Szent István University

KINETICS AND KINEMATICS OF THE HUMAN KNEE JOINT UNDER STANDARD AND NON-STANDARD SQUAT MOVEMENT

Scientific theses

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1 INTRODUCTION, AIMS

1.1 Introduction

In this doctoral thesis, two novel mechanical models were presented: an analytical-kinetical model with regard to the kinetics of the human knee joint under non-standard squatting, and a numerical-kinematical model with regard to the local kinematics of the knee joint under standard squatting.

The horizontal movement of the centre of gravity is a known parameter regarding the kinetics of the human knee joint, although it might cause significant change in the patellofemoral and tibiofemoral forces.

My first aim is to create a new analytical-kinetical model, which can take the above mentioned parameter into account and show how it alters the patellofemoral and tibiofemoral kinetics.

Regarding the kinematics of the knee joint, the sliding-rolling phenomenon is examined between the connecting surfaces of commercial knee prostheses under standard squatting movement. This phenomenon has a governing effect on the wear of knee implants, although its ratio, magnitude and evolution along the functional arc of the flexion angle are currently unknown.

For this reason my second aim is the give a numerical description about this phenomenon.

1.2 Description of the aims

1.2.1 Effect of center of gravity

The kinetic description of squatting in the literature is limited to the so-called *standard squat*, where the torso is restrained to carry out only vertical motion, which means that practically the centre of gravity does not change its position during the squat. In contrast with this modelling approach, the *non-standard squat* includes the horizontal movement of the center of gravity as well.

The simplification, regarding the standard-squat model, has been widely used in the literature, and so far only a few authors pointed out, that the moving centre of gravity might have a significant effect on the kinetics of the knee joint. For this reason, the non-standard squat has been chosen for the object of investigation.

Several authors [Denham and Bishop, 1978, Schindler and Scott, 2011, Perry et al., 1975, Amis and Farahmand, 1996] bethought and surmised that the movement of the center of gravity should influence the patellofemoral forces, nevertheless none of these authors studied the phenomenon in depth.

Taking into consideration the earlier studies, a new analytical-kinetical model has been created, which among others parameters, includes the horizontal movement of the center of gravity as a dimensionless function.

By the use of the model, the patellofemoral and tibiofemoral forces can be calculated by simple algebraic equations, which are compared to experimental data from the available literature.

The results of the new analytical-kinetical model serve three goals: firstly, the obtained results can be used in other experiments or modelling issues as known loads during a specific type of squatting, secondly, the results show that how the moderate trunk motion – the horizontal movement of the center of gravity – affects the knee joint, and thirdly the results might be also useful for strength-based calculations in total knee replacement design.

A mention must be made that this model is aimed to describe the human, physiological knee joint.

1.2.2 Local kinematics in the knee joint

The other prior aim of this thesis is the connected to the local kinematics of the knee joint during squatting, more precisely, the **relative motion between the femoral and tibial condyles**. This motion is characterised in the literature as sliding-rolling. Due to its importance, this phenomenon is deeply investigated in the subject of gear-connections, however the available literature regarding the knee joint is rather limited.

The phenomenon of sliding-rolling has major impact on the mechanism of wear, therefore it has a significant effect on the lifetime and survivor-rate of the implants as well.

Up to now, only a limited number of articles dealt with this phenomenon. **These studies investigated the sliding-rolling with significant simplifications, for example: only in the initial phase of the motion (up to 20-30º of flexion angle), planar motion was considered and the geometry was also substantially simplified.**

Regarding its ratio, only rough estimations are available by the authors, and that is related to the beginning of the motion between 0˚ to 20-30˚. These results claim that in this initial segment, rolling is dominant, while above these certain angles sliding is primer.

These articles have also not investigated the ratio independently on the lateral and medial sides of the knee joint along the active functional arc (20-120º of flexion angle).

The determination of the correct sliding-rolling ratio along the functional arc of the knee joint (20-120º of flexion angle) has as substantial result for the development of implants. The reason lies in the fact, that the presence of this phenomenon causes different material abrasion compared to pure sliding or rolling alone, therefore, as a parameter during tribological tests, it has to be correctly given.

Until now, results from the earlier models, based on notable simplifications, were normative, which very likely underestimated the ratio of sliding-rolling.

In order to address this problem, which has been so far not studied in such depth regarding knee prostheses, multibody models are created from commercial total knee replacements.

These models consider real three-dimensional geometries, the effect of friction between the condyles, and collateral ligaments as well.

Another mention must be made that this kinematic model is aimed to describe the sliding-rolling phenomenon with regard to knee prostheses, and not to the physiological knee joint.

2 ANALYTICAL-KINETICAL MODEL

2.1 Model creation

After the comprehensive review of the literature, general conclusions have to be drawn in order to create a new model that is able to answer questions that until now have not been dealt with. In order to establish this new model, let us look at the main modelling questions and make decisions towards the model creation with a thorough explanation.

1st QUESTION: Which human locomotion should be modelled?

ANSWER: Considering two simple facts, it is adequate to choose the locomotion of squatting:

a) According to the studies of Reilly et al. [Reilly et al., 1972] and Dahlqvist et al. [Dahlqvist et al., 1982], the greatest magnitude of the patellofemoral forces ($\boldsymbol{F}_{pf}$, $\boldsymbol{F}_{pt}$, $\boldsymbol{F}_{q}$) appears during squatting motion,

b) A daily-used movement, which also has clinical importance as being a rehabilitation exercise,

c) From the mathematical point of view, the squatting movement provides more possibility to create a simpler but accurate analytical model.

For these reasons, the chosen locomotion is the squat.

2nd QUESTION: Should numerical or analytical model be used?

ANSWER: Although most of the earlier published mathematical models are considered as analytical models, only the work of Denham and Bishop [Denham and Bishop, 1978], Nisell et al. [Nisell et al., 1986] and Mason et al. [Mason et al., 2008] provide a closed-form analytical solution.

The rest of the mathematical models describe the phenomenon by non-linear equation systems that make the calculation clumsy. In addition, if a numerical model is used no closed form analytical correspondence can be created between the biomechanical factors such as patellar length-height, patellar tendon length or the anatomical angles.

As a major aim of this thesis, an analytical-kinetical model will be created thus the forces can be analytically derived from equilibrium equations.

3rd QUESTION: Should we consider static of dynamic model?

ANSWER: A significant question in the biomechanical research whether the human locomotion should be modelled with static or dynamic models.

The static-dynamic choice actually depends more on the locomotion. Regarding the running, it is adequate that the model is dynamic since the motion is carried out rapidly, thus significant inertial forces may arise.

Squatting in contrary is only in special cases rapid, in general carried out rather slowly. The clinical relevance of the squatting on the one hand is a lower-extremity strengthening exercise, while on the other hand a postoperative ACL rehabilitation program. A mention must be made that for patients with total knee arthroplasty, rapid squatting is contraindicated.

For the sake of clarity, the following duration(s) can be credited to normal squat exercise: Innocenti et al. [Innocenti et al., 2011] reported 20 sec of descending time in their study from 0˚ to 120˚ of flexion angle, while Fukagawa et al. [Fukagawa et al., 2012] discovered age-related correlation about deep squat kinematics. Their findings showed that the average normal deep squat duration situates between 3 and 6 sec as a function of age.

Based on this fact, we can conclude that the inertial effect on the patellofemoral forces under squat movement can be neglected as well.

Consequently, a static squat model will be used.

4th QUESTION: Should two- or three dimensional model be used?

ANSWER: Two-dimensional modelling is widely accepted and used in case of kinetic investigation, since the forces have their major effect in the sagittal plane and minor effect in the coronal plane [Singerman et al., 1994, Miller, 1991].

As for the modelling point of view, regarding the patellofemoral forces, a two-dimensional model can also provide accurate results with the advantage of easy handling.

Thus, the new analytical-kinetical model is consequently two-dimensional.

5th QUESTION: Should the geometry of the contact be considered?

ANSWER: The analytical-kinetical model will examine only the patellofemoral and tibiofemoral kinetics. The studies of Powers et al. [Powers et al., 2006], Innocenti et al. [Innocenti et al., 2011] and some practical applications regarding prostheses (GetAroundKnee™) suggest that a simple connection such as the hinge is applicable and satisfactory if only the kinetics is considered.

Therefore, the connection between the femur and tibia is represented with a hinge in the new analytical-kinetical model.

6th QUESTION: What muscles should be taken account and what can be disregarded?

ANSWER: The roll of the quadriceps tendon and the patellar tendon are indispensable, but ultimately what other ligaments and tendons can we neglect from our investigation?

In the study of Denham and Bishop [Denham and Bishop, 1978], it was well demonstrated with simultaneous electromyograph tracings that in case of balanced equilibrium the extensor effect upon the knee is minorly affected by actions in the hamstrings or the gastrocnemius muscles. Major activity was only reported in the quadriceps and in the soleus, while only occasional burst of activity, which helps to maintain balance, was noticed in the other muscle groups, so their effect can be safely disregarded.

According to the above-mentioned facts, only the quadriceps tendon and the patellar tendon are considered in the new mechanical model.

7th QUESTION: Should rigid of flexible bodies be used in the modelling?

ANSWER: Firstly, disregarding the deformation of the bones considerably simplifies the calculation, while only associates a moderate error to it and secondly it is a commonly applied simplification if we look at the earlier presented models in the literature review.

In the new, proposed mechanical model, the bodies are considered rigid.

8th QUESTION: Should force ratios or individual forces be used?

ANSWER: In several studies [Denham and Bishop, 1978, Van Eijden et al., 1986, Yamaguchi and Zajac, 1989, Hefzy and Yang, 1993, Gill and O'Connor, 1996] only the ratio of the patellofemoral forces can be obtained in a way that the quadriceps force is always assumed as a constant known force.

To all intents and purposes, these models neglect the fact that the quadriceps force changes during the motion and the change could be derived analytically.

In spite of the common assumption, another major aim of the new analytical-kinetical model is to derive the forces individually, thus the change, as a function of flexion angle, can be monitored and further studied.

9th QUESTION: Should the moving centre of gravity be implemented into the model?

ANSWER: The movement of the centre of gravity is a known phenomenon although it has been only slightly investigated how it alters the forces in the knee joint.

Firstly, it was shortly discussed by Perry et al. [Perry et al., 1975] that according to clinical experiences by locking the hip bone and leaning forward, practically moving the centre of gravity towards the knee, the amount of quadriceps force needed to stabilize the posture will be decreased thus the knee flexion will be carried out easier in case of patients with paresis. Although it was an appreciation of necessity, the question was not further analyzed.

They suspected that “**leaning forward a couple of centimetres could halve the patellofemoral forces**”, although they did not prove this hypothesis.

Due to the currently unknown effect of the moving center of gravity on the patellofemoral forces, this phenomenon, as a novel factor, will be implemented into the new analytical model.

