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BIOMECHANICAL MULTIBODY KNEE MODEL WITH CONDYLE CONTACT: MODELING, SIMULATION AND ANALYSIS

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ABSTRACT: Over the last decades, the study of human body motion as a multibody system has undergone enormous developments. In a broad sense, most of the research in simulation of human tasks is based on the assumption that the human joints constraining the relative motion of the anatomical segments are ideal joints. However, this assumption fails to capture obvious physical properties of the natural human articulations. The purpose of this work is to develop a computational multibody model able to capture some of the basic properties of the knee joint and simulate human gait during the stance phase including the kinetics of the real knee.

1 INTRODUCTION

Over the last decades, the study of human body motion as a multibody system has received the attention of a good number of researchers. In a broad sense, most of the research in simulation of human tasks is based on the assumption that the human joints constraining the relative motion of the anatomical segments are ideal joints. However, this assumption fails to capture obvious physical properties of natural human articulations [1].

One biomechanical system that has fascinated the human spirit over the history is the human knee joint. The knee is the biggest, most complicated and possibly the most incongruent joint in human body. It is located between the body's two longest lever-arms: it sustains high forces, being also susceptible to chronic diseases and injuries. However, in spite of the considerable interest attracted over the last few decades and the progress made in understanding this complex joint, some aspects of its kinematics still remain uncertain [2] and uncertainties remain concerning its basic load-carrying mechanisms. The history of modeling and analyzing of the knee joint is long. Most of the available models are based on simple formulation because the knee is one of the most complex joints. In this sense, mathematical knee joint models have been used to obtain a better understanding of the complex mechanical behavior of the substructures, which include the human musculoskeletal system with knee joint. Already in 1917 Strasser [3] developed a simple knee model based on a four-bar mechanism, in which two bars represent the cruciate ligaments, and the remaining bars denote the femur and tibia bones. The same planar model was subsequently improved in 1974 by Menschik [4] by inclusion of two curves representing the femur and tibia articular surfaces. In 1976, Crowninshield et *al.* [5] presented an analytical model to study the biomechanics of the knee joint by a so-called inverse method. Wismans et *al.* (1980) [6] developed a 3D analytical model of the knee joint. This model takes into account not only the knee geometric properties, but also the static equilibrium of the system. They included a 3-D curved geometry of the tibia and femur surfaces, as well as a nonlinear elastic spring to model ligaments. Moeinzadeh et *al.* (1983) [7] developed a two dimensional dynamic model of the knee including ligament resistance, and specified a force and moment on the femur. Abdel-Rahman and Hefzy (1993) [8] and Engin and Tumer (1993) [9], presented more efficient approaches to solve the 2D problem identified by Moeinzadeh et *al.* [7], which was later extended to 3D. In this 3-D model, the femur and the tibia were considered as rigid bodies. External loads were specified, and ligamentous and contact forces were determined. In the 2-D dynamical model by Engin and Tumer [9] was included a patella component. In the literature for the patella-femoral joint, the most relevant models are the models developed by Essinger, et al. [12], Hirokawa (1991) and Heegaard et *al.* [11]. All of these models are quasi-static. In Hirokava's [10] 3-D model, each articular surface was divided into small patches, the quadriceps tendon was modeled using two strings; the patellar ligament was modeled using two linear springs. Heegard's [11] 3-D model was based on the finite element method and included a deformable patella and a rigid femur. The model by Essinger et *al.* [12] is characterized by being the only quasi-static model that incorporates the tibio-femoral as well as the patella-femoral joints. However, this model contains a limitation on the geometry of the patella and the patellar motions were simplified. Blankevoort and Huiskes [13], and Mommersteeg et *al.* [14], developed and experimentally demonstrated a 3-D knee model with surrounding soft tissue. More recently, Piazza and Delp (2001) presented a rigid body dynamic model of a total knee replacement performing a step up task [15]. Machado et *al.* (2010) [1] developed a planar multibody knee model.

The main purpose of this work is to develop a computational multibody model capable of capturing some of the basic properties of the knee joint and simulating human gait during the stance phase including the kinetics of the real knee. The aim of the model is not to be a perfect multibody knee model but rather to provide an understanding of the components of knee kinematics.

