BIOMECHANICAL RESEARCH OF SZENT ISTVÁN UNIVERSITY

B. Csizmadia[1], G. Fekete[1]

[1] Szent István University, Institute of Mechanics and Machinery, Hungary

Abstract This paper represents a short summary of the human knee joint modeling. The Biomechanical Team of the Szent István University investigates the motion of the human knee in the case of squatting. In the analysis of the previously mentioned motion the emphasis is laid on the kinematical properties, which is approximated by experimental and numerical ways. Primarily the sliding and rolling properties of the human condyles are examined, especially I in those domains where pure rolling, rolling and sliding jointly, and pure sliding may occur. Since only the two extremities are well known in the theoretical and applied mechanics, this paper wishes to furnish further information to the subject. On the other hand, in order to determine special additional features the global experimental investigations of the knee are also crucial. These results can be essential for further investigations of the phenomena of the combination of rolling and sliding and to show a new path of creating prosthesis.

Keywords knee joint, experimental, computational model, kinematics

1 INTRODUCTION

The aim of the Biomechanics Team of Szent István University is to reach a deeper understanding of the human knee joint, by carrying out both experimental and theoretical investigations. Different tasks are being explored such as the determination of the axis of revolution by Bíró [1], the governing angles of motion of knee by Csizmadia [2], the global determination of the acting forces or the local motions by Fekete [3]. The team has carried out significant work in this area, and has already presented experimental results. The final aim of the project is to develop a new kind of prosthesis which surface and topography is designed according to the data of multiple cadaver experiments. The steps to successfully carry out this project are the followings:

- Creating a mechanical model to provide initial and boundary conditions,
- Carrying out experimental results in order to determine the acting forces and the governing angles in the knee joint and its surroundings. The results can be useful to verify and even to correct the mechanical model,
- Creating a computational model to ease the calculations and to take into consideration (if necessary) complex geometry. According to the large number of simulations the results can be extended to the level of creating an averaged condyle topology of the human knee.

Worthy of note, there are two governing hypothesizes in the research of the human knee. The first one attributes the importance of the human knee motion properties to the connecting surfaces of the bones (the contact points of the femur and tibia on the condyles), while the function of ligaments remains as a kind of control of the kinematical trajectory of the knee joint. The second one claims the opposite and hereby it must be mentioned that the SZIE Biomechanical Team follows the first hypothesis. Thus, beside many others, the precise mapping of the condyles geometry is an important issue in our research.

2 ANALYTICAL GLOBAL MECHANICAL MODEL

In the 1st phase, the SZIE Biomechanical Team started using both cadaver and intact living knees in their experimental researches. In order to carry out the experiments a special apparatus was needed which was proper to clutch a cadaver knee, and load the joint as well, in the case of squatting (*Figure 1.).*

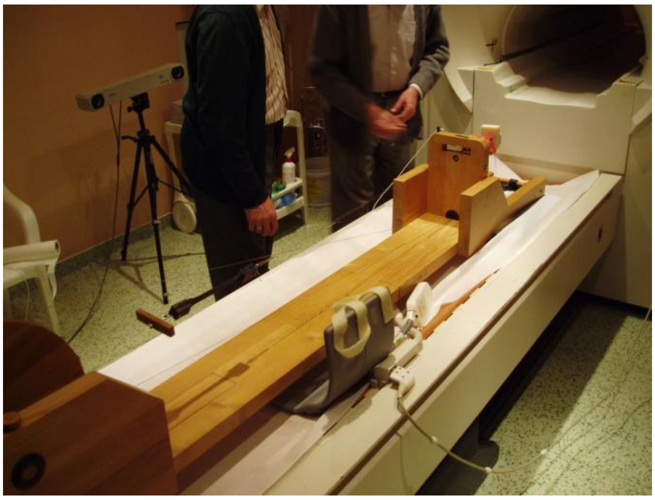

Figure 1.

During the load of the test rig, the governing angles are being recorded by the POLARIS system, which can simultaneously determine the tracks of many isolated points while they are in the move. Besides the test rig, the Team created a 2D, global mechanical model [4] to determine the forces in the muscles of the human knee joint (*Figure 2. and Figure 3.)*.