2.2 Description of the analytical-kinetical model

2.2.1 Limitations and advancements

The following simplifications were considered regarding the new model:

- The model is quasi-static,
- The femur, tibia and patellar are considered as rigid bodies,
- The patellar tendon and the quadriceps tendon are inextensible,
- The line of action of the quadriceps is parallel with the femoral axis,
- The model is two-dimensional, the forces are only investigated in the sagittal plane,
- No contact forces ($\boldsymbol{F_s}$, $\boldsymbol{F_N}$) between the surfaces are considered,
- The connection between the femur and tibia is described with a hinge with one degree of freedom (no instantaneous center of rotation is considered),
- The load is derived from the total bodyweight of the person.

The new model is built to complement the earlier models, thus it holds several new features:

- Both standard and non-standard squatting movement can be investigated with this model,
- The body weight vector ($\boldsymbol{BW}$) can move vertically and horizontally (the centre of gravity is not fixed horizontally),
- The angle between the axis of tibia and the patellar tendon (β) is considered,
- The angle between the axis of tibia and the line of action of $\boldsymbol{BW}$ (γ) is considered,
- The angle between the axis of femur and the line of action of $\boldsymbol{BW}$ ($\delta = \alpha - \gamma$) is considered,
- The angle between the axis of tibia and the tibiofemoral force vector (φ) is considered,
- The rotation of the femur and tibia are not synchronized, but independent of each other,
- The experimentally determined dimensionless moment arms of the quadriceps force (λ_f) patellar tendon force (λ_p) and tibiofemoral force (λ_t) are taken into account.
- The patellofemoral compression force ($\boldsymbol{F_{pf}}$), the patellar tendon force ($\boldsymbol{F_{pt}}$), the quadriceps force ($\boldsymbol{F_q}$) and the tibiofemoral force ($\boldsymbol{F_{tf}}$) can be derived analytically in a closed form.

2.2.2 The analytical-kinetical model

The analytical-kinetical model is built up from three interconnected bodies: femur, tibia and patella. The model consists of equilibrium equations, which describe the forces, connected to the femur, tibia and patella during the squat (Figure 1).

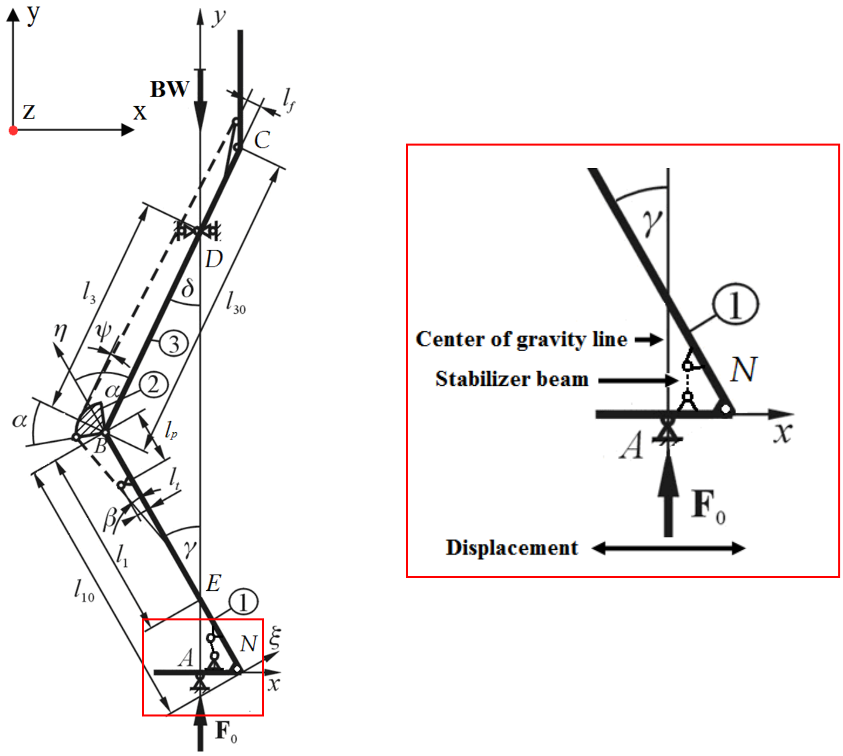


Figure 1: The analytical-kinetical model

The patella is assumed to rotate about z-axis at point B and so does the tibia similar to the model of Smidt [Smidt, 1973], Denham and Bishop [Denham and Bishop, 1978] or Mason et al. [Mason et al., 2008]. The line of action of $\boldsymbol{BW}$ intersects with the theoretical line of femur and tibia in point D and E. In order to keep the balance of the system, a stabilizer element has been incorporated into the model (Figure 1). The stabilizer beam has the feature that its length can change during the movement, thus moment can be transmitted at the ankle. Mention must be made that the kinetics of the ankle is not considered in this thesis.

At point D, a roller is applied which can move along the axis of femur, while another roller is applied at point A which can move along the axis of foot.

At point A, the ground reaction force is represented as $\boldsymbol{F_0}$ force, which equals to $\boldsymbol{BW}$. Strings, representing the quadriceps and patellar tendons, attach the rigid bodies to each other. The elongation of these strings is neglected.

PARAMETER	**DENOTATION**	**DEPENDENCY OF *α***
Length of tibia	l_{10}	No
Length of femur	l_{30}	No
Length of patellar tendon	l_p	No
Moment arm between the axis of tibia and the tibial tuberosity	l_t	No
Moment arm between the axis of femur and the line of action of the quadriceps force	l_f	No
Angle between the axis of femur and the quadriceps force vector	ψ	No
Intersected length of the axis of tibia and the instantaneous line of action of the ***BW***	l_1	Yes
Intersected length of the axis of femur and the instantaneous line of action of ***BW***	l_3	Yes
Angle between the axis of tibia and the patellar tendon	β	Yes
Angle between the axis of tibia and the line of action of ***BW***	γ	Yes
Angle between the axis of femur and the line of action of ***BW*** ($\delta = \alpha - \gamma$)	δ	Yes
Angle between the axis of tibia and the tibiofemoral force vector	φ	Yes

Table 1: Parameters of the analytical model

2.2.3 Mathematical description of the analytical-kinetical model

The aim is to derive the $\boldsymbol{F_q}$ quadriceps force, the $\boldsymbol{F_{pf}}$ patellofemoral compression force, the $\boldsymbol{F_{pt}}$ patellar tendon force and the $\boldsymbol{F_{tf}}$ tibiofemoral compression force. The calculation is carried out by the use of static equilibrium equations as a function of flexion angle. The free-body diagram of the model is shown in Figure 2.

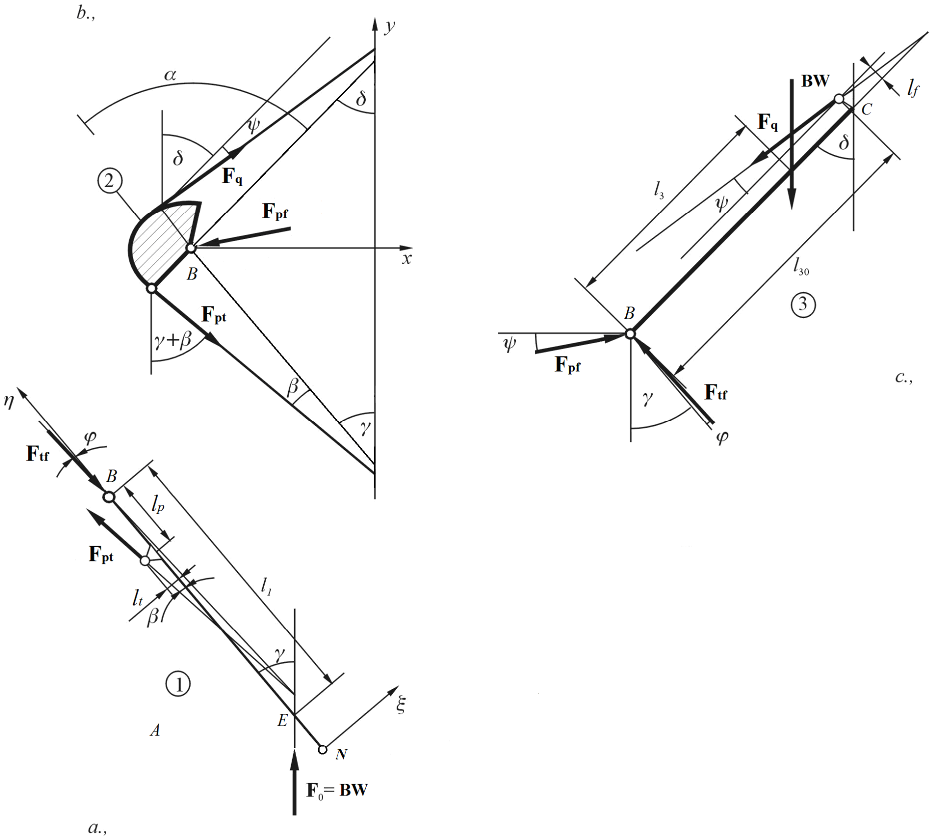


Figure 2: Free-body diagram (a, b, c)

The moment equation applied about the *z*-axis through point *B* on the tibia (Figure 2-a):

$$\sum M_{B1z} = 0 = -l_p \cdot F_{pt} \cdot \sin\beta(\alpha) - l_t \cdot F_{pt} \cdot \cos\beta(\alpha) + l_1(\alpha) \cdot BW \cdot \sin\gamma(\alpha) \quad (3.1)$$

From Eq. (3.1), the patellar tendon force can be derived as:

$$F_{pt}(\alpha) = BW \cdot \frac{l_1(\alpha) \cdot \sin\gamma(\alpha)}{l_p \cdot \sin\beta(\alpha) + l_t \cdot \cos\beta(\alpha)} \quad (3.2)$$

In order to simplify the results, dimensionless variables will be introduced in Table 2.