2 MODELING OF KNEE JOINT

2.1 GEOMETRIC DEFINITION

The biomechanical multibody system knee model proposed in this work is developed in the AnyBody Modeling System version 4.2 (AnyBody Technology, Aalborg, Denmark) [20]. It is composed of two rigid bodies, the tibia and the femur, whose characteristics are functions of the geometric and anatomic properties of the real bones. The segment masses are estimated according to Winter's work [16]. The extent of the tibia is defined by two nodes representing its superior/inferior extremities, i.e., knee and ankle respectively. Similarly, two nodes that correspond to the superior/inferior joints of the femur bone were defined. To create a model with more physiological realism, real bones geometries were added.

The AnyBody Modeling System uses a multibody system formulation based on a Cartesian approach that allows for the handling of closed kinematic chains. In the course of this process, the fundamental anatomical and biomechanical aspects of the knee and human locomotion are reviewed [1]. Traditionally, the tibiofemoral joints are developed by using revolute joints. In the proposed model, the tibio-femoral joint is created between an ellipsoid and a point using the facilities of AnyScript language. The method used implements the equation of an ellipsoid directly. It uses a linear measure, i.e., the length, between the center of the ellipsoid and the point. The first step to creating an ellipsoid joint is to identify ellipses that represent actual bone shapes. This fundamental idea is based on works that consider the profile shape of the femoral condyles the between 0° flexion contact point and the 90° flexion contact point as the ellipse curve [17].

To identify the ellipses that represent actual bone shapes, two planes for each condyle, are considered. The medial and lateral condyles are different and individual approximation of them by means of ellipses includes this basic property in the model. The relative position vector, defining the position of each plane, is defined, as a sagittal plane of each condyle. The following step is related to the creation of points on condyle surface defining the condyle surface. The main reason of this step was to implement the points' coordinates in ellipse equation. In the present work an ellipse with semi-axes of sizes a and b, centered on position (x_c,y_c) and whose axes rotated counter-clockwise by an angle θ was considered. Taking into account this ellipse features, the ellipse equation used in the present work is described as Eq. 1.

$$\left(\frac{\left(\cos(\theta)(x_i-x_c)+\sin(\theta)(y_i-y_c)\right)^2}{a^2}+\frac{\left(-\sin(\theta)(x_i-x_c)+\cos(\theta)(y_i-y_c)\right)^2}{b^2}-1\right)^2=0 \quad (1)$$

Please notice that the equation is squared to ensure that the values are always positive.

It turned out that the real distance from a point to an ellipse leads to complex expressions, so instead of the distance function, the value of the equation when a point is inserted, residual value, was used. Based on the ellipse equation it is possible to observe that every point that fulfills the ellipse equation will lead to a zero residual when inserted and any point outside the ellipse returns a non-zero residual when inserted into the equation. Thus, minimizing the residual will lead to an identification of the optimal ellipse.

Having identified the two ellipses that best represent each condyle, the joints between the condyles and the tibial plateau are defined by means of contact points on the tibia segment. For each condyle, two points representing the anterior and posterior horns of the lateral and medial menisci respectively were defined [18]. Figure 1 shows the knee model.

Fig. 1 Representation of the developed knee model. Two contact points defined on the anterior side are present in addition to the visible posterior contact points,

3 Results and Discussion

Kinematic data in the form of flexion/extension patterns are imposed on the model, and the resulting patterns of secondary knee movements (internal/external rotation and abduction/adduction) are observed and compared with data obtained from in vivo bone pin studies performed by Lafortune et *al.* [19].

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?

It was observed that the knee model approximated the average secondary motion patterns observed by Lafortune et *al.* (Figure 2).

However, the bone pin studies also revealed considerable inter-individual differences in the secondary motion patterns, and it was subsequently investigated whether the model was able to reproduce these differences with reasonable variations of bone shape parameters. This was accomplished by a parameter study in which the main parameters that define the geometry of the condyles were considered.

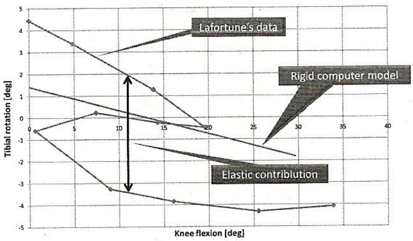

Fig. 2 Angular pattern of internal/external rotation [Deg] versus knee flexion angle [Deg] during stance from Lafortune's data and from the proposed knee model.

