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Abstract

PhD THESIS

RESEARCH ON THE BIOMECHANICAL STUDY OF THE HUMAN KNEE JOINT WITH APPLICATIONS IN PROSTHETICS

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CONTENT

Cap.1. INTRODUCTION, STRUCTURE AND OBJECTIVES OF PhD THESIS

The aim and objectives of this PhD thesis have as a starting point the practical necessity of developing an optimized knee prosthesis model, referring in particular to the endoprosthesis of the knee, starting from the data gathered from the literature that highlight, on the one hand, a statistical increase in the incidence of articular knee joint pathology and, on the other hand, a number of shortcomings found in medical practice of existing knee prosthesis models clinically translated by a higher rate of complications arising from this surgical procedure. Worldwide research is being carried out to improve the design and manufacture of knee endoprostheses, a research developed in several directions: in terms of constituent materials, realization and design, and how to fix the implant. On the other hand, from the analysis of the literature at national level there is a still limited concern regarding the area of knee joint endoprosthesis.

The objectives of this PhD thesis are the biomechanical and human knee joint prosthesis study to reduce the medical consequences that the deficiencies or constructive limitations of existing knee prosthesis models have on the recovery of human knee joint functions in the idea of reproducing or improving its functionality. Starting from this general objective, this paper has the following specific objectives:

- performing experimental, kinematic and kinetostatic biomechanical evaluations on a sample of healthy subjects and on a sample of patients with osteoarthritic knees, flexion-extension movements of the human knee joint in order to obtain characteristic kinematic parameters and their comparison;
- the study of human knee joint endoprosthesis, existing knee endoprosthetic types currently used for human lower limb joint rehabilitation;
- elaboration of the complex parameterized virtual model of the human knee prosthetic joint assembly for its use for numerical simulations and for finite element analyses;
- developing the virtual model of a knee prosthesis commonly used in endoprosthesis for numerical simulations and finite element analyses of prosthetic joint assembly for different angles of inclination in varus and for two cases of anteroposterior tibial inclination;
- developing the optimized parameterized optimized model of a new knee prosthesis model;
- numerical simulations and finite element analyses of virtual knee prosthesis models proposed and prosthetic knee joint assembly for different angles of inclination in varus and for three cases of anteroposterior tibial inclination;
- the use of rapid prototyping technology to obtain the prototype of the new knee prosthesis model for its use for experimental implantation on cadaver bones;
- determining the surgical technique of implanting the prototype of the new knee prosthesis model;

The finite element method will be used to model and simulate the prosthetic human knee joint assembly to study its behavior, to study the stresses and transfer of load between the implant and bone, as well as between prosthetic components, using 3D finite elements.

The consistency of the research activities proposed in this project with the global trends in the field is also highlighted by the specialized bibliography. Thus, research on the materials and design of implants / prostheses, the way of fixing them, their type-dimensional optimization, have been treated and presented in specialized papers, which confirms the actuality of the research topic proposed by this PhD thesis.

The research in this paper will be carried out through an interdisciplinary approach using classical and modern engineering methods, calculation and experimental methods.

Based on the presented objectives, the paper is structured in 7 chapters, followed by bibliography and annexes.

After a brief presentation of the aim and objectives of this PhD thesis in Chapter 1, Chapter 2 presents anatomy and biomechanical elements of the human knee joint, as well as the main types of knee endoprostheses used for diseases such as knee osteoarthritis.

Chapter 3 is dedicated to the experimental biomechanical evaluation of the normal, osteoarthritic and prosthetic human knee joint. This chapter presents the gathering and processing of data on the variation of flexion-extension angles of human joints from lower limbs and the variation of reaction forces with the ground while the subject performs various tests, such as walking on a set of force platforms, ascending and descending the stairs, as well as sitting-standing from the chair.

Chapter 4 presents the stages of developing the three-dimensional virtual models of the components of a mannequin used to study the human walking kinematics and dynamics in the ADAMS virtual simulation environment based on the data previously gathered. Research is focused on normal and rapid human walking, as well as ascending and descending the stairs.

Chapter 5 presents the virtual modeling of the classical prosthesis and prosthetic joint assemblies for the studied varus cases, which were used for the simulation by finite element method, thus obtaining the distribution maps and the maximum values of the von Mises stresses and displacements.

Chapter 6 is dedicated to the three-dimensional virtual modeling of a new optimized knee prosthesis in order to restore the flexion range to a level closest to that of the intact knee as well as to the studies and analyses of a set of knee joint assemblies with the new prosthesis, corresponding to several angles of inclination in varus, and several angles of tibial anteroposterior inclination. In this chapter, the prototype of the new prosthesis model obtained by rapid prototyping, as well as the operative technique for implanting the knee prosthesis into a joint formed from bones taken from the corpse, are presented. The viability of this new prosthetic model was confirmed by virtual and experimental simulations and tests.

The thesis ends with the presentation in Chapter 7 of how the results of the research carried out for this thesis have been valued, the personal contributions of the author, as well as the future research directions in the field approached.

Cap.2. ELEMENTS OF KNEE ARTROPLASTY

2.1. Elements of human knee anatomy

The human knee is one of the most complex human joints, by the number of components, the loads to which it is subjected, the complicated spatial geometry of the components, and the existence of multiple contacts between the different components.