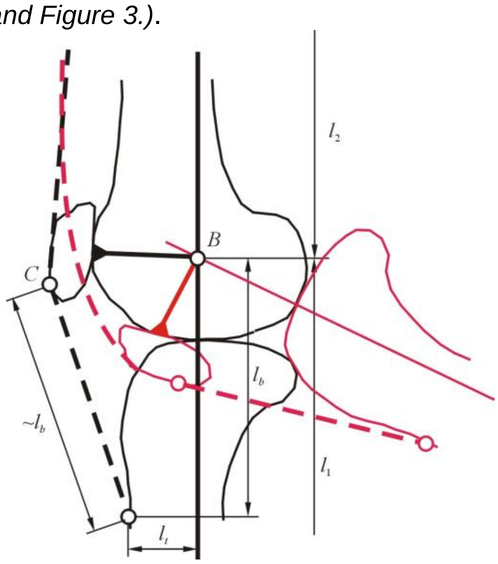

Figure 2.

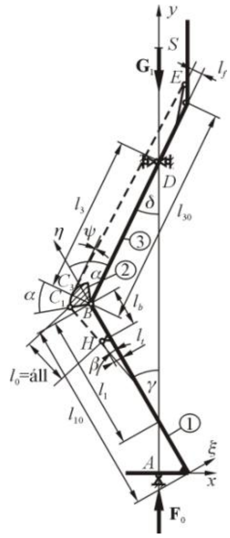

Figure 3.

The model provides the evaluation of the forces of the connecting ligaments in the function of an arbitrarily α angle, in the ratio of the load desribed by Csizmadia [4]. The main problem of the model was the lack of initial and boundary conditions which are represented as missing functions and constants. These are the following parameters:

- λ_1 (l_1/l_{10}): Active proportional length of tibia.
- λ_3 (l_3/l_{30}): Active proportional length of femur.
- λ_b (l_b/l_{10}): Proportional length of ligmentum patellae.
- λ_f (l_f/l_{30}): Proportional thickness of thigh.
- λ_t (l_t/l_{10}): Proportional thickness of shin.
- β: The angle of the ligamentum patellae compared to the tibia.
- ψ (γ/α): The proportional angle of the γ.

Among the parameters, the λ_1, λ_3, ψ, β are the functions of α, while the λ_t, λ_f, λ_b are proportional constants.

3 EXPERIMENTAL METHODS, RESULTS, AND VALIDATION

3.1 Method

In order to obtain these parameters, experiments were carried out on fourteen human persons. The experimental persons were between in the age of 21 to 27, nine male, and five female in normal physical condition. The aim of the experiment was to measure the center of gravity, and to construct the so-called bone-axes (namely the femur and tibia) so the demanded proportional values could be measured as well.

For the experiment, a SPIDER-8 data acquisition system was used, controlled by the Catman Express 3.0 program. Three KALIBER type dynamometers were connected to the SPIDER-8. Before the measurement, certain boundary conditions were given to the experimental persons, such as:

- stretched arms
- straight spine
- heel adapted to the apparatus
- 3 second steady-state in each experimental setting.

The experiments were carried out by having each person stand on to the board shown on the *Figure 4*,

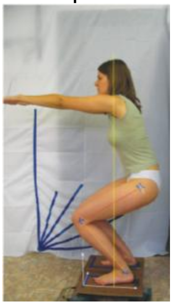

Figure 4.

while the 'x' and 'y' components of center of gravity were measured in seven static positions. The average value and its standard deviation were also determined with the statistical error, calculated by the Student criteria**.** Since the prospective values were available, the bone-axes could be constructed. Markers were put on to the end of the bones, so if they are connected with a line, the theoretical axes will be given in each status. This method was correct in the first status of the squatting, so the axes were constructed, but in the other statuses, the connective tissue was shifted, so the marker did not represent the real position of the end of the bone. The solution of the problem was given by the functional anatomy. In case of squatting, the posterior muscles of the thigh are in a totally laxed condition, while the same statement is valid to the front muscles of the shin [5]. This fact can be expended as follows: during the motion of squatting the muscles do not act, or the action is not significant in the above mentioned accentuated areas thus the gradient of the volume is zero. So these areas can be modeled as rigid bodies. In the view of this conclusion, two more auxiliary vertices can be taken, and now, with the use of basic construction methods, the demanded vertices can be easily allocated.