FORMULA	DESCRIPTION OF THE GEOMETRIC FUNCTIONS
$\lambda_1(\alpha) = l_1(\alpha)/l_{10}$	Dimensionless, intersected tibia length function
$\lambda_3(\alpha) = l_3(\alpha)/l_{30}$	Dimensionless, intersected femur length function
$\lambda_p(\alpha) = l_p / l_{10}$	Dimensionless length of patellar tendon
$\lambda_t(\alpha) = l_t / l_{10}$	Dimensionless thickness of shin
$\lambda_f(\alpha) = l_f / l_{30}$	Dimensionless thickness of thigh

Table 2: Dimensionless functions and constants

The acting forces will be calculated in a normalized form with respect to the force derived from the body weight (***BW***):

$$\frac{F_{pt}(\alpha)}{BW} = \frac{\lambda_1(\alpha) \cdot \sin\gamma(\alpha)}{\lambda_p \cdot \sin\beta(\alpha) + \lambda_t \cdot \cos\beta(\alpha)} \tag{3.3}$$

The scalar equilibrium equations related to the ξ - η coordinate system (fixed to the tibia) are the followings (Figure 2-a):

$$\sum F_{i\eta} = 0 = -F_{tf} \cdot \cos\varphi(\alpha) + F_{pt} \cdot \cos\beta(\alpha) + BW \cdot \cos\gamma(\alpha) \tag{3.4}$$

$$\sum F_{i\xi} = 0 = F_{tf} \cdot \sin\varphi(\alpha) - F_{pt} \cdot \sin\beta(\alpha) + BW \cdot \sin\gamma(\alpha) \tag{3.5}$$

After some simplifications we obtain:

$$\varphi(\alpha) = arctg\left[\frac{(\lambda_1(\alpha) - \lambda_p) \cdot tg\beta(\alpha) - \lambda_t}{\lambda_1(\alpha) \cdot tg\gamma(\alpha) + \lambda_p \cdot tg\beta(\alpha) + \lambda_t} \cdot tg\gamma(\alpha)\right] \tag{3.6}$$

By the use of angle φ the tibiofemoral force can be derived from Eq. (3.4) or Eq. (3.5) as:

$$\frac{F_{tf}(\alpha)}{BW} = \frac{F_{pt}}{BW} \cdot \frac{\cos\beta(\alpha)}{\cos\varphi(\alpha)} + \frac{\cos\gamma(\alpha)}{\cos\varphi(\alpha)} \tag{3.7}$$

The moment equilibrium equation applied about the z-axis through point B on the femur (Figure 2-c):

$$\sum M_{ib3} = 0 = l_f \cdot F_q \cdot \cos\psi(\alpha) + l_{30} \cdot F_q \cdot \sin\psi(\alpha) - l_3(\alpha) \cdot BW \cdot \sin\delta(\alpha) \tag{3.8}$$

Taking into account that $\delta = \alpha - \gamma$, and assuming that $\psi = 0$, the quadriceps force in the tendon is:

$$\frac{F_q(\alpha)}{BW} = \frac{\lambda_3(\alpha) \cdot \sin(\alpha - \gamma(\alpha))}{\lambda_f} \tag{3.9}$$

The $\psi = 0$ assumption means that the direction of the resultant acting forces in the quadriceps muscle are parallel to the axis of femur. This is a widely accepted and used approximation.

The scalar equilibrium equations related to the patella in the x - y coordinate system (Figure 2-b):

$$\sum F_{ix} = 0 = F_q(\alpha)\cdot \sin\delta(\alpha) + F_{pt}(\alpha)\cdot \sin\big(\gamma(\alpha)+\beta(\alpha)\big) + F_{pf_x} \qquad (3.10)$$

$$\sum F_{iy} = 0 = F_q(\alpha)\cdot \cos\delta(\alpha) - F_{pt}(\alpha)\cdot \cos\big(\gamma(\alpha)+\beta(\alpha)\big) + F_{pf_y} \qquad (3.11)$$

From Eq. (3.10) and Eq. (3.11) the magnitude of the patellofemoral compression force can be derived by using x,y coordinates with respect to the body weight force:

$$\frac{F_{pf}(\alpha)}{G} = \frac{\sqrt{F_{pf_x}^2 + F_{pf_y}^2}}{G} = \frac{\sqrt{F_q(\alpha)^2 + F_{pt}(\alpha)^2 - 2\cdot F_q(\alpha)\cdot F_{pt}(\alpha)\cdot \cos(\beta(\alpha)+\delta(\alpha)+\gamma(\alpha))}}{BW} \qquad (3.12)$$

2.2.4 Remarks about the model

Since all the forces are mathematically described by the use of the above-mentioned equations, the patellofemoral forces can be estimated in the knee joint during squatting. Nevertheless, the derived equations include multiple dimensionless functions and constants such as $\lambda_1(\alpha)$, $\lambda_3(\alpha)$, λ_p, λ_t, λ_f, $\beta(\alpha)$, $\gamma(\alpha)$, which are currently unknown.

Without these parameters, the analytical-kinetical model cannot be solved and used, thus as another aim of this thesis is to determine these certain parameters and variables by means of experiments.

3 EXPERIMENTS ON HUMAN SUBJECTS

3.1 Aim of the experiment

The new analytical-kinetical model of the non-standard squat has been fully described, but as it was mentioned, seven important factors and variables ($\lambda_1(\alpha)$, $\lambda_3(\alpha)$, λ_p, λ_t, λ_f, $\beta(\alpha)$ and $\gamma(\alpha)$) are missing to solve the equations.

Among the above mentioned functions, only $\beta(\alpha)$ function has been investigated and published earlier by several authors [Van Eijden et al., 1986, Yamaguchi and Zajac, 1989, Gill and O'Connor, 1996, Victor et al., 2010], while the $\lambda_1(\alpha)$, $\lambda_3(\alpha)$ and $\gamma(\alpha)$ functions have not yet appeared in other models or publications. Thus, by all means, they have to be determined experimentally. Regarding the $\beta(\alpha)$ function, the authors were all in agreement.

Beside the motivation to comply the mathematical model with the necessary parameters, the experiments are also meant to prove the following hypotheses:

1. The horizontal movement of the center of gravity line changes its position during squatting, in contrary with other assumption [Cohen et al., 2001],
2. The horizontal movement of the center of gravity line can be derived with empirical function during squatting.

3.2 The measurement setup and the experiment

The geometrical parameters were defined in Figure 3:

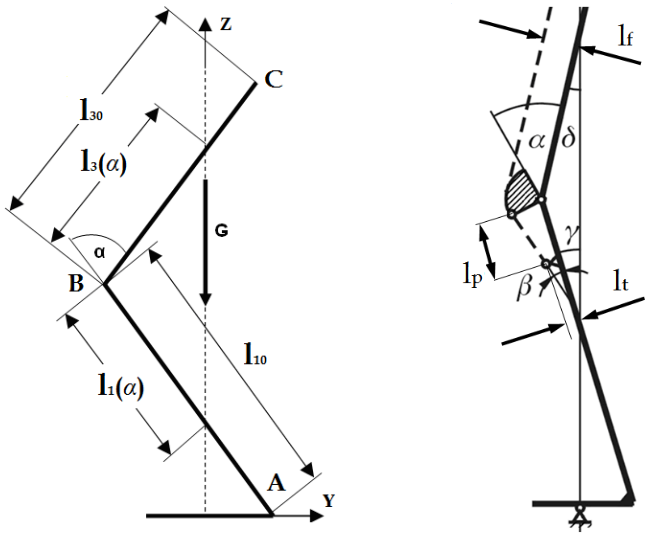


Figure 3: Geometrical parameters

These quantities are listed in Table 1 and Table 2.

For the experiments, MOM type "A" class ETP 7922 dynamometers [Kaliber] were used from the Kaliber Ltd, while for data process, a Spider 8 multi-channel PC measurement electronics [HBM] was used from the HBM GmbH.

The experiments were carried out on 16 persons (9 males and 7 females) between 21 and 27 years old. The mean (± standard deviation) weight of all participants was 72.2 ± 17.4 kg respectively. The measurements were carried out in two parts. 9 people at the first experiment and couple of weeks later the other 7.

The measurement was carried out as follows: after calibration, three dynamometers were set under the force platform (Figure 4).

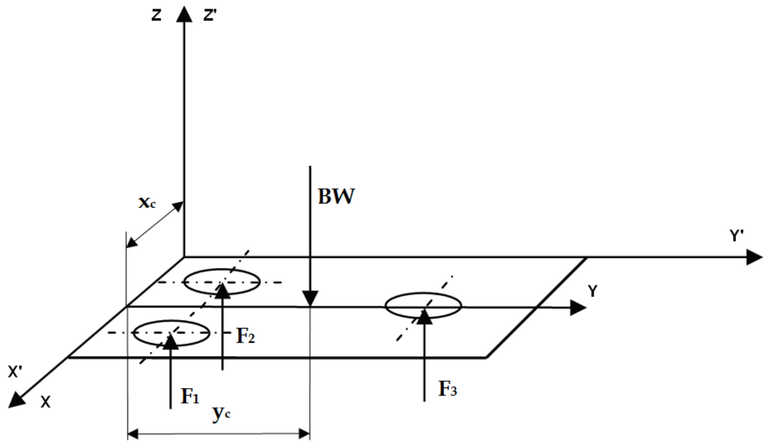


Figure 4: Position of dynamometers

When the markers were fixed, the subjects steped up on the plates and their center of gravity were measured in six positions. Both x_c and y_c position of the center of gravity line could be measured, although only the y_c component was investigated.

Figure 5: Squatting positions

During the squatting motion, the subject has to keep three conditions:

1. stretched arms,
2. heels adjusted to the metal frame at initial standing position,
3. keeping the different positions for 3 seconds.

During squatting, the heel naturally ascends which is allowable for the test. After the measurements of the center of gravity position (y_c) the averaged value and its standard deviation have been calculated in each squatting status.

3.3 Construction of the dimensionless quantities

After measuring the yc coordinate of the center of gravity line of all persons, the theoretical lines of the bone axes and the intersection of the center of gravity had to be constructed. The constructions were carried out in the AUTOCAD by importing the photos into the program. Since all of the dimensions of the platforms were known, the measured y_c component of the center of gravity could be drawn in each position by the software (Figure 5), and with a construction method, all the demanded quantites could be determined in each squatting status.

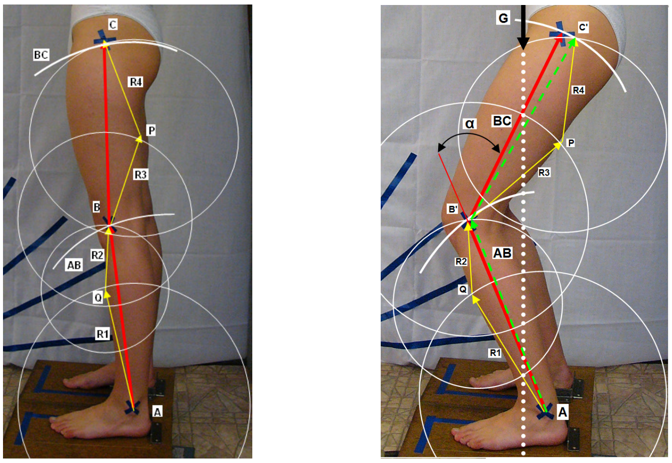


Figure 5: Construction of the center of gravity line

Every parameter has been determined in each status (Figure 6):

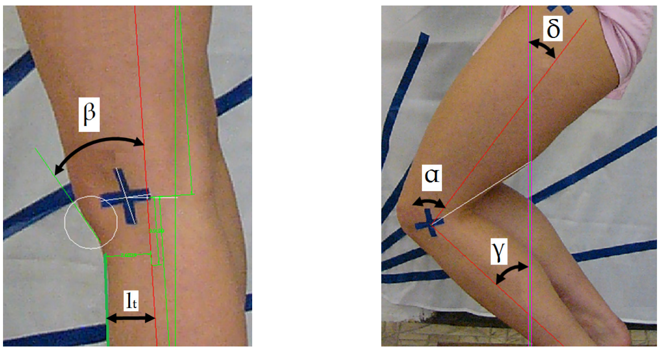


Figure 6: Determining the parameters

The construction was followed by statistical process, where both the contstants and functions were determined and summarized in a Table 3. $\gamma(\alpha)$ function was also given in a dimensionless form as follows:: $\Phi(\alpha)=\gamma(\alpha)/\alpha$. Table 3 also includes the standard deviation of the parameters.