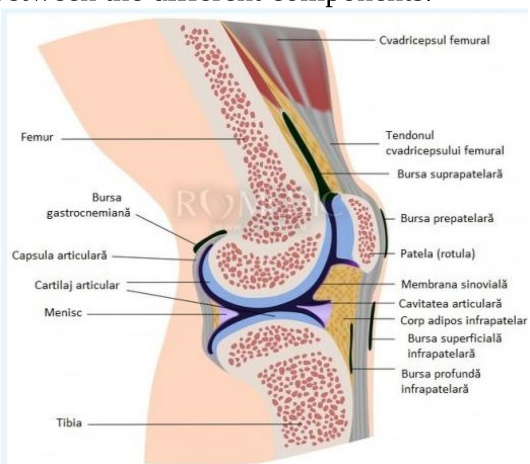


Fig. 2.3. Components of the knee [WAN]

2.3. Knee osteoarthritis (Gonarthrosis) [DEN_1977]

Articular osteoarthritis, one of the major chronic illnesses commonly found in middle-aged and elderly people, affects a very large number of people. This disease is accompanied by pain and can lead to constraints related to mobility, long-term disability and increased morbidity. The World Health Organization estimates that hundreds of millions of people already suffer from bone and joint disease, including osteoarthritis, with significant increases expected due to doubling the number of people over 50 by 2020. It is estimated that due to the drastic increase in osteoarthritis cases, by 2030, the total number of total hip arthroplasties will increase by 572,000 (approximately 174%), while total knee arthroplasties will increase by 3.48 million procedures (approximately 673% between 2005 and 2030 [KUR_2007] in the USA.

Gonarthrosis (knee osteoarthritis) is the most common form of arthritis, especially in the elderly people. Mostly called degenerative joint disease, it affects the cartilage, that is, the tissues that cover the bone at the end that engages in the joint. The role of cartilage is to contribute to the movement, but also to absorb the shocks during the movement. In gonarthrosis, the surface of the cartilage can crack, causing pain, swelling and loss of joint mobility.

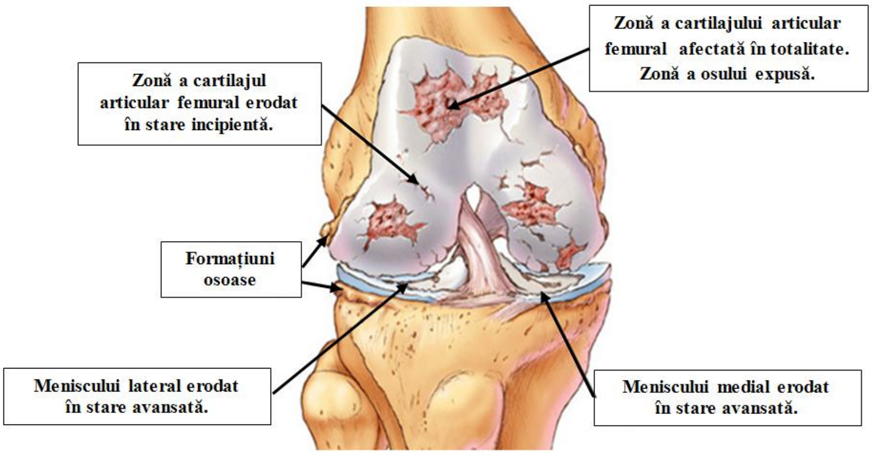


Fig. 2.7. Frontal view of human knee joint affected by advanced gonarthrosis [WAL]

Knee joint gonarthrosis is more common in women, with primary arthrosis accounting for 70%; men are more likely to have secondary arthrosis (53%). Osteoarthritis is the fourth most common cause of women's health problems and the 8th most common cause in men.

• Approximately 40% of all people over 70 are affected by knee osteoarthritis.

• Approximately 80% of the people with osteoarthritis suffer from limited mobility.

• Approximately 25% of these people can no longer carry out the most important activities of everyday life.

The causes that can lead to gonarthrosis are:

a. Deviation of the mechanical axes of the femur and tibia in the frontal plane
b. Misalignment of knee axes in the sagittal plane
c. Overweight
d. Excessive sports activities
e. Traumas
f. Biological causes
g. Meniscal lesions
h. Instability caused by knee ligaments lesions

Endoprosthetic ***arthroplasty*** is defined as a reconstructive surgical intervention with prosthetic replacement of the articular components and bone sacrifice. It is an operation that restores joint mobility as well as the normal functioning of ligaments, muscles and other periarticular structures that perform joint movement.

The aims of endoprosthetic arthroplasty are:

- alleviating the suffering of the patient through the disappearance of pain;
- recovering mobility and joint stability, correcting existing deformities.

The effectiveness of arthroplasty depends on:

1) the quality of articular and mechanical reconstruction of the artificial joint;
2) the integrity and biomechanical balance of periarticular muscles.

The articular endoprosthesis must meet certain characteristics in order to achieve good efficacy:

a. biocompatibility;
b. the fitting must be effective, solid and durable;
c. mechanical low friction functioning between prosthetic components;
d. the design of prosthetic components must reproduce the joint as accurately as possible.

The objectives of prosthetic implant design are as follows:

- simplicity of design and insertion;
- preservation characterized by minimal loss of bone tissue;
- durability;
- cost;
- safety regarding the failure of prosthesis;
- service characterized by increasing the technical options in the surgery for the repair of damaged prosthetic components.

Knee arthroplasty is an intervention that targets patients who generally suffer from gonarthrosis. The objectives of total knee arthroplasty are: redistribution of loads that must be as uniform as possible, axial realignment, pain relief and optimization of mobility. The alignment of both components, both femoral and tibial, takes into account the restoration of the knee transverse axis that must be parallel to the ground. The alignment of the femoral component in the frontal plane must achieve the valgus inclination of the distal femur. At the femoral level, the edges of the prosthesis should overlap with the edges of the osteotomy section [BAC_1986], [SEO_2005], [SIS_2006], [WAL], [WAL_1991].