3.2 Results

When the axes of the bones are constructed in each status, the prospective values of the center of gravity can be plotted as a line, and by the intersections, the unknown values can be appointed. Due to the applied methods, the following results can be reported. All of the constant proportional values have been estimated with their deviation.

$\lambda_t = 0{,}10 \pm 0{,}037$ $\quad\quad$ $\lambda_b = 0{,}158 \pm 0{,}087$ $\quad\quad$ $\lambda_f = 0{,}164 \pm 0{,}057$

In addition, all the proportional functions are given as well [6]. In the case of β function, another result is put in the diagram, given by the Marcus-Wimmer-Adriacchi formula [7]. Unequivocally, it is the segment of the determined domain. Eventually, the results in diagrams are the followings (*Figure 5.).*

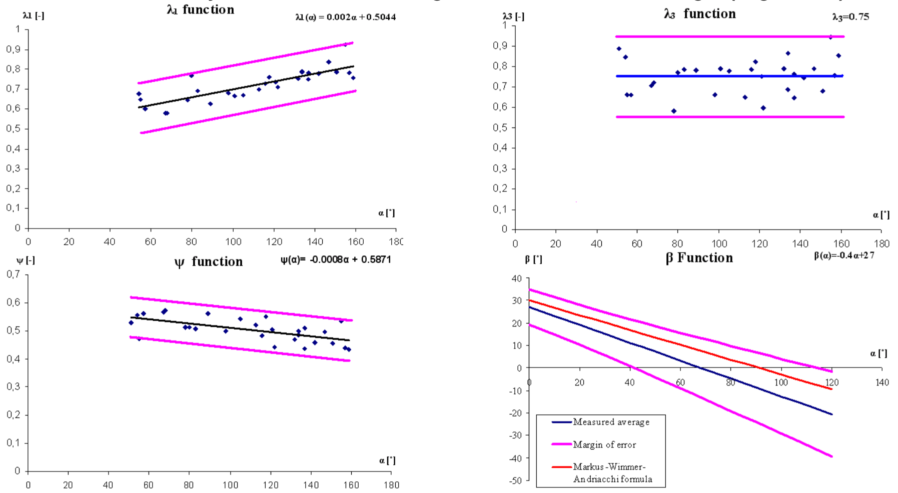

Figure 5.

With the introduction of these constants and functions, the mechanical model turned to be ready to execute. The solution of the proportional quadriceps force function with the average values of the proportional parameters during flexion is shown on the *Figure 6*.

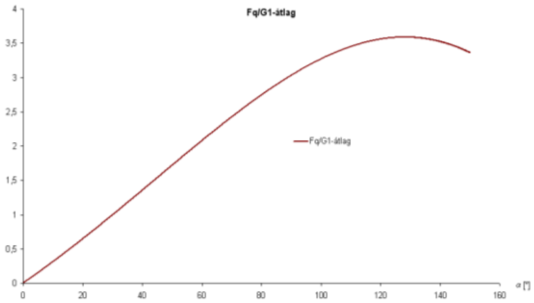

Figure 6.

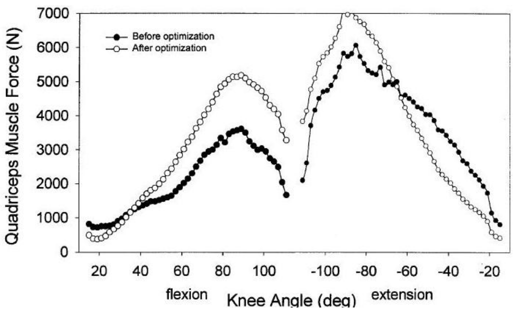

Figure 7.

Approximately, the function is linear to 80˚, and besides the theoretical result of the SZIE mechanical model, an experimental result is shown carried out by Zheng [8] **(***Figure 7.***)**. Apparently, the character of the model and the experimental result is very similar in the case of flexion.

3.3 Comparison of results

Since results derived by pure theoretic models can show great differences between model and reality, in association with the Ghent University verification tests were done on a totally different test rig. The Ghent University set-up is capable to measure the force in the quadriceps, but the proportional parameters can not be adjusted, so the given results are not an entirely appropriate model of the human squatting.
The *Figure 8.* represents the result of the artificial knee according to the mechanical model, which is only segment to 55˚.