	C1	C2	SD	r^2
$\lambda_1(\alpha)$ [-]	0.492	0.0024	0.15	0.65
$\lambda_3(\alpha)$ [-]	0.86	-0.0022	0.22	0.63
$\beta(\alpha)$ [°]	26.56	-0.3861	14	0.95
$\Phi(\alpha)$ [-]	0.567	-0.0026	0.081	0.735
λ_p [-]	0.11	0	0.018	-
λ_p [-]	0.1475	0	0.043	-
λ_f [-]	0.164	0	0.028	-

Table 3: Functions* and constants of the analytical-kinetical model

*The form of the function: $f(\alpha) = C1 + C2\cdot \alpha$

3.4 Remarks

In summary, a new method was presented to experimentally determine the horizontal movement of the center of gravity line and other anthropometrical constants-functions.

By knowing the above-mentioned parameters, the results can be extended for further use: the earlier introduced analytical-kinetical model – where the load case is based on the obtained λ functions – is able to predict now both the patellofemoral- and tibiofemoral forces.

4 LOCAL KINEMATICS OF THE KNEE JOINT

4.1 Model creation

After reviewing the advancements of other authors in the modelling of sliding-rolling phenomenon, several questions were conceived regarding the white spots of the research. By gathering these questions, solid directions about the properties of the new model could be drawn and controversial or disregarded factors could be re-evaluated.

1st QUESTION: Should numerical or analytical model be used?

ANSWER: The complex geometry of the condyles and the challenging contact issue between the bodies make the description of the phenomenon impossible with algebraic equations, thus an analytical model is not advised.

Due to the complexity of the geometry and the phenomenon itself, only a numerical model is fitting for use.

2nd QUESTION: Which human locomotion should be modelled?

ANSWER: On the one hand, our analytical model is based on the squatting movement, thus it is adequate to use this motion as basis. Moreover, the load of the knee joint during squatting is certainly higher than in most of other activities, therefore it is a good reason to work further on this movement.

For these reasons, the chosen locomotion is the squat.

3rd QUESTION: Should rigid or flexible bodies be used in the modelling?

ANSWER: It has been proven that the use of rigid bodies causes negligible error in the kinematical [Baldwin et al., 2009, Halloran et al., 2005a, Halloran et al., 2005b] or in the kinetical [Baldwin et al., 2009] investigations, while the calculation time is only the half, one forth of the simulations with flexible bodies.

In the new, proposed numerical model, the bodies are rigid.

4th QUESTION: Should two- or three dimensional model be used?

ANSWER: The human knee joint is practically a three-dimensional joint that incorporates secondary rotations in the frontal (represented as abduction/adduction) and transverse (represented as axial rotation) planes of motion. The assumption that knee joint movements can be represented by planar motion in the sagittal plane excludes the potential effect of axial rotation (the so-called "screw home mechanism") on the calculation of the sliding-rolling phenomenon.

Thus, one principal limitation of the earlier published models [O'Connor et al., 1990, Chittajallu and Kohrt, 1999, Hollman et al., 2002] is that the contact geometry of the knee joint is oversimplified. According to O'Connor et al. [O'Connor et al., 1990] the slip ratio (thus the sliding-rolling ratio as well) is sensitive to the shape, or the assumed shape, of the tibia plateau. Considering this fact, simplification of the geometry very likely has a significant effect on the sliding-rolling ratio.

In addition, several authors agree, that their approach [Wilson et al., 1998, Hollman et al., 2002] is only a rough approximation due to the simplified geometry.

Thus, the new numerical model is consequently three-dimensional.

5th QUESTION: Should the sliding-rolling phenomenon be examined between the tibiofemoral or the patellofemoral connection?

ANSWER: Typically, wear (regarding knee replacements) appears between the tibiofemoral contact due to the constant sliding and rolling motion. For this reason, almost with no exceptions, most studies put the emphasis on the tibiofemoral connection [Wimmer and Andriacchi, 1997, O'Brien et al., 2013, Blunn et al., 1992, Hood et al., 1983, Wimmer et al., 1998, Blunn et al., 1991, Blunn et al., 1994, Davidson et al., 1992]. According to these studies, the new numerical model will also be designed to examine the tibiofemoral contact with regard to the sliding-rolling phenomenon.

According to the above-mentioned studies, the new numerical model sets the emphasis on the tibiofemoral connection.

6th QUESTION: What muscles should be taken account and what can be disregarded?

ANSWER: *Only the quadriceps tendon and the patellar tendon are considered in the new numerical-kinematical model, similarly to the analytical-kinetical model.*

7th QUESTION: Should friction between the bodies be defined?

ANSWER: The earlier were in agreement that, due to the synovial fluid, the friction between the condyles can be neglected (0.001-0.004), although no studies were reported about the possible effect of friction on the sliding-rolling ratio.

Since multibody models can easily incorporate contacts with friction, it is worth involving this specific factor.

For this reason, friction is incorporated into the numerical model.

8th QUESTION: Should the slip ratio or other quantity be used to define the sliding-rolling phenomenon?

ANSWER: In the literature, several types of slip ratios, sliding-rolling ratios, etc, appear. Many of these sources refer to a so-called slip ratio defined by O'Connor et al. [O'Connor et al., 1996]. The slip ratio is defined as follows: the slip ratio of one represents pure rolling, and a slip-ratio of infinity represents pure slip while intermediate values represent combination of roll and slip together.

This definition does not make the phenomenon easily understandable, or gives a well-defined ratio, since between one and infinity the difference is infinite. Another ratio has to be introduced, which can describe this local motion preferably as a percentage.

For this reason, a new ratio will be introduced which can describe the sliding-rolling phenomenon as a percentage.

9th QUESTION: Should real bone structure geometry be examined or prosthesis geometry?

ANSWER: The condyles are covered by meniscus, which fulfil several purposes: on the one hand, it stabilizes the knee that no severe lateral or medial slip would occur, and on the other hand, it disperses the load on the surface. In order to model real human bone geometry, the meniscus system should be modelled as well, which highly complicates the work.

It is more advisable to work with current prosthesis geometries, where no meniscus modelling is included.

Therefore, prosthesis geometries are used in the numerical model.

10th QUESTION: Between what angles should the sliding-rolling ratio be examined?

By summarizing the findings of the experimental and mathematical (numerical) literature, in case of experimental testing of prosthesis materials the sliding-rolling ratios are widely applied between 0.3-0.46. This assumption has been proven correct, although at higher flexion angles, presumably, the sliding-rolling ratio changes significantly [Nägerl et al., 2008, Reinholz et al., 1998], but the results related to the sliding-rolling ratio above 30˚ of flexion angle are rather limited.

Since the pattern of the sliding-rolling phenomenon has not been thoroughly investigated in full extension, the aims are the followings:

I. The pattern and magnitude of the sliding-rolling ratio have to be determined between 20-120˚ of flexion angle on several prosthesis geometries. This segment is considered as the fundamental active arc, which is totally under muscular control and involves most of our daily activities.

II. The change of the sliding-rolling ratio has to be investigated, as a function of different commercial and prototype prostheses. This should help to find the lower and upper limit of the sliding-rolling ratio between the condyles.

III. The possible effect of the lateral and medial collateral ligaments on the sliding-rolling ratio should be examined. It is unknown how much influence has the ligaments on the local kinematics, therefore as a first step, an investigation will be carried out by involving them into the multibody system.

4.2 Description of the numerical-kinematical model

4.2.1 Limitations and advancements

In this new model, the investigation is restricted to the sliding-rolling ratio and the contact kinetics under standard squat movement. The new numerical-kinematical model includes some simplifications as follows:

- The bones, such as the tibia, patella and femur were assumed as rigid bodies, since the influence of deformation in this study is neglected,
- The patellar tendon modelled as an inextensible spring,
- The quadriceps is modelled as one single linear spring,
- No cruciate ligaments were modelled.

The new model complements the earlier models in some extent, thus it holds new features:

- The numerical-kinematical model is three-dimensional, based on commercial prosthesis geometries,
- Both lateral and medial sliding-rolling ratio can be studied due to the three dimensional surfaces,
- Realistic friction condition is considered between the contact surfaces e.g. patellofemoral and tibiofemoral connection,
- Kinetical investigation is also possible with this model.

4.2.2 Kinematical constrains and properties

For the calculation, five multibody models were built with MSC.ADAMS program system. The geometric models were mapped by a monochrome scanner at the Szent István University. The following boundary conditions were applied on each model:

- According to the literature, the spring and damping constant of the quadriceps were set to 40 N/mm és 0.15 Ns/mm,
- The body weight (***BW***) was set to 800 N, and it was applied on the femur distalis,
- The femur distalis was constrained by a GENERAL POINT MOTION, where all the coordinates can be prescribed (Figure 7). Only one prescription was set: the endpoint of the femur (distalis) can only perform translational motion along the y-axis.
- The ankle part of the model was constrained by a SPHERICAL JOINT, which allows rotation about all axes, but no translational motions are permitted in that point (Figure 7).
- Between the femur, tibia and patella, CONTACT constraints were set according to Coulomb's law with respect to the very low static and dynamic friction coefficients ($\mu_s = 0.003$, $\mu_d = 0.001$) similarly to real joints.
- The following material properties were set: Young modulus$_{Femur}$: 19 GPa, Poisson ratio$_{Femur}$: 0.3, Young modulus$_{Tibia}$: 1 GPa, Poisson ratio$_{Tibia}$: 0.46.

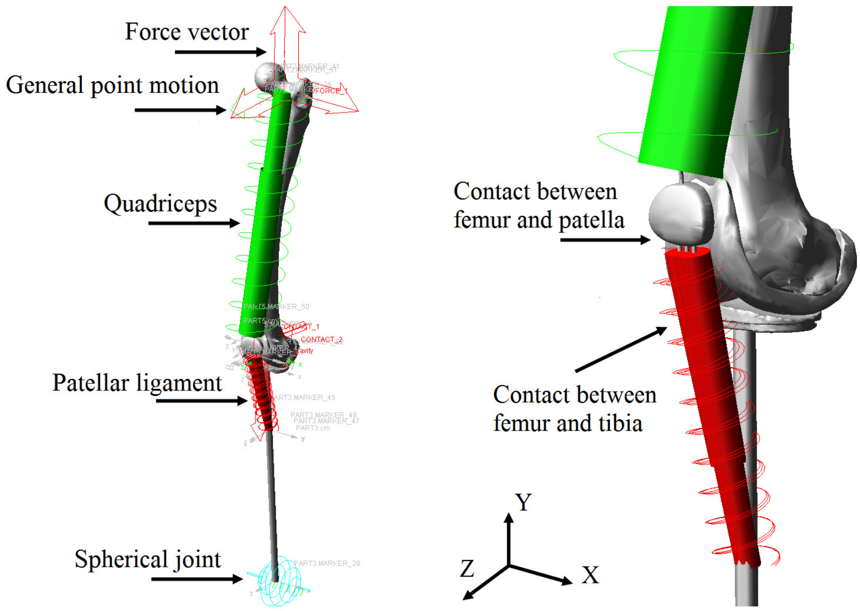


Figure 7: Multibody model in MSC.ADAMS

4.2.3 Calculation method

The following kinematic quantities can be directly calculated by the MSC.ADAMS during the simulation of the motion as a function of time:

- $\bar{r}_{Ci}(t)$: Vector-scalar function, which determines the instantaneous position of the connecting points of two bodies defined in the absolute coordinate system (Figure 3.34). If i = 1, contact between femur and tibia, if i = 2, contact between femur and patella.
- $\bar{r}_{CMF}(t)$, $\bar{r}_{CMT}(t)$, $\bar{v}_{CMF}(t)$, $\bar{v}_{CMT}(t)$, $\bar{\omega}_{CMF}(t)$, $\bar{\omega}_{CMT}(t)$: Vector-scalar functions, which determine the instantaneous position of the center of mass (CM_i), velocity and angular velocity of the femur (F) and the tibia (T) defined in the absolute coordinate system (Figure 8).
- $\bar{e}_{Ci}(t)$: Vector-scalar function (unit-vector), which determines the instantaneous tangent vector respectively to the contact path defined in the absolute coordinate system (Figure 9).