Knee prostheses must fulfill several criteria:

- functionality as close to normal as possible;
- ability to transfer the articular reaction force to the subjacent bone;
- best fitting of the prosthetic components;
- the highest wear resistance.

Cap.3 Biomechanical evaluations of the normal, osteoarthritic and prosthetic human knee joint

3.2. Experimental protocol

The first objective of this study is to measure the variation in flexion-extension angles of the human joints from the knees of the two lower limbs while the subject performs various tests such as walking on the ground on a set of force platforms of different speeds, ascending-descending the stairs. The experimental data series obtained will be entered as entry data into the joints of a virtual mannequin and a virtual walking simulation will be performed in the ADAMS environmental software. Variation of ground reaction forces will be obtained by experimental data and virtual simulation and will be compared. The second objective involves the determination by numerical simulation of the reaction forces developed in the knee joint in order to perform a finite element method analysis and to obtain stress and deformity maps for normal (healthy) knee joints, for osteoarthritic joints and prosthetic joints.

Measurements were performed on a sample of 7 healthy subjects without pain or musculoskeletal disorders and a sample of 7 patients with a high degree of osteoarthritis in one of the knees. In the case of patients, the measurements were made before the prosthesis operation and after the prosthesis operation. The research was approved by the Ethics Committee of the University of Craiova. Tests performed by healthy subjects were conducted in the Biomechanical Research Laboratory of the Research Platform of the University of Craiova, INCESA. The subjects were equipped with shorts, flat soles so as not to affect the activity and performance of the subjects or data gathering. The biomechanical evaluations of the patients were performed in the Department of Orthopedics-Traumatology of the Emergency County Hospital of Craiova. The tests were performed the day before the prosthesis surgery, and after 2 months the tests were performed again in order to observe the evolution of the patients and the impact of the prosthesis on the kinetic parameters of walking.

3.2.1. Equipment

The data gathering and processing system used is the Biometrics system [WBI], which is commonly used for dynamic motion analysis, in research, as assessment systems and in clinical rehabilitation programs. *Integrated complex 3D analyzing equipment for the human movement* is geared towards research in a wide variety of fields such as biomechanics, robotics, medical bioengineering, traumatology, prosthesis, ergonomics, recovery, sports medicine and sports performance, veterinary biomechanics and pharmacology.

An advantage in using Biometrics is the possibility of simultaneous use and data gathering from a maximum of 24 different sensors, such as electrogoniometers and EMG sensors, simultaneously with the use of force platforms.

3.2.1.1.DataLOG MWX8 is the equipment developed by Biometrics Ltd to monitor and gather data outside the lab. It allows data gathering both in analogue and digital format, and data gathering is done by connecting a transfer cable connector to one of the 8 channels of the **DataLOG** and the second transfer cable connector to the device used to take over data (Goniometers, force platforms, EMG sensors, accelerometers, etc.).

Fig. 3.1. DataLOG Biometrics

Fig. 3.2. Biometrics Electrogoniometers [WBI]

3.2.1.2.Electrogoniometers are sensors that can be used to study the biomechanics of human joints. Biometrics goniometers are designed for fast measurement of human body joint movements in 2 directions with an accuracy of ± 2º for a measured range of minimum 90º.

3.2.1.3.Force platforms

Force platforms in the Biometrics system can connect directly via Bluetooth to dataLINK and DATALog systems for data gathering and reactive force analysis across a wide range of applications. A force platform consists of a sandwich of 2 uniform metal plates with 4 load cells mounted between them.

Fig. 3.4. Force platforms FP4 [WBI]

Force platforms are used in walking analysis to measure ground reaction forces generated by the contact between the sole and the ground when the subject walks on them.

Fig. 3.6. Mounting of Biometrics equipment

The entire real-time data gathering process, viewing charts and settings is carried out through the Biometrics DataLOG software. The software allows exporting / importing .txt formats that can be used later.

For the data gathering during the experimental tests performed on the sample of subjects and on the sample of patients, the following were used:

-2 electrogoniometers SG 150 for knee joint

-2 electrogoniometers SG 150 for hip joint.

-2 electrogoniometers SG 110, for ankle joint.

-6 force platforms FP 4 mounted according to fig. 3.5.

-3 DataLOG wireless data gathering devices, two for the 6 electrogoniometers (hence 12 simultaneous data gathering) and the 3rd one for the 6 force platforms.

3.2.2. Subjects and patients

Two samples were subjected to biomechanical assessments: the sample consisting of 7 healthy subjects and the sample consisting of 5 patients suffering from advanced osteoarthritis.

3.2.3. Tests

The experimental tests were carried out in the Biomechanics Laboratory of the Applied Sciences Research Platform, INCESA of the University of Craiova, for the sample of 7 healthy subjects and at the Emergency Hospital of Craiova for the sample of 5 patients suffering from advanced osteoarthritis.