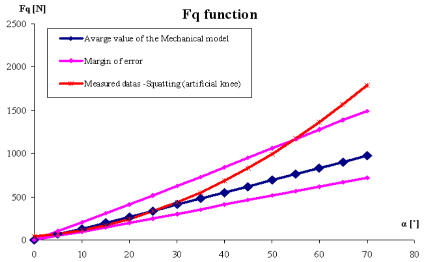

Figure 8.

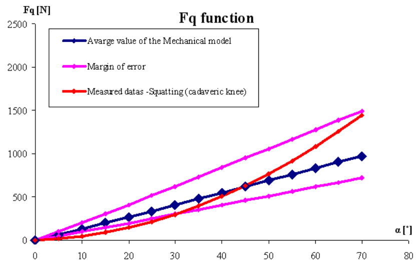

Figure 9.

With the cadaver knee, the given results are better; the segment goes to 70˚ (*Figure 9.*). This way the demanded parameters have been estimated with their statistical error, and in the case of the β function, compared with the reference formula. Using the obtained parameters, the solution of the quadriceps force function correlates with the experimental results of the reference. For further use, the functions of the mechanical model can be applied in the computation knee joint model as more realistic and validated boundary conditions, according to the former 'constant loaded' conditions [9, 10]. The applicability of the test rigs is also determined with these calculations and experiments.

4 EXPERIMENTAL METHODS, RESULTS, AND VALIDATION

4.1 Internal motions and their background

Since different results about the human knee joint are available, the internal motions and special addition features such as the combination of sliding and rolling must be examined. Two kinds of motions occur inside of the human knee joint during action, which are namely sliding and rolling. Their gradient and proportion is currently unknown. Zuppinger was the first among the medical researchers who claimed according to X-ray images that the relation between the tibia and femur was basically rolling until 20° of flexion and beyond that point, sliding and rolling jointly occurred. Making a decision about the proportion of movements and the behavior of the phenomena by using only X-ray images is simply impossible. This is the reason why the other aim of the team is to found a new and applicable method to correctly investigate and determined the phenomena of *sliding-roll*. To do so, a special proportional value will be introduced.

4.2 Model creation

In order to correct these hypothesizes and to make further steps in the research the MSC.ADAMS program system was used to carry out computational examination and to build a complex knee model. MR images were taken and a model was mapped from them. With basic CAD methods a geometric model was generated and imported into the ADAMS (*Figure 10.).*