Besides the kinematic quantities, MSC.ADAMS software can calculate kinetic quantities as well, for example:

- Contact forces between the contact surfaces, reaction forces and moments in the applied constraints or forces in the springs.

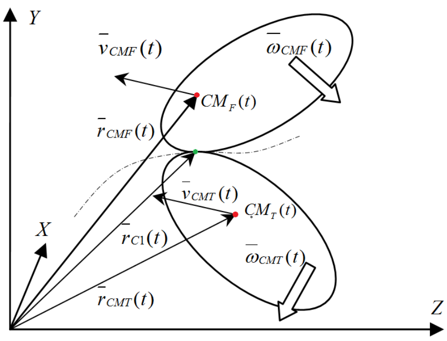


Figure 8: Kinematical quantities I.

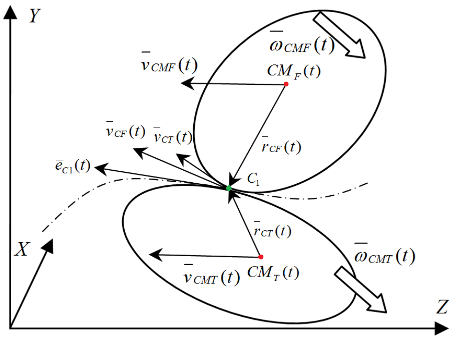


Figure 9: Kinematical quantities II.

To obtain the velocity of a point – in our case point C_1 – the following calculation algorithm is applied [Csizmadia and Nándori, 1997]:

$$\bar{v}_{CF}(t) = \bar{v}_{CMF}(t) + \bar{\omega}_{CMF}(t) \times \bar{r}_{CF}(t) \tag{3.43}$$

$$\bar{v}_{CT}(t) = \bar{v}_{CMT}(t) + \bar{\omega}_{CMT}(t) \times \bar{r}_{CT}(t) \tag{3.44}$$

where,

$$\bar{r}_{C1}(t) = \bar{r}_{CMF}(t) + \bar{r}_{CF}(t) \rightarrow \bar{r}_{CF}(t) = \bar{r}_{C1}(t) - \bar{r}_{CMF}(t) \tag{3.45}$$

$$\bar{r}_{C1}(t) = \bar{r}_{CMT}(t) + \bar{r}_{CT}(t) \rightarrow \bar{r}_{CT}(t) = \bar{r}_{C1}(t) - \bar{r}_{CMT}(t) \tag{3.46}$$

By substituting Eq. (3.45) into Eq. (3.43) and Eq. (3.46) into Eq. (3.44) we obtain:

$$\bar{v}_{CF}(t) = \bar{v}_{CMF}(t) + \bar{\omega}_{CMF}(t) \times \left(\bar{r}_{C1}(t) - \bar{r}_{CMF}(t)\right) \tag{3.47}$$

$$\bar{v}_{CT}(t) = \bar{v}_{CMT}(t) + \bar{\omega}_{CMT}(t) \times \left(\bar{r}_{C1}(t) - \bar{r}_{CMT}(t)\right) \tag{3.48}$$

Now, the velocities with respect to the femur and tibia are determined in the contact point, in the absolute coordinate system (Figure 10). By multiplying equation (3.47) and (3.48) with the $\bar{e}_{C1}(t)$ unit vector, we can derive the tangential scalar component of the femoral and tibial contact velocities with respect to the contact path:

$$v_{CFt}(t) = \left[\bar{v}_{CMF}(t) + \bar{\omega}_{CMF}(t) \times \left(\bar{r}_{C1}(t) - \bar{r}_{CMF}(t)\right)\right] \cdot \bar{e}_{C1}(t) \tag{3.49}$$

$$v_{CTt}(t) = \left[\bar{v}_{CMT}(t) + \bar{\omega}_{CMT}(t) \times \left(\bar{r}_{C1}(t) - \bar{r}_{CMT}(t)\right)\right] \cdot \bar{e}_{C1}(t) \tag{3.50}$$

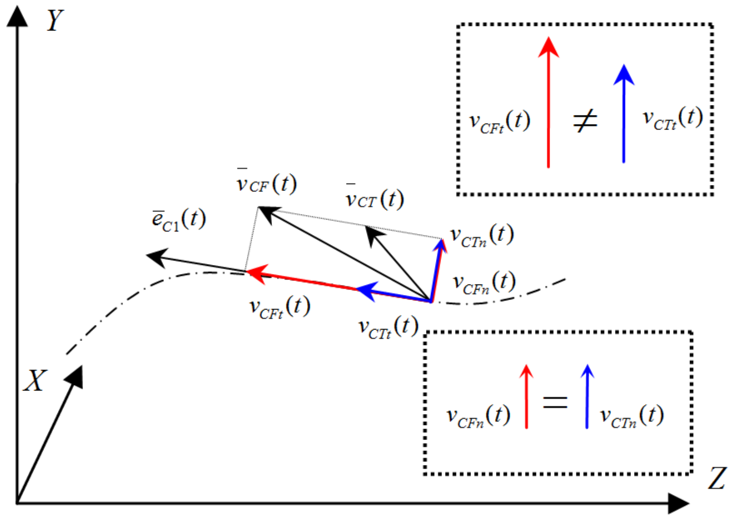


Figure 10: Tangential- and normal velocity components

The tangential scalar components are only valid, if the following condition is satisfied [Szendrő, 2007, Vörös, 1970]:

$$v_{CF\,n}(t) = v_{CT\,n}(t) \tag{3.51}$$

This means that the normal scalar components of the femoral and tibial contact velocities have to be equal, otherwise, the two surfaces either would be crushed into each other or would be separated.

Since the scalar contact-velocities are available, by integrating them over time the connecting arc lengths with respect to the femur and tibia can be calculated as:

$$s_{femur}(t) = \int v_{CFt}(t)\cdot dt = \int \left[\bar{v}_{CMF}(t) + \bar{\omega}_{CMF}(t)\times\left(\bar{r}_{C1}(t) - \bar{r}_{CMF}(t)\right)\right]\cdot \bar{e}_{C1}(t)\cdot dt \tag{3.52}$$

$$s_{tibia}(t) = \int v_{CTt}(t)\cdot dt = \int \left[\bar{v}_{CMT}(t) + \bar{\omega}_{CMT}(t)\times\left(\bar{r}_{C1}(t) - \bar{r}_{CMT}(t)\right)\right]\cdot \bar{e}_{C1}(t)\cdot dt \tag{3.53}$$

By having determined the arc lengths on both connecting bodies, the sliding-rolling ratio can be introduced and denoted as follows:

$$\chi(t) = \frac{\Delta s_{tibiaN}(t) - \Delta s_{femurN}(t)}{\Delta s_{tibiaN}(t)} \tag{3.54}$$

where,

$$\Delta s_{femurN}(t) = s_{femurN}(t) - s_{femurN-1}(t) \tag{3.55}$$

$$\Delta s_{tibiaN}(t) = s_{tibiaN}(t) - s_{tibiaN-1}(t) \tag{3.56}$$

are the corresponding incremental differences of the connecting arc lengths.

The sliding-rolling function, or sliding-rolling ratio, is defined as the difference between of an incremental distance travelled (Δs_{tibiaN}) on the tibia and the incremental distance travelled (Δs_{femurN}) on the femur over the incremental distance travelled (Δs_{tibiaN}) on the tibia. *N* denotes an arbitrary arc length during the connection.

By this function, exact conclusions can be drawn about the sliding and rolling features of the motion. A sliding-rolling ratio of zero indicates pure rolling, while one describes pure sliding. If the ratio is between zero and one, the movement is characterized as partial rolling and sliding. For example, a sliding-rolling ratio of 0.4 means 40% of sliding and 60% of rolling. A positive ratio shows the slip of the femur compared to the tibia. If the sign is negative, than the tibia has higher slip compared to the femur.

It is desirable to determine the sliding-rolling ratio as a function of flexion angle rather than as a function of time. To do so, the flexion angle (α) was derived by integrating the angular velocities of the femur and tibia about the *x*-axis over time and taking into account that the model was set in an initial 20 degree of squat at the beginning of the motion.

$$\alpha(t) = \int \omega_{CMFx} \cdot dt + \int \omega_{CMTx} \cdot dt + 20 \tag{3.57}$$

Since $\alpha(t)$ function has been determined, time can be exchanged to flexion angle and the sliding-rolling function can be plotted as a function of flexion angle:

$$\chi(\alpha) = \frac{\Delta S_{tibiaN}(\alpha) - \Delta S_{femurN}(\alpha)}{\Delta S_{tibiaN}(\alpha)} \tag{3.58}$$

5 RESULTS

5.1 Results regarding the analytical-kinetical model

Since the required parameters and variables are available, the analytical-kinetical model can be evaluated and compared to the results of other authors. However, let us first investigate the effect of the horizontally moving center of gravity on the standard squat model described by Mason et al. [Mason et al., 2008], and after the results of the non-standard squat model.

5.1.1 Effect of center of gravity – Standard squat model

Mason et al. [Mason et al., 2008] published a comprehensive review about the patellofemoral joint forces, where they combined the results and models of other authors in order to give a fully analytical approach for calculating the patellofemoral forces. By the use of this model, the so-called „*net knee moment”* has been determined in its original- (standard squat) and modified (non-standard squat) state (Figure 11).