The tests performed by the subjects in the healthy sample are:
- 1) Test 1 – walking on the ground on force platforms approximately 10 m, for 25 sec.
- 2) Test 2 – walking on the ground on force platforms approximately 10 m, for 20 sec.
- 3) Test 3 – walking on the ground on force platforms approximately 10 m, for 15 sec.
- 4) Test 4 – ascending the stairs (12 stairs)
- 5) Test 5 – descending the stairs (12 stairs)
-6) Test 6 – sitting-standing up from the chair

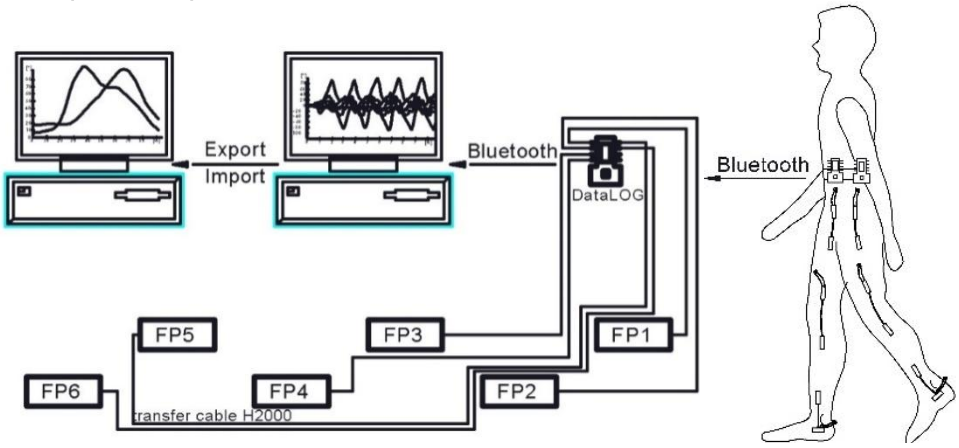


Fig. 3.7. Schema block of data gathering-processing system

3.2.4. Stages of experimental data processing

To obtain the average cycle charts corresponding to each test, both at the individual and sample level, following steps are undertaken:

1. Data are gathered with the Biometrics system, resulting in data files and graphs corresponding to each active channel, using the Biometrics software.
2. Exported files are imported in.txt format from the Biometrics program into the SimiMotion program.
3. The models of cycles and phases required for the subsequent subdivision of imported files, are edited.
4. The import file is divided into cycles and phases based on models edited at the previous stage.
5. 7 consecutive cycles of that chart are selected and normalized to obtain the mean cycle, but first it is recommended to remove the first 2-3 walking cycles and the last 2-3 walking cycles, i.e. those cycles with transient, non-uniform motion, unrepresentative for the test.
6. The selected consecutive cycles are normalized. To get the charts, the Cut Into Phase application that extracts and normalizes each cycle of the imported file was used.

3.3. Results

3.3.1. Results for Healthy subjects – Tests 1-3

The main kinematic parameters obtained by gathering the data corresponding to tests 1-3 for all healthy subjects are found in Tables 3.3.-3.5.

Covering the processing stages of the data collected during the experimental tests, the left and right knee mean cycles for each healthy subject and each test were obtained. In Fig. 3.12.-3.13. the mean knee cycles corresponding to each knee (right and left) and a comparative graph of the two mean cycles corresponding to the 1-5 tests performed by Subject 2, are presented. Similar charts were obtained for each test performed by each subject.

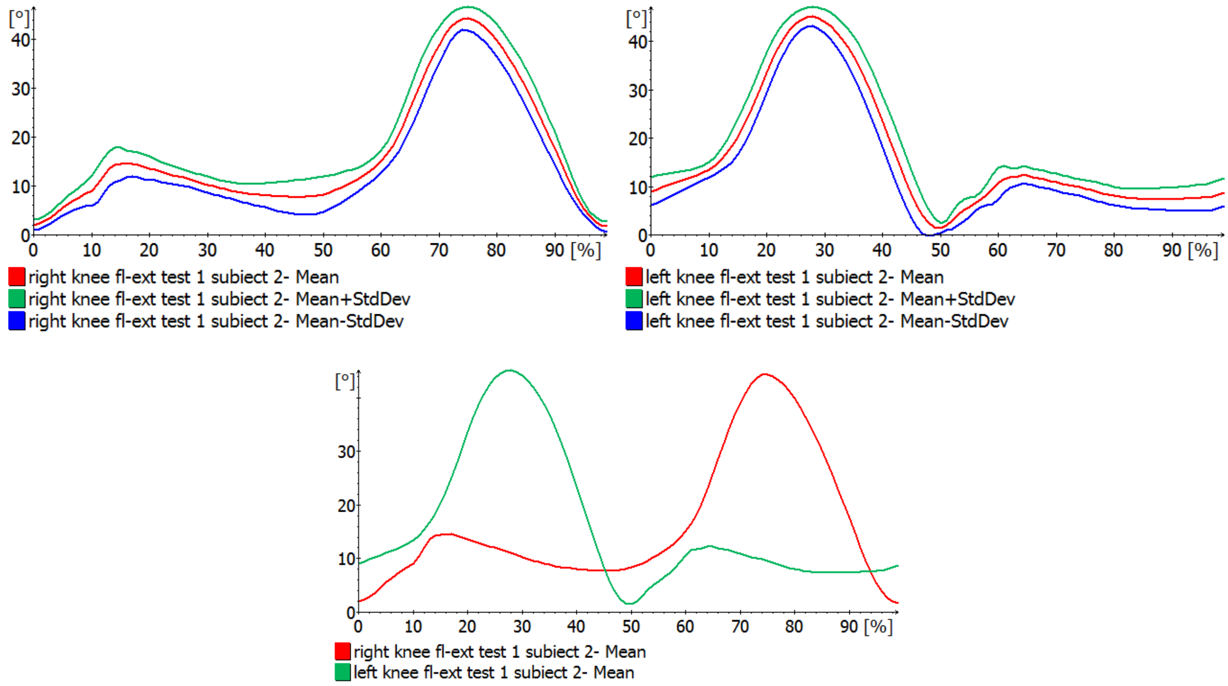


Fig. 3.12. Mean cycle for right knee and left knee and their comparison - Test 1, Subject 2

Next, for each test, corresponding to each of the two knees (right and left), the mean normalized cycle was determined in the whole sample of healthy subjects, starting from the normalized mean cycles of flexion-extension angles of each subject in the sample as input data. In Fig. 3.14.-3.15. the mean sampled cycles for both knees are presented for Test 3. Similarly, mean sample-level cycles were obtained for all tests performed by healthy subjects and by patients.