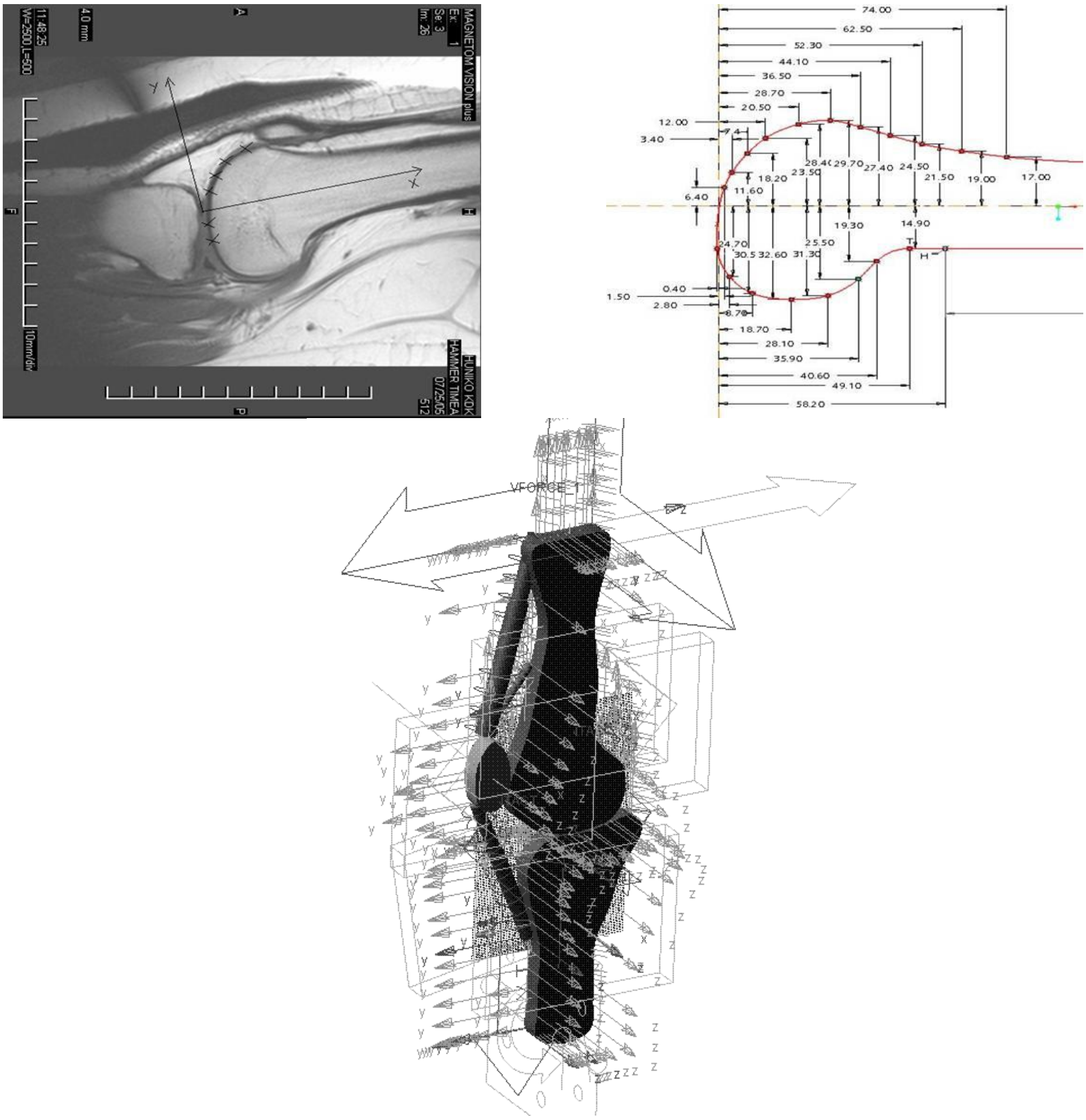

Figure 10. The geometrical model

The following simplifications were applied:

- 2.5D model (it involves a third finite extent)
- Bones are considered to be rigid bodies
- The damping and mechanical properties of the meniscuses are neglected but their lubricating behavior is taken into consideration

In the interest of to carry out squatting motion, kinematical constraints must be set to the model elements. The elements such as the femur, tibia and patella must be fixed to each other and to the environment. The following constraints and boundary conditions were applied in the program:

- Tibia, femur and patella can only perform planar motion (PLANAR JOINT),
- The top of the femur is attached to the hip-bone, thus deflection in the ‘x’ direction is not possible (MOTION JOINT),
- Bodies are attached to each other by springs (SPRING), therefore the muscle model is introduced,
- The tibia is attached to the environment by a hinge, thus the rotation around the ‘z’ axis is permitted (HINGE),
- The static friction coefficient is 0.11, while according to Wang [8] the dynamic friction coefficient is 0.008,
- One force in the ‘y’ direction is acting on the femur with a magnitude of 60 N.

4.3 Calculation method

Let a new specific value be introduced, which shall be titled as *sliding-rolling coefficient*, and denoted with χ. Let χ defined as:

$$\chi = \frac{\Delta s_{2i} - \Delta s_{3i}}{\Delta s_{2i}} \qquad (1)$$

- $\Delta s_{2i} = s_{2i} - s_{2K}$: Difference of the i[th] arc length extracted the initiative K arc length of the 2[nd] body,
- $\Delta s_{3i} = s_{3i} - s_{3K}$: Difference of the i[th] arc length extracted the initiative K arc length of the 3[rd] body.

