The investigations revealed that this was indeed the case. The menisci in this model also showed a difference in segmental motion between the lateral and medial menisci. However, the obtained displacement for each meniscus is a little higher for this small knee flexion angle [30]. The biomechanical role of the meniscus is an expression of its gross and ultra structural anatomy and of its relationship to the surrounding intra-articular and extra-articular structures. Thus, this model considers that the anatomy of the menisci is based on simple geometric elements, which do not correspond to the real anatomy. This anatomic difference is able to explain these results. On the other hand, another reason for the obtained results is that in this model the muscles and the ligaments were not considered [40].

Finally, it must be stated that the main feature of the approach proposed in this work is to complex relative motion between anatomical segments can be explained, from the kinematic point of view, as one degree of freedom joint. This concept can be understood in a similar way to the formulation introduced by Li [53, 54], who pointed out that in the case of human joints, the motor coordination establishes functional relations among the joint variables. The same idea has been also considered by Page and co-authors, who determined the optimal path traced by the instantaneous screw axis of human joints in cyclical motions with one functional degree of freedom [55, 56].

6 CONCLUSIONS

A comprehensive three-dimensional multibody model of the knee articulation during stance phase has been presented and discussed in this work. The kinematics of the knee model is obtained through the implementation of real kinematics data followed by an optimization study. The model shows that the combination of flexion/extension, internal/external rotation and abduction/adduction can be explained by the shapes of condyles if they are taken into account in the definition of the knee joint. In addition to the nominal kinematics produced by the condylar contact with the points representing the meniscus comes a component that appears to be elastic in the sense that it cannot be explained by a rigid-body mechanism. Constraint equations requiring each of these four points pairwise to remain in contact with each ellipsoid surface are now defined. Building on AnyBody's ability to handle implicit constraints, these equations simply take the form of the standard quadratic equation for an ellipsoid. The investigations reveal that this component can be fully explained by realistic elastic deformations of the soft tissues. In short, the internal/external rotation can be perceived as the sum of two terms: the baseline rotation from condyle shapes and the elastic deformation's contribution. It is also possible to conclude that the inter-individual difference can be explained by differences in condyles shapes. The implication is that the anatomical knee rather than being perceived as a six degree-of-freedom joint can be fully explained kinematically as a one degree-of-freedom joint with a complex behavior and an elastic component to it. This perception is very relevant in our continued search for valid mechanical models of the human knee.

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