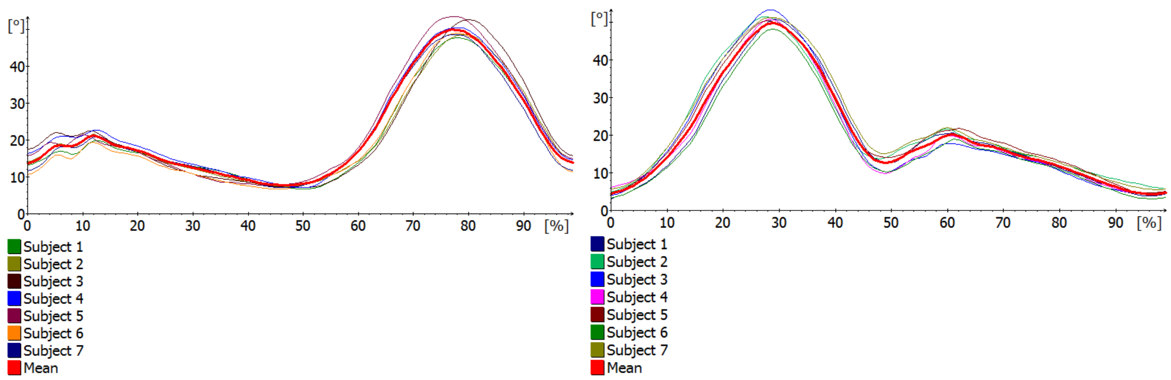


Fig. 3.14. Mean cycle (Test 3), at sample level: a) right knee; b) left knee

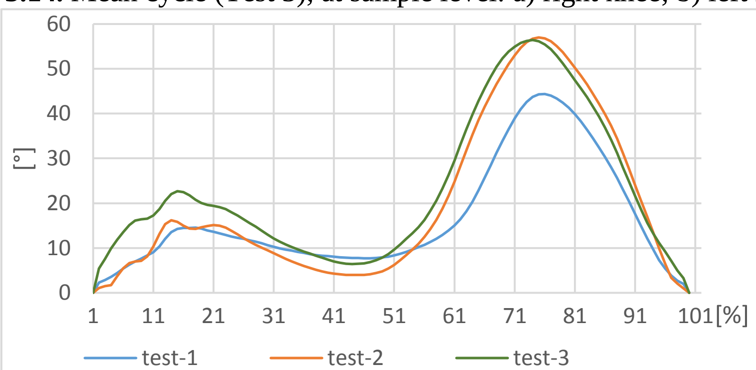


Fig. 3.15. Mean cycles at the sample level of healthy subjects for tests 1, 2 and 3.

The mean cycle of the experimental reaction forces obtained in the sample of 7 healthy subjects on the 6 force platforms (platforms 1, 3 and 5 – for right lower limb and platforms 2, 4 and 6 for left lower limb), corresponding to test 2 (normal walking) are shown in Fig. 3.17.

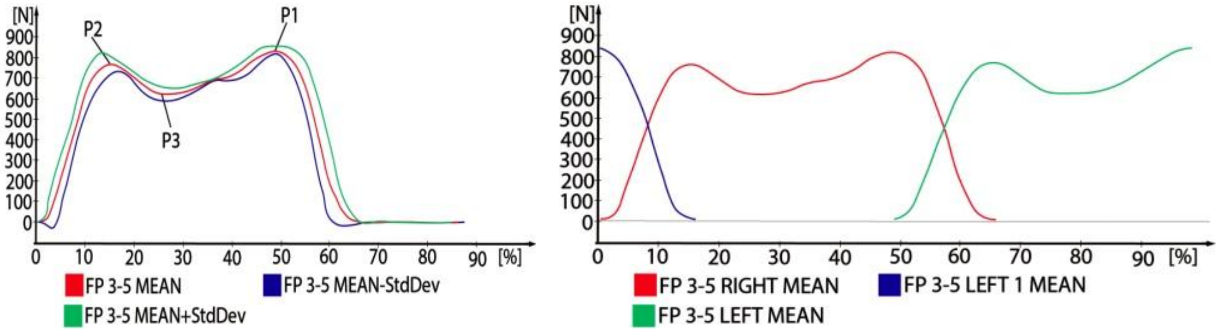


Fig. 3.17. a) Mean cycle of reaction forces determined experimentally for the right lower limb; b) Mean cycles of experimental reaction forces for the right lower limb and for the left lower limb corresponding to a full cycle of the right lower limb

The results (values and allure of the shape of variation graphs of the reaction forces) are similar to those obtained in the papers [NUT_2008, MYL_2006], in which maximum values in the range of [1.15 BW, 1.25 BW] and minimum values (corresponding to point P3) in the range of [0.82 BW; 095 BW] (BW – body weight)- are reported.