The above-mentioned quantity shows the proportion of the connecting arc lengths in percentages, thus exact conclusions can be drawn to the sliding and rolling feature of the motion. If the value of χ equals to 1, then it is pure sliding, and if it equals to 0, then it is pure rolling.
To carry out this calculation, the arc lengths must be determined. The following kinematic quantities can be calculated by the MSC.ADAMS during the simulation of the motion:

- $\bar{r}_K(t)$ vector-scalar function, which determines the instantaneous position of the connecting points of the two bodies (*Figure 11*.),

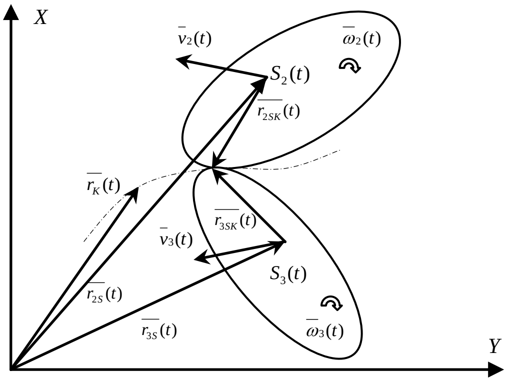

Figure 11.

- $\bar{r}_{2S}(t)$, $\bar{r}_{3S}(t)$, $\bar{v}_{2S}(t)$, $\bar{v}_{3S}(t)$, $\bar{\omega}_2(t)$, $\bar{\omega}_3(t)$ vector-scalar functions, which sequentially determine the instantaneous position of the center of gravity, velocity and angular velocity of the 2[nd] and 3[rd] bodies (*Figure 12*.).

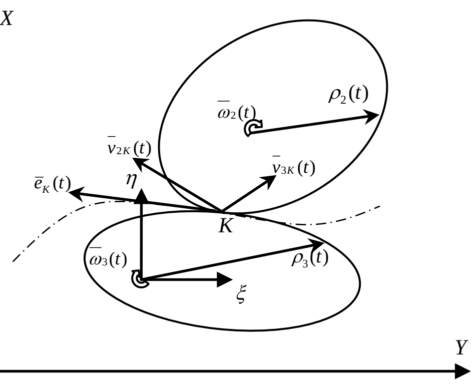

Figure 12.

In addition the following kinematic quantities can be directly estimated:

- Instantaneous values of F_f and F_N, which are the normal and frictional components of the connecting surfaces,

In order to examine the proportion of sliding and rolling, let us consider the following well-known formula from the results given by the MSC.ADAMS:

$$d\bar{r} = \bar{r}_{2SK}(t) = \bar{r}_K(t) - \bar{r}_{2S}(t) \qquad (2)$$

Substituting the results of (1) equation into the (2) and (3):

$$\bar{v}_{2K} = \bar{v}_{2S} + \bar{\omega}_2 \times \bar{r}_{2SK}(t) \qquad (3)$$

$$\bar{v}_{3K} = \bar{v}_{3S} + \bar{\omega}_3 \times \bar{r}_{3SK}(t) \qquad (4)$$

Thus the instantaneous velocity of the connecting points of the bodies can be calculated in the $x - y$ coordinate system. If the instantaneous tangent unit vector of the connecting curvature can be determined:

$$\bar{e}_K(t) = \frac{\dot{\bar{r}}_K(t)}{\left|\dot{\bar{r}}_K(t)\right|} \quad (5)$$

Then by multiplying the velocity vectors with it, the sliding components of the velocities become quantifiable:

$$v_{2N} = \bar{v}_{2K} \cdot \bar{e}_K(t) \text{, and } v_{3N} = \bar{v}_{3K} \cdot \bar{e}_K(t) \quad (6)$$

By these results the sliding components can be calculated:

$$\Delta v_N = v_{3N} - v_{2N} \quad (7)$$

After the execution of the ADAMS program, the following numerical results were produced *(Figure 13.).* The proportion of the Ff/Fn function *(Figure 14.)* can be directly plotted.

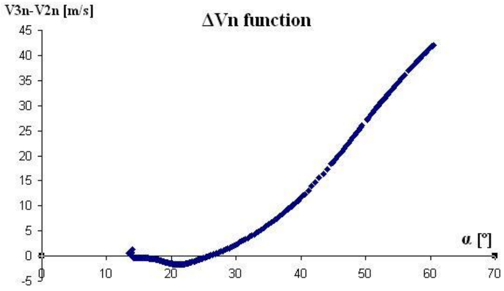

Figure 13.