Regarding the experimentally observed inter-individual differences of the internal/external rotation, the model showed that they could be fully explained by differences in condylar shapes, as Figs 3-12 show. In the lateral and medial ellipsoids, the highest difference of the internal/external rotation is observed in the anterior/posterior direction (x direction) as was expected, because this direction corresponds to the knee flexion.
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 (Fig. 2). The model was therefore used to investigate whether this observed behavior could be explained by reasonable elastic deformations of the points representing the menisci in the model (Fig. 13). The investigations revealed that this was indeed the case. The menisci in this model showed a segmental difference in mobility with the posterior horns exhibiting less posterior displacement than the anterior horns in relation to the tibia during 11.459 deg of knee flexion. 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. 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 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.

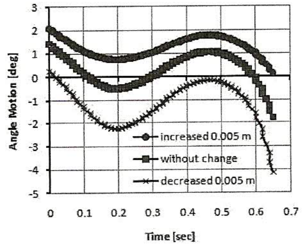

Fig. 3 Internal/external rotation with lateral semi-axe a change.

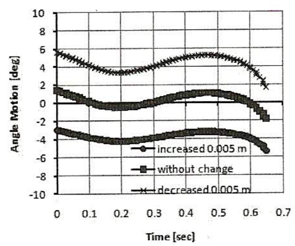

Fig. 4 Internal/external rotation with lateral semi-axe b change.

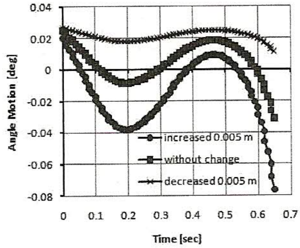

Fig. 5 Internal/external rotation with lateral center x_c change.

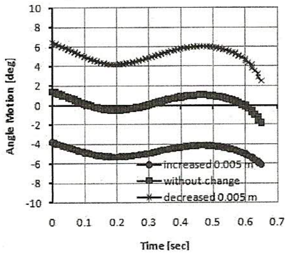

Fig. 6 Internal/external rotation with lateral center y_c change.

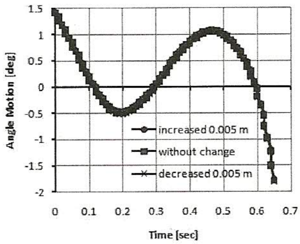

Fig. 7 Internal/external rotation with lateral rotation θ change.

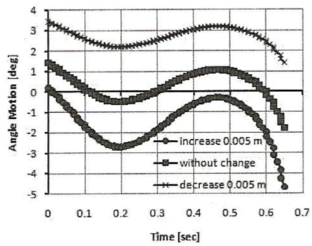

Fig. 8 Internal/external rotation with medial semi-axe a change.

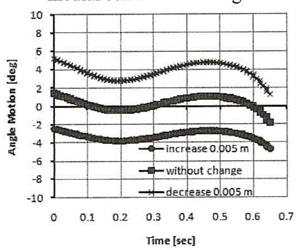

Fig. 9 Internal/external rotation with medial semi-axe b change.

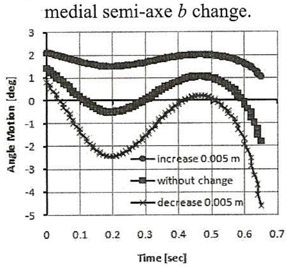

Fig. 10 Internal/external rotation with medial center x_c change.

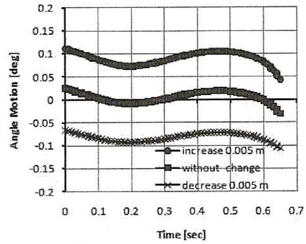

Fig. 11 Internal/external rotation with medial center y_c change.

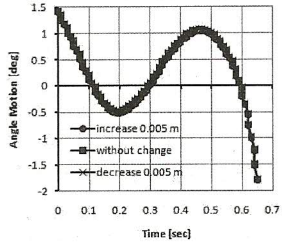

Fig. 12 Internal/external rotation with medial rotation θ change.

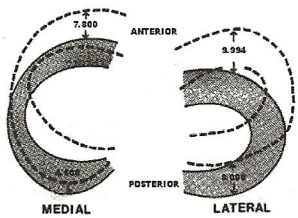

Fig. 13 Diagram of mean displacements (in millimeters) along tibial plateau.

4 Conclusions

A comprehensive three-dimensional model of the knee joint 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. The investigations reveal that this component can be fully explains 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 the different 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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