3.3.2. Results of patients – Tests 1-3

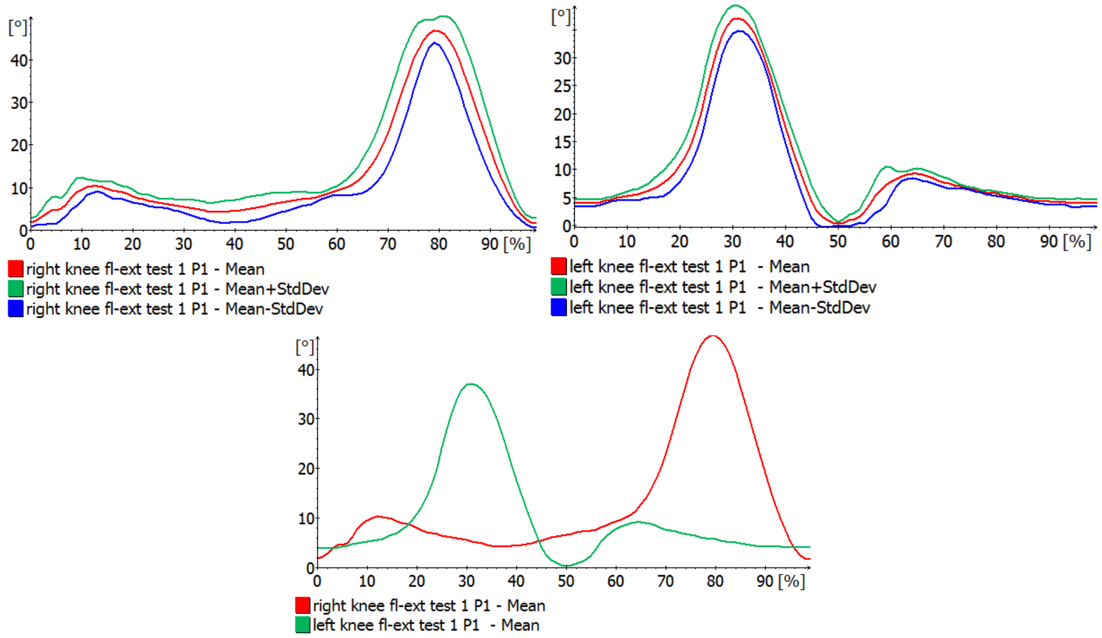


Fig. 3.19. Mean cycle for right knee and left knee and their comparison - Test 1, Patient 1

The patients also performed the 6 tests after a period of 4 months after the prosthesis surgery, the time required to alleviate specific post-operative pains, as well as to undergo a kineto-therapeutic recovery program.

3.3.3.Results – Tests 4 and 5 – Ascending and descending the stairs

The objective of this study is to measure, on the sample of 7 healthy subjects and 5 patients, the reaction forces and variation of flexion-extension angle of knees from both lower limbs during ascending and descending the stairs. The diagrams of the ground reaction forces when descending the stairs, corresponding to Subject 1, obtained for the six force platforms are shown in Fig. 3.28. a) and the corresponding charts of flexion-extension and rotation angles in frontal, ankle, knee and hip plane at both lower limbs, gathered with the Biometrics data acquisition system are presented in Fig. 3.28. b) and c).

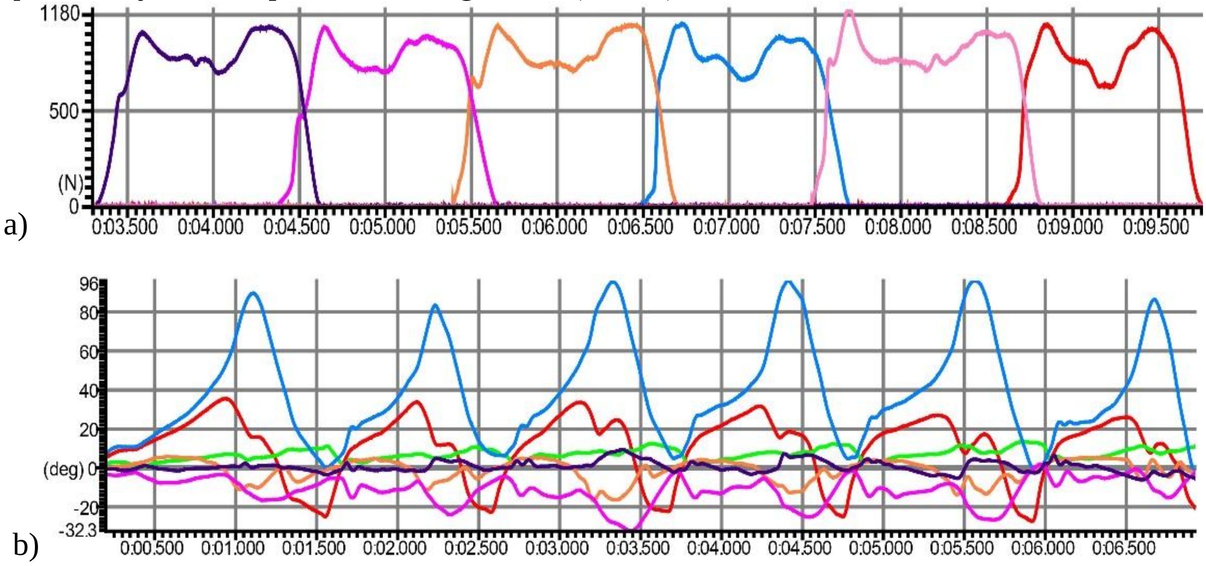
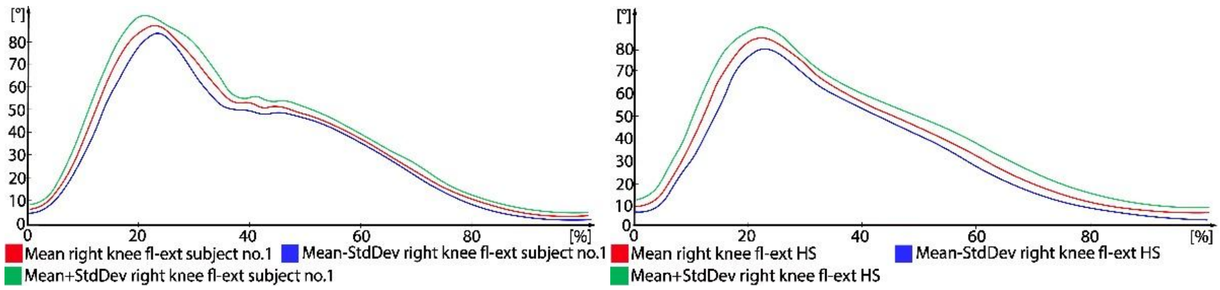