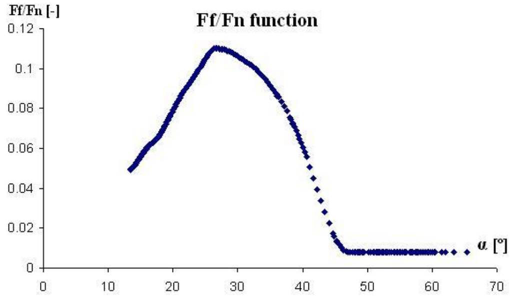

Figure 14.

Figure 13. correctly demonstrates that the sliding velocity components turn to be zero at approximately 26º, and by examining Figure 14. it shows the peak of the *Ff/Fn* function is also around 26º. That is the threshold where the first sliding occurs, and from that point until cca. 45º, the actions of sliding and rolling are combined. After 45º, the motion is characterized as pure sliding. Unfortunately, the integration of the sliding velocity component is not yet finished, but since the algorithm is already elaborated, the results will be soon presented.

5 FURTHER RESULTS AND APPLICATIONS

The above presented method provides a new way to investigate the phenomena of sliding and rolling in the case of planar motion. The method will be expanded to 3D investigations in the case of the sliding and rolling proportions of the condyles and prothesis. The goal with this method is partly to find the dominant part of the 3D surface, which takes on the majority portion during motion, and partly to investigate these quantities on several human subjects. In this way, by averaging the given results, hopefully a general conclusion will be drawn about the possible domains of the surface where the upcoming wear would be demolishing. In this part of the project, statistical methods will be applied on the set of data in order to determine the sliding-rolling proportions of the human knee. On the other hand, after creating a fairly accurate model, the idea is that according to the measured quantities of angles, the computational model would be adjusted to these statistical results. This can be carried out by an iterative modification of the geometry until the point where the numeric result correctly approximates the result measured by the POLARIS.

REFERENCES

[1] I. Bíró – B. M. Csizmadia – G. Katona: Determination of instantaneous axis of rotation of tibia and its role in the kinematical investigation of human knee joint, Proceedings of the Third Hungarian Conference on Biomechanics, Budapest, July 4-5, 2008, p. 55-62, ISBN 978 963 06 4307 8

[2] B. M. Csizmadia – G. Katona: Some result on the motion analysis executed on the experimental model of the human knee. 23rd Danubia Adria Symposium on Experimental Methods in Solid Mechanics – Extended abstracts, Podbanské – Zilina, Slovak Republic, 2006.11.26-27.

[3] G. Fekete: Numerical methods for determining local motions of human knee joint, Zilele Tehnice Studentesti - Editia a XII-A, Timisoara, 12-18 Mai (In press), 2008.

[4] B. M. Csizmadia – I. Bíró – G. Katona – Z. Szakál: A térdizület számítógépes megjelenítésén alapuló stereotaxisos navigációs műtéti eljárások (Stereo-taxis navigational methods of surgery, based on computational monitoring of human knee joint). NKFP/1B/0009/2002 pályázat keretében készült kutatási jelentés.

[5] Szentághotai János: Funkcionális Anatómia (Functional anatomy). Medicina Kiadó, Budapest – 1975.

[6] G. Fekete: Experimental methods for determining of mechanical model of human knee, Zilele Tehnice Studentesti - Editia a XI-A,Timisoara, 6-13 Mai, p. 1-8, ISSN 1843 - 1917, 2007.

[7] A. Marcus, Wimmer, T.P. Adriacchi: Tractive forces during rolling motion of the knee: Implications of wear in total knee replacement. Journal of Biomechanics 30 (1997), 131-137.

[8] N. Zheng, G.S. Fleisig, R.F. Escamilla, S.Barrentine: Analytical model of the knee for estimation of internal forces during exercise. Journal of Biomechanics 31 (1998), 963-967.

[9] J.Heegaard, P.F. Leyvraz, A. Curnier, L. Rakotomanana, R. Huiskes: The biomechanics of the human patella during passive knee flexion. Journal of Biomechanics 28 (1995), 1265-1279.

[10] W. Mesfar, A. Shirazi-Adl: Biomechanics of the knee joint in flexion under various quadriceps forces. The Knee 12 (2005), 424-434.