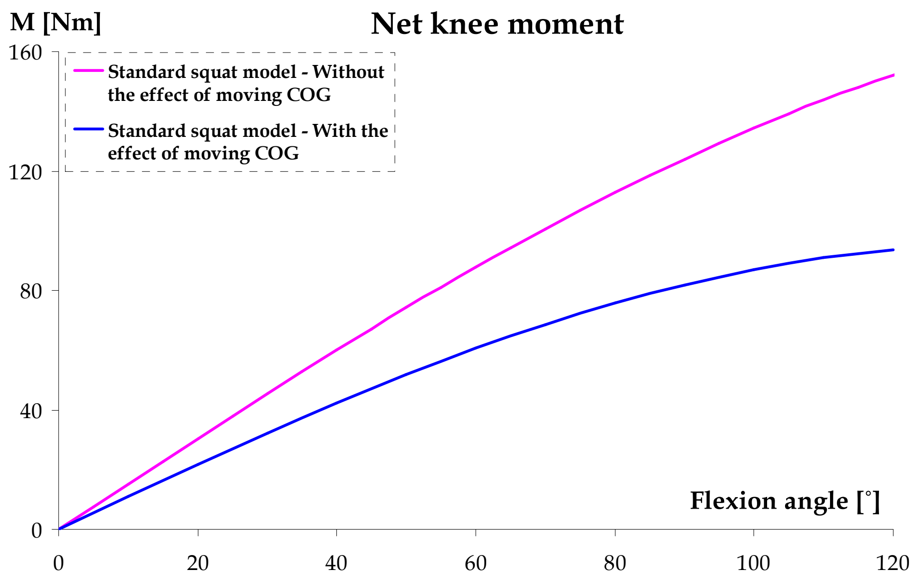


Figure 11: Net knee moments of the model of Mason et al. [Mason et al., 2008]

In order to see the influence of the moving center of gravity in numbers, the patellofemoral forces and the net knee moments have been recalculated, as percentage difference, and compared between the standard (fixed center of gravity) and non-standard (moving center of gravity) squat in this model [Mason et al., 2008].

Flexion angle (α)	ΔM_h	ΔF_q	ΔF_{pt}	ΔF_{pt}
30°	20%	17%	17%	18%
60°	28%	24%	24%	24%
90°	34%	38%	38%	38%
120°	44%	25%	25%	25%

Table 4: Percentage difference between Standard and Non-standard squat [Mason et al., 2008]

The obtained results were calculated as follows:

$$\Delta K = \left(1 - \frac{K_{nem-standard}}{K_{standard}}\right) \cdot 100 \qquad (4.2)$$

Where, *K* can be any quantity (force, moment or displacement). *ΔK* can provide a percentage difference of a standard quantity compared to a non-standard quantity (here standard and non-standard relates to the squat motion). The incorporation of the moving center of gravity significantly lowers the patellofemoral forces (17-38%) along the calculated domain. This lowering effect on the patellofemoral forces (average 27.5%) corresponds very well with the result of Kulas et al. [Kulas et al., 2012] who also investigated the effect of the moderate forward trunk lean condition and observed 24% lower peak ACL forces!

By these results, not only the necessity of this factor in the modelling has been confirmed, but it also has been shown that this factor surely decreases the forces in the tendons (and ligaments). For this reason the movement of the center of gravity, as a new parameter in the modelling, is eminently valid.

5.1.2 Effect of center of gravity – Non-standard squat model

Let us look at the results of the new analytical-kinetical model.

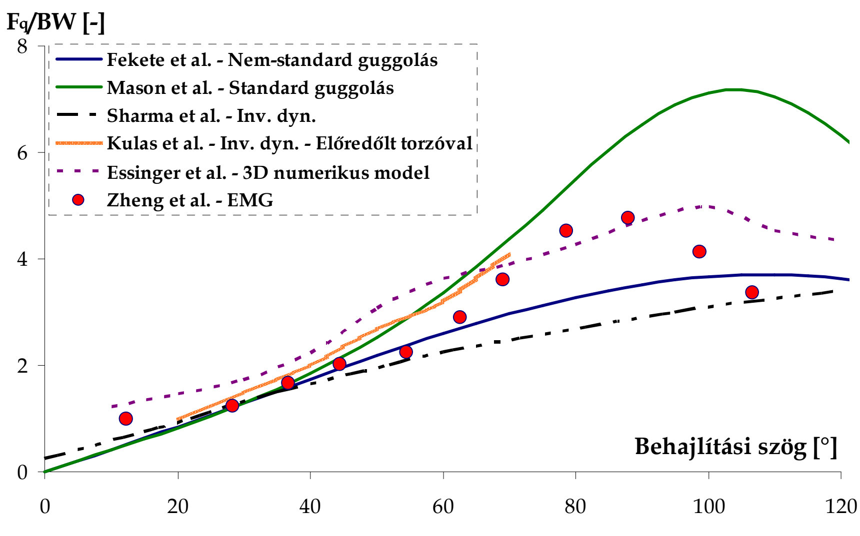


Figure 12: Quadriceps tendon force

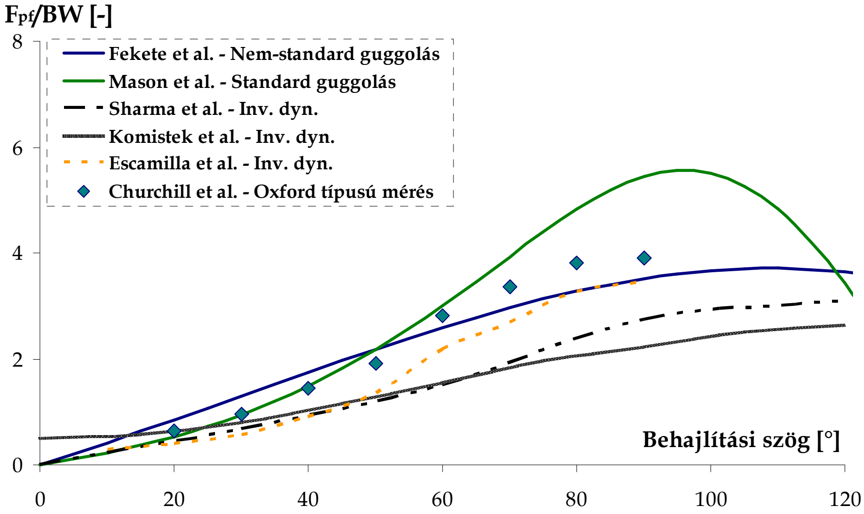


Figure 13: Patellofemoral compression force

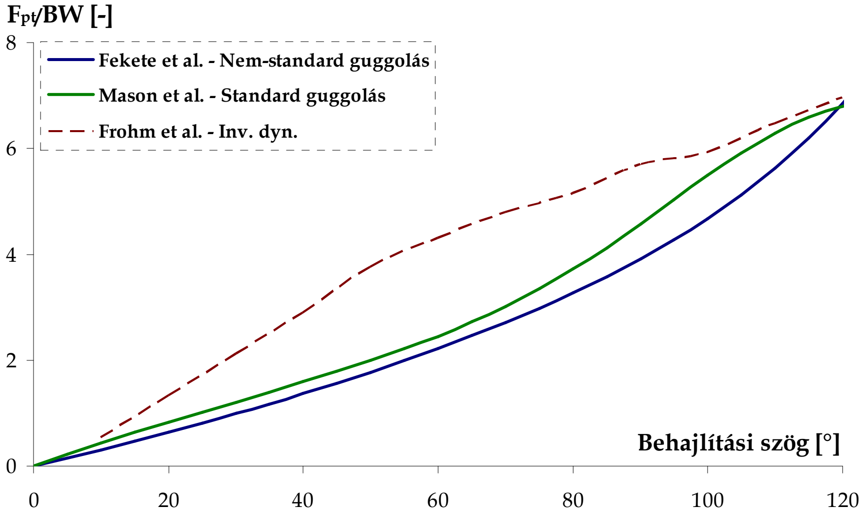


Figure 14: Patellar tendon force

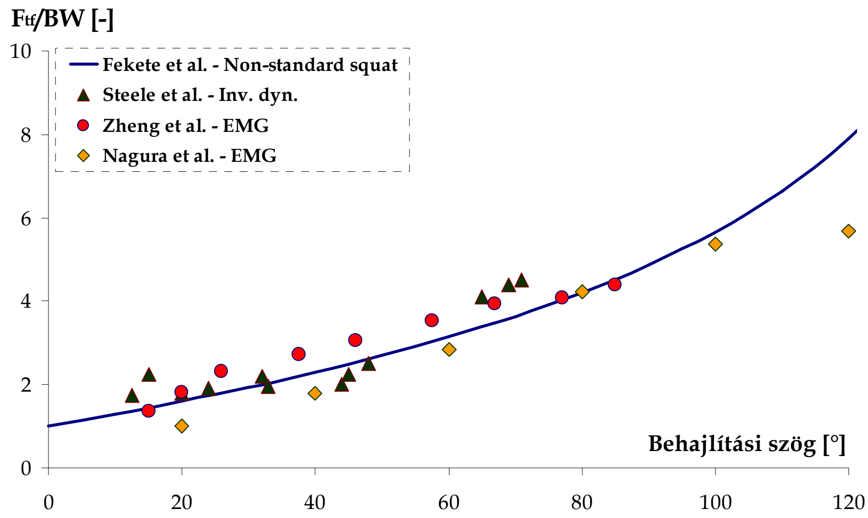


Figure 15: Tibiofemoral compression force

In Figure 12, the quadriceps tendon force of the non-standard squat model corresponds well with the result of Kulas et al. [Kulas et al., 2012], Essinger et al. [Essinger et al., 1989] and Zheng et al. [Zheng et al., 1998]. Among the three authors, the most important comparison is considered with Kulas et al. [Kulas et al., 2012], since their study involves the effect of moderate forward movement of the trunk. The non-standard squat model and the model of Sharma et al. [Sharma et al., 2008] estimate the peak force at 120˚ of flexion angle, while the model of Essinger et al. [Essinger et al., 1989] approaches the peak at 100˚ of flexion angle. The peak force of the non-standard squat model is estimated to 3.63 ***BW***.

In Figure 13, the patellar tendon force is plotted. The correlation is very strong between the standard and non-standard models regarding this force. Their characteristics, magnitudes and peak locations are in good accordance with each other. The experimental result of Frohm et al. [Frohm et al., 2007] shares more or less the same location and magnitude, but it has different, degressive, characteristic. According to these corresponding results, the estimated peak force is 6.8 ***BW*** and the peak location is at 120˚ of flexion angle.

In Figure 14, the patellofemoral compression force is plotted. The deviation between the forces is higher, compared to other forces ($\boldsymbol{F_q}$ or $\boldsymbol{F_{pt}}$). By considering the plotted results, the non-standard squat model correlates with the results of Sharma et al. [Sharma et al., 2008], Komistek et al. [Komistek et al., 2005] and Escamilla et al. [Escamilla et al., 2008], although with some overestimation. Komistek et al. [Komistek et al., 2005] and Escamilla et al. [Escamilla et al., 2008] estimated the peak force between 2.6 and 3.5 ***BW***. The estimated peak angle of the non-standard squat model, in this case, is located around 110˚ of flexion angle and the peak force is approximately 3.6 ***BW***. The only exception is the result of Escamilla et al. [Escamilla et al., 2008], which was only carried out up to a 90˚ of flexion angle.

In Figure 15, the tibiofemoral force is presented. The standard squat model by Mason et al. [Mason et al., 2008] is not able to predict this force, thus no comparison could be carried out between the two analytical models. The new analytical-kinetical model was compared to the results of Zheng et al. [Zheng et al., 1998], Nagura et al. [Nagura et al., 2010] and Steele et al. [Steele et al., 2012]. As it is seen, the four results have very good correlation with each other, although the experimental result of Zheng et al. [Zheng et al., 1998] and Steele et al. [Steele et al., 2012] provide prediction only until 90° and 70° of flexion angle. Here, the peak force is estimated between 7.8 ***BW***.