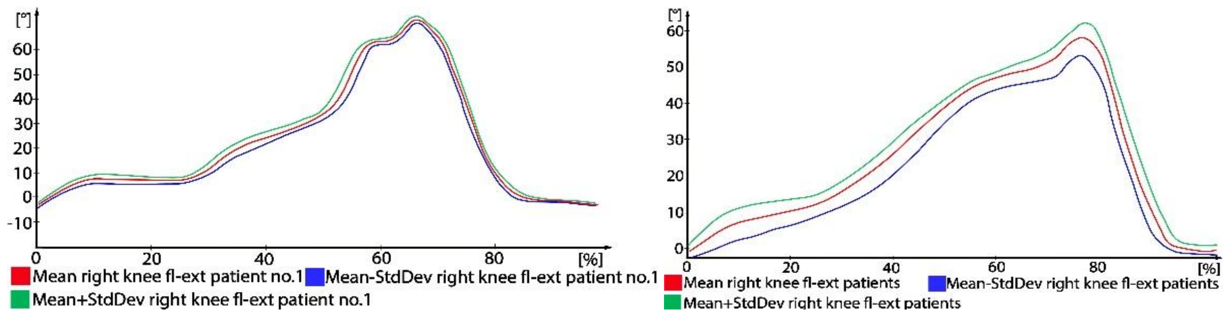
Fig. 3.28. Charts obtained by Biometrics System for the descending test of healthy subject no. 1: a) variation of reaction forces for six force platforms; b) Variation of flexion extension angle and rotating angle in frontal plane for ankle, knee and hip of right lower limb.

For the Stair Ascending Test, the mean flexion-extension cycle for right knee joint of the healthy subject no. 1 is shown in Figure 3.29. a), while in Fig. 3.29. b) the mean flexion-extension cycle is presented at the level of the whole sample. For the Stair Descending Test, the mean flexion-extension cycle for right knee joint of the patient healthy no. 1 is shown in Figure 3.32. a), while in Fig. 3.32. b) the mean flexion-extension cycle is presented at the level of the whole sample.



a) b)

Fig. 3.29. Mean flexion-extension cycle for right knee when ascending the stairs (test 4)*:* a) for subject no.1; b) for the whole sample of healthy patients



a) b)

Fig. 3.32. Mean flexion-extension cycle for right knee OA when descending the stairs (Test 5)*:* a) for patient no.1; b) for the whole sample of patients.

Tables 3.9. and 3.10. show the maximum values of flexion-extension angles and reaction forces corresponding to healthy subjects, and patients, respectively.

Table 3.9. Maximum values of flexion-extension and reaction forces

Subject	Descending		Ascending	
	Max. angle [^{0}]	Max. reaction force [N]	Max. angle [^{0}]	Max. reaction force [N]
1	89	1180	88	1035
2	95	1082	92	942
3	95	1146	90	934
4	85	1144	85	1006
5	93	1116	90	925
6	88	1093	86	982
7	90	1096	87	951
Mean Cycle	90.71	1122.43	88.29	967.86

Table 3.10. Maximum values of flexion-extension and reaction forces

Patient	Descending		Ascending	
	Max. angle [^{0}]	Max. reaction force [N]	Max. angle [^{0}]	Max. reaction force [N]
1	68	1082	61	940
2	61	1044	56	910
3	66	860	60	765
4	63	982	59	844
5	65	1027	59	905
Mean Cycle	64.60	999.00	59.00	872.80

Experimental tests have shown that the maximum values of flexion-extension angles are lower by about 20-26 degrees for the osteoarthritic knee than for the healthy knee for both types of tests: ascending and descending. Variation curves have a similar pattern for patients and for healthy subjects. Also, flexion-extension angle values are higher for descending than for ascending for all experimental test subjects: both healthy subjects, and patients. The study proved that the ground reaction force at the beginning of the support phase is greater than at the end of the support phase. The shape of the vertical reaction force cycle has changed only slightly from normal walking to ascending the stairs, but it has changed considerably at stair descending, with significant differences. This finding is consistent with [STA_2005] and other researchers. The conclusion is that the experimental results obtained in this thesis in the ascending and descending tests correspond, to a great extent, to the results obtained internationally by other researchers.

3.3.4. Results Test 6- Sitting-standing up from the chair

The purpose of the standing-sitting experimental test is to compare the variation range and the amplitude of the flexion-extension angle for the knees of the subjects in the healthy sample and the osteoarthritic knee of the patients before and after the prosthesis operation. All people in the experiment executed 15 consecutive standing-sitting cycles. The data files with the angular amplitudes of knee flexion-extension were obtained during the standing-sitting movement for each person from the report generated by the acquisition system. Fig. 3.40 shows the mean standing sitting cycle corresponding to patient 1. Similar charts were obtained for all patients and for all healthy subjects.