As a validation, the analytically obtained forces are compared to results derived by inverse dynamics approach, oxford-type test rigs, and other analytical models.

The inverse dynamics approach is based on the following method: if the acting force-system (or acting moments) and the moment of inertia (or mass) are known, then by double-integration the displacement of the body (or particle) can be deduced:

Forward dynamics

F → $F = m \cdot \ddot{x}$ → ∬ → x

On the other hand, if the moment of inertia (or mass) and the displacement are known, then similarly with a double derivation the acting force-system (or moments) can be deduced:

Inverse dynamics

x → d^2/dt^2 → $F = m \cdot \ddot{x}$ → F

With regard to human locomotion, the limbs are represented as rigid links, where given the kinematics of each part, the inverse dynamics approach determines the forces (and moments) responsible for the individual movements

Although, no direct measurement was performed to validate the obtained results, a comparison between the current predictions and the ones found in the literature can estimate the validity of this new analytical-kinetical model (Table 4.2). The comparison was done at 90° of flexion angle, since that was the angle until all sources had results.

AUTHOR	MODEL TYPE	F_{pf}/BW	F_{pt}/BW	F_{tf}/BW	F_q/BW
Mason et al., 2008	Hinge	5.4	4.5	-	7.1
Dahlkvist et al., 1982	Hinge	7.4	-	5.1	5.3
Steele et al., 2012	Hinge (OpenSim)	-	-	7.6	9.6
Essinger et al., 1989	Three-dimensional	-	-	-	4.7
Kulas et al., 2012	Inverse dynamics	-	-	-	4.1
Sharma et al., 2008	Inverse dynamics	2.7	1.5	-	3
Frohm et al., 2007	Inverse dynamics	-	5.7	-	-
Escamilla et al., 2008	Inverse dynamics	3.5	-	-	-
Komistek et al., 2005	Inverse dynamics	2.5	-	-	-
Nagura et al., 2006	EMG	-	-	4.7	4.5
Zheng et al., 1998	EMG	-	-	4.4	4.7
Churchill et al., 2001	Oxford	3.9	-	-	-
Mean		4.3	3.9	5.45	5.37
Standard deviation		1.86	2.16	1.46	2.06
Present model	**Hinge**	**3.51**	**3.9**	**4.86**	**3.52**

Table 5: Peak muscle force predictions from literature and present model at 90˚ of flexion angle

According to Table 5, the present model shows very good correlation with the results from the literature. In spite of the simplicity of the model, the predicted forces, compared to the calculated mean values, only differed by 0-1.85 SD respectively.

By summarizing the findings, it can be concluded that the new analytical-kinetical model is able to predict correctly the patellofemoral and tibiofemoral forces under standard and non-standard squat movement. The comparison of the results with inverse dynaics methods shows convincing accuracy, while the analytical model has also the advantage that by simple algebraic equations the forces can be calculated.

5.2 Results regarding the numerical-kinematical model

The new numerical-kinematical model was primarily used to determine the sliding-rolling pattern between the connecting condyles of prosthesis geometries. One the on hand, the obtained results confirmed the study of Wilson et al. [Wilson et al., 1998], who stated that sliding-rolling ratio is slightly (5-8%) higher on the medial side than the lateral side, thus these medial results were taken as reference functions (Figure 16).

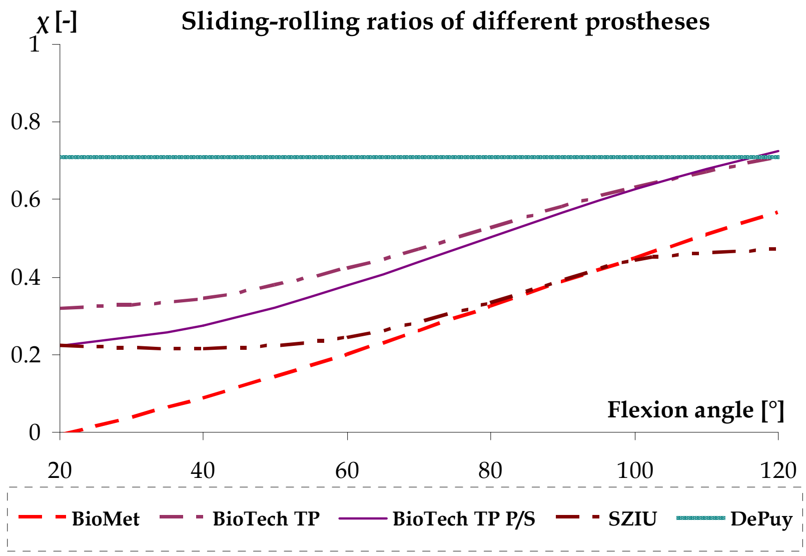


Figure 16: Sliding-rolling functions

From Figure 16, a well-visible trend appears along the flexion angle for the SZIU, Biotech TP, Biotech TP P/S and the BioMet Oxford models. The DePuy model although falls completely out of the range, as appears to be a constant function, thus it has been removed from the further investigation.

To generalize the results, the obtained functions have been averaged and the average function has been plotted in Figure 17 with the standard deviation.

$$\chi(\alpha) = -5.16\cdot 10^{-7}\cdot\alpha^3 + 1.235\cdot 10^{-4}\cdot\alpha^2 - 4.113\cdot 10^{-3}\cdot\alpha + 0.226 \tag{4.3}$$

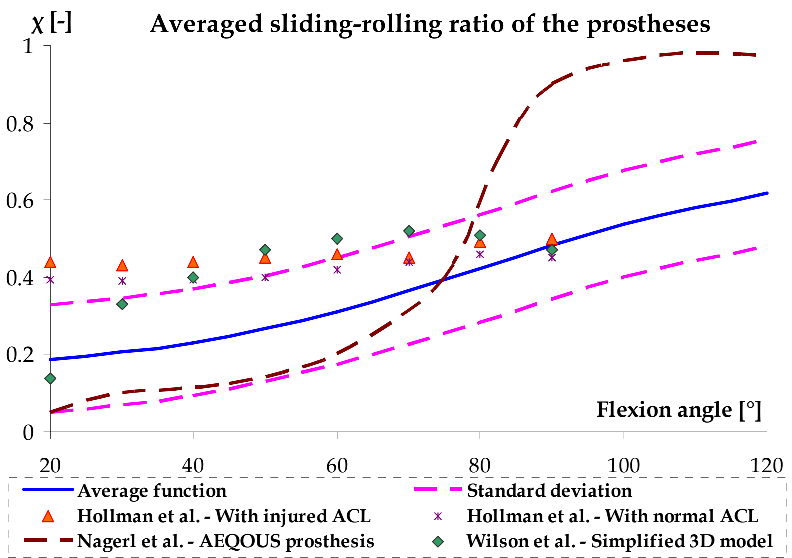


Figure 17: Averaged sliding-rolling function

During the comparison of the results, a mention must be made reagarding the methods: Hollman et al. [Hollman et al., 2002] used the path of instantaneous center of rotation (PICR) method which is a simplified two-dimensional approach that corresponds well with the quazi three-dimensional result of Wilson et al. [Wilson et al., 1998].

Both of the authors agreed that their main limitation is the geometry, which might cause that the sliding-rolling ratio is underestimated in the higher flexion angles.

In contrary, Nägerl et al. [Nägerl et al., 2008] used unique prosthesis geometry (AEQUOS-G1), which was designed to maintain primarily rolling attributes during the stance phase in order to avoid wear due to the sliding friction. Their result corresponds well in the lower region, although they assume that the sliding-rolling ratio reaches its maximum already at 90˚ of flexion angle. A mention must be made: the result of Nägerl et al. [Nägerl et al., 2008] represents a single prosthesis, while the new obtained results, determining a well-defined area, shows the trend of five commercial prostheses and provides a more general view about the phenomenon.

6 NEW SCIENTIFIC THESES

In this doctoral thesis, the kinetics (term of classical mechanics, that concerns the cause of the motion generated by forces and moments) and local kinematics (term of classical mechanics, that studies the movement of particles or bodies without considering the causes of the motion) of the standard and non-standard squat has been comprehensively studied.

Regarding the kinetics of the human knee joint, my aim was to demonstrate how the horizontal movement of the center of gravity – as a new parameter in the theory of squatting – influences its kinetics.

As for the kinematical point of view, the sliding-rolling ratio has been investigated on the complete function arc of different prostheses regarding both lateral and medial condyles.

These aims were accomplished by creating a new analytical-kinetical model that can estimate the patellofemoral and tibiofemoral forces under standard or non-standard squatting, and a numerical-mechanical model that is suitable to investigate the local kinematics of the knee, more precisely the sliding-rolling phenomenon.

The analytical-kinetical model includes new parameters which were experimentally determined by involving 16 human subjects into the research. By the use of this new analytical-kinetical model, closed-form solution could be derived to estimate the patellofemoral and tibiofemoral forces as a function of flexion angle. The obtained results correspond well with inverse dynamics results from the relevant literature. The model – beside its accuracy and simplicity – has the considerable advantage that no measuring device or other instrument is needed to calculate patellofemoral or tibiofemoral forces during standard or non-standard squatting.

In the numerical part of the thesis, the relative motion between the contact surfaces of the knee joint has been comprehensively studied by means of multibody dynamics approach. The observed motion was defined with a new sliding-rolling ratio, which has a significant roll among the wear test parameters regarding to total knee replacements. Earlier, this ratio was only known in the initial part of the movement, approximately up to 30° of flexion angle.

The numerical modelling and simulation were carried out in the MSC.ADAMS program, involving a number of commercially used prostheses from different manufacturers.
The new scientific results have been summarized in three theses.

1st Thesis: A new analytical-kinetical model has been created that can provide a closed-form solution regarding the patellofemoral and tibiofemoral forces by taking the horizontal movement of the center of gravity – as a new parameter in the squat literature – into account. It has also been proven by the analytical-kinetical model that this new parameter has significant effect on the patellofemoral kinetics.

By taking into consideration the earlier published knee models, a new analytical-kinetical model has been created which involves 7 anthropometrical parameters in order to describe the evolution of the patellofemoral and tibiofemoral forces between 0° and 120° of flexion angle. The model can calculate the forces with respect to standard and non-standard squat.

$$\frac{F_{pt}(\alpha)}{G} = \frac{\lambda_1(\alpha) \cdot \sin\gamma(\alpha)}{\lambda_p \cdot \sin\beta(\alpha) + \lambda_t \cdot \cos\beta(\alpha)} \qquad (3.3)$$

$$\frac{F_{tf}(\alpha)}{G} = \frac{F_{pt}}{G} \cdot \frac{\cos\beta(\alpha)}{\cos\varphi(\alpha)} + \frac{\cos\gamma(\alpha)}{\cos\varphi(\alpha)} \qquad (3.7)$$

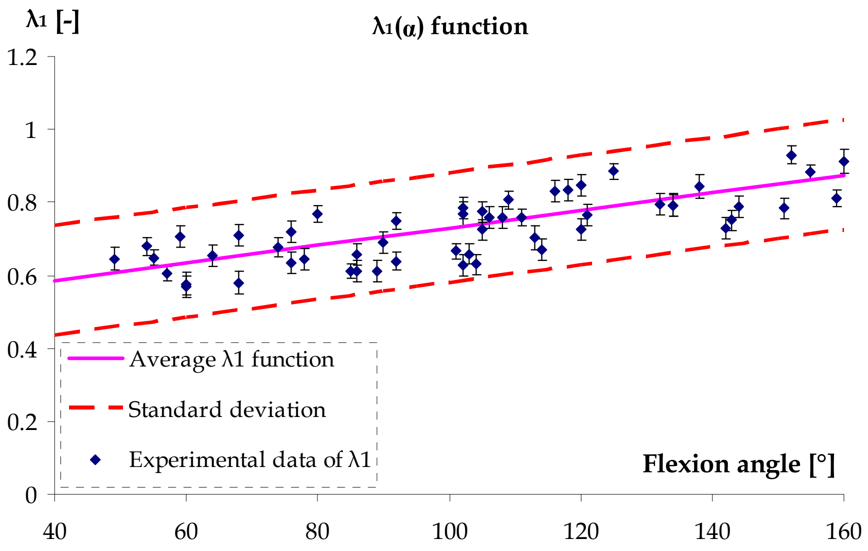
$$\frac{F_q(\alpha)}{G} = \frac{\lambda_3(\alpha) \cdot \sin(\alpha - \gamma(\alpha))}{\lambda_f} \qquad (3.9)$$

$$\frac{F_{pf}(\alpha)}{G} = \frac{\sqrt{F_q(\alpha)^2 + F_{pt}(\alpha)^2 - 2 \cdot F_q(\alpha) \cdot F_{pt}(\alpha) \cdot \cos(\beta(\alpha) + \delta(\alpha) + \gamma(\alpha))}}{G} \qquad (3.12)$$

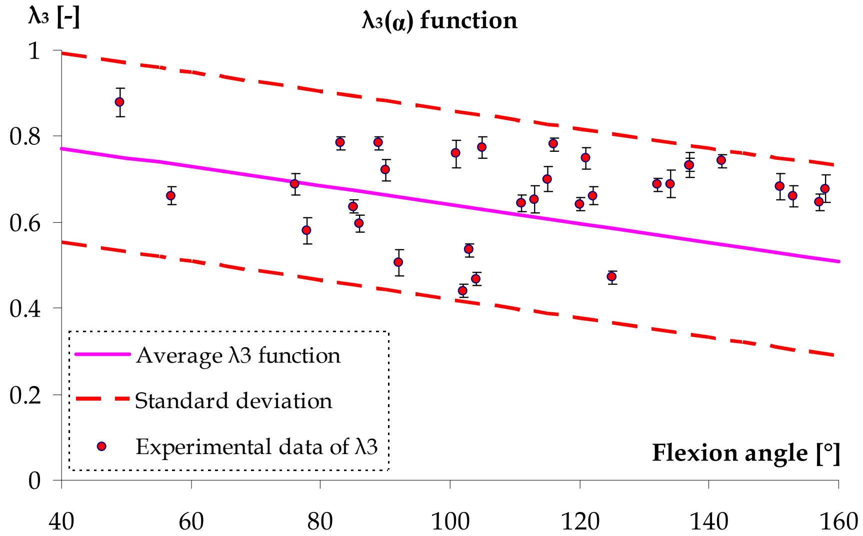
Applicabilty limit of the model: $0^\circ \leq \alpha \leq 120^\circ$

2nd Thesis: By means of experimental methods the horizontal movement of the center of gravity during non-standard squat has been experimentally described as a function of flexion angle

As a parameter in demand for the analytical-kinetical model, the center of gravity functions were determined by experimental methods carried out on 16 human subjects, under non-standard squatting motion. The human subjects had to carry out the movement under certain conditions (stretched out hands, adjusted heels, holding the position for 3 second), thus the functions describe one certain squatting motion.



$$\lambda_1(\alpha) = 0.0024 \cdot \alpha + 0.4925 \pm t \cdot 0.15 \tag{3.28}$$



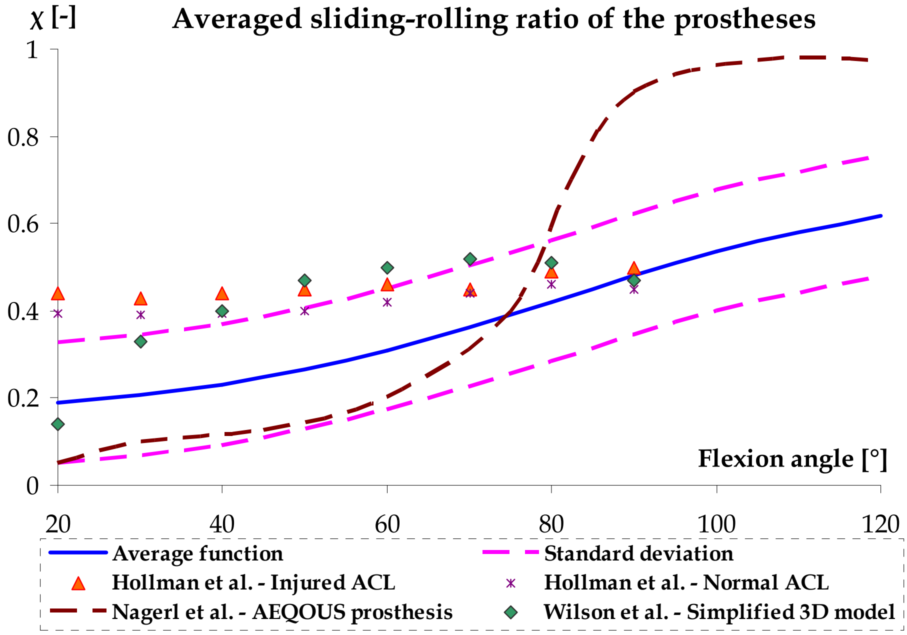
$$\lambda_3(\alpha) = -0.0022 \cdot \alpha + 0.86 \pm t \cdot 0.22 \tag{3.29}$$

Applicabilty limit of the model: $40^\circ \leq \alpha \leq 160^\circ$

3rd Thesis: Based on Multibody approach, the sliding-rolling ratio between the contact surfaces has been numerically determined along the complete functional arc with regard to actual prosthesis geometries.

The sliding-rolling ratio (with its maximum and minimum values) on both lateral and medial side has been determined by the use of commercial prosthesis models. Earlier, the ratio was only known in the initial movement (0˚ ≤ α ≤ 20-30˚) thus now the phenomenon – and its evolution – has been discovered, under certain circumstances, along the complete functional arc of the knee joint.

$$\chi(\alpha) = -5.16 \cdot 10^{-7} \cdot \alpha^3 + 1.235 \cdot 10^{-4} \cdot \alpha^2 - 4.113 \cdot 10^{-3} \cdot \alpha + 0.226 \qquad (4.3)$$



Applicability limit of the model: 20˚ ≤ α ≤ 120˚

7 PUBLICATIONS RELATED TO THE THESIS

Peer-reviewed journals publications included in SCI:

1. **G. Fekete**, B. Csizmadia, M. A. Wahab, P. De Baets: "*Experimental determination of horizontal motion of human center of gravity during squatting*", Experimental Techniques, Accepted, 2011, DOI: 10.1111/j.1747-1567.2011.00768.x. IF: 0.505
2. **G. Fekete**, B. M. Csizmadia, M. A. Wahab, P. De Baets, G. Katona, L. V. Vanegas-Useche, J. A. Solanilla: *"Sliding-rolling ratio during deep squat with regard to different knee prostheses",* Acta Polytechnica Hungarica, 9 (5), 5-24, 2012. IF: 0.385.
3. **G. Fekete**, B. M. Csizmadia, M.A. Wahab, P. De Baets, I. Bíró: *"Effect of the horizontal movement of the center of gravity on the patellofemoral biomechanics",* Dyna Colombia, Under review, 2013.

Peer-reviewed journal publications

1. **G. Fekete**, B. Csizmadia, P. De Baets, M. A. Wahab: *"Review of current knee biomechanical modelling techniques",* Mechanical Engineering Letters, 5, 30-36, 2011.
2. **G. Fekete**, B. Csizmadia: *"Biomechanics of the human knee joint",* Mechanical Engineering Letters, 1, 146-158, 2008.
3. **G. Fekete**, B. Csizmadia: *"Csúszva gördülés értelmezése a térdízület biomechanikai vizsgálatához",* Gép, 12(59), 4-8, 2008.
4. **G. Fekete**, B. Csizmadia: *"Interpretation of sliding-roll phenomena in the examination of knee biomechanics",* Bulletin of Szent István University, 339-347, 2008.
5. **G. Fekete**, B. Csizmadia: *"Computational human knee joint model for determining sliding-rolling properties",* Scientific Bulletin of Politehnica University Timisoara – Transaction on Mechanics, 53 (67), 305-309, 2008.

Conference Proceedings

1. **G. Fekete**, B. Csizmadia, M.A. Wahab, P. De Baets: *"Analytical patellofemoral knee models: Past and Present",* Synergy in the technical development of agriculture and food industry, Gödöllő, Hungary, October 9-16, 2011.
2. **G. Fekete**, B. Csizmadia, M.A. Wahab, P. De Baets: *"Analytical and computational estimation of patellofemoral forces in the knee under squatting and isometric motion",* Sustainable Construction and Design, 2, 246-257, Gent, Belgium, February 16-17, 2011.
3. **G. Fekete**, B. Csizmadia: *"Biomechanical research of Szent István University",* Sustainable Construction and Design, 1, 107-114, Gent, Belgium, February 10, 2010.

4. **G. Fekete**, B. Csizmadia: *“Numerical methods for determining local motions of human knee joint”,* Zilele Technice Studentesti, 12, 204-210. Timisoara, Romania, May 11-18, 2008.
5. **G. Fekete**, L. Kátai: *“MSC.ADAMS programrendszer felhasználása a biomechanikai modellezésben”,* Fiatal Műszakiak Tudományos Ülésszaka, 13, 1-4. Cluj-Napoca, Romania, March 13-14, 2008.

Scientific student theses:

1. **G**. **Fekete**: „*Experimental methods for determining of mechanical model of human knee*”. Zilele Technice Studenţeşti (Műszaki Hallgatói Napok), Timisoara, Romania, 2007.
   1st place.
2. **G. Fekete**: „*Kísérleti módszerek a térdízület mechanikai modelljének számításához”*. OTDK dolgozat, Győr, Hungary, 2007. **3rd place.**
3. **G. Fekete**: „*Kísérleti módszerek a térdízület mechanikai modelljének számításához”*. TDK dolgozat, Gödöllő, Hungary, 2005. **2nd place.**
4. **G. Fekete**, M. Kassai: „*Térd egyszerű kinetikai modellje”*. TDK dolgozat, Gödöllő, Hungary, 2004. **Special prize.**

Peer-reviewer activity in SCI journals:

1. Experimental Techniques (1)
2. Clinical Biomechanics (1)