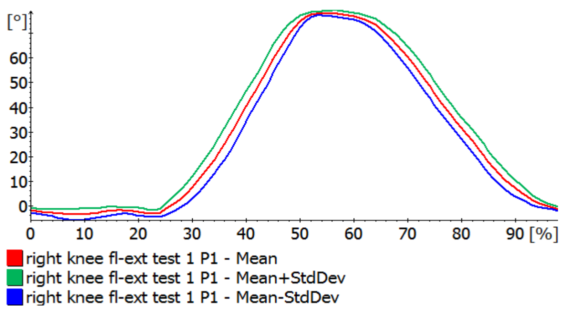


Fig. 3.40. Mean cycle of standing-sitting on the chair for OA knee, patient 2

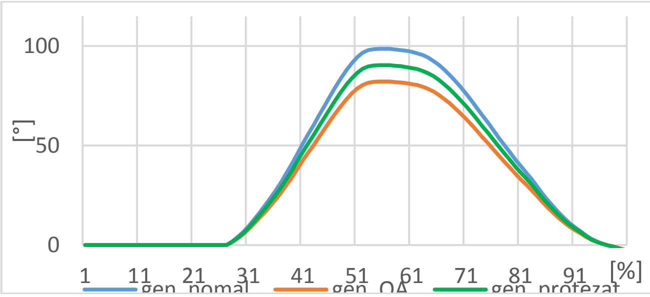


Fig 3.41. Mean flexion cycle at sample level, corresponding to standing-sitting test for a) the normal knee, (b) the osteoarthritic knee (before surgery), c) the prosthetic knee (4 months after surgery)

In Tables 3.15 and 3.17 the maximum values of the flexion-extension angle of the mean cycle corresponding to each knee of each subject and patient are presented in tabular form.

Table 3.15. Maximum value of flexion-extension angle of the mean cycle corresponding to each knee of each subject for each test

	Subject 1		**Subject 2**		**Subject 3**		**Subject 4**		**Subject 5**		**Subject 6**		**Subject 7**		**Mean**	
Knee	**R**	**L**	**R**	**L**	**R**	**L**	**R**	**L**	**R**	**L**	**R**	**L**	**R**	**L**	**R**	**L**
Test 1 [°]	45.2	45.5	51.5	50.4	54.1	52.7	48.1	51.8	54.4	49.2	46.7	47.8	51.1	50.2	50.16	49.65
Test 2 [°]	55.8	50.4	57.7	60.8	52.1	55.2	60.5	58.5	55.6	55.1	52.8	50.2	57.3	56.4	55.97	55.23
Test 3 [°]	57.4	56.7	58.3	60.7	60.4	60.5	59.8	60.3	62.8	60.7	56.5	52.7	55.1	57.4	58.61	58.43
Test 4 [°]	92.9	92.8	82.6	83.7	86.2	81.2	86.1	86.5	93.4	89.5	88.2	88.7	88.2	84.5	88.23	86.7
Test 5 [°]	93.4	93.5	83.4	84.2	88.3	89.3	89.8	89.7	96.1	94.8	91.3	91.5	89.3	91.8	90.23	90.68
Test 6 [°]	102.73	99.5	102.86	99.8	97.6	99.4	96.38	98.9	103.32	98.7	90.93	98.5	91.42	90.2	97.89	98.46

Table 3.17. Maximum value of flexion-extension angle of the mean cycle corresponding to each knee of each patient for each test

	Patient 1		**Patient 2**		**Patient 3**		**Patient 4**		**Patient 5**		**Mean**	
Knee	**R**	**L**	**R**	**L**	**R**	**L**	**R**	**L**	**R**	**L**	**R**	**L**
Test 1 [°]	34.3	35.2	31.8	32.1	31.7	31.2	32.7	33.1	33.4	32.7	32.78	32.86
Test 2 [°]	37.1	36.3	34.2	35.1	33.2	35.4	36.1	37.6	37.2	36.2	35.56	36.12
Test 3 [°]	42.7	42.2	40.4	41.1	41.4	42.6	42.2	41.8	43.2	43.5	41.98	42.24
Test 4 [°]	58.2	56.8	60.1	58.4	59.2	57.8	58.8	58.9	59.7	59.1	59.2	58.2
Test 5 [°]	64.5	63.7	65.9	64.2	64.9	62.8	63.3	63.9	62.8	62.7	64.28	63.46
Test 6 [°]	84.3	88.78	78.12	81.5	86.22	82.1	87.72	82.7	79.2	83.1	80.90	84.84

Cap. 4 WALKING SIMULATION OF A MANNEQUIN, WITH ADAMS PROGRAM

4.1. Introduction

The purpose of this chapter is research on cinematics and dynamics of human walking, carried out in the virtual simulation environment ADAMS [MSC_2013]. Research is focused on normal and fast human walking, as well as ascending and descending stairs. To reach this goal, we went through the main stages described below:

-We used the experimental database with kinematic data measured with electrogoniometer sensors;

- Based on the mean anthropometric sizes corresponding to the sample of healthy subjects used in the experimental tests, we have developed a virtual mannequin in Solid Works;

- We developed a multibody model of the mannequin in ADAMS in the following situations: normal walking, ascending and descending stairs.

- Processing and analysis of the results obtained.

4.2. Building the virtual model of the mannequin in Solid Works

The virtual model of the human mannequin was developed in Solid Works, based on the mean anthropometric data of the sample of human subjects used for experimental analysis. Table 1 shows these data. The final shape of the virtual model of the mannequin is shown in Fig.4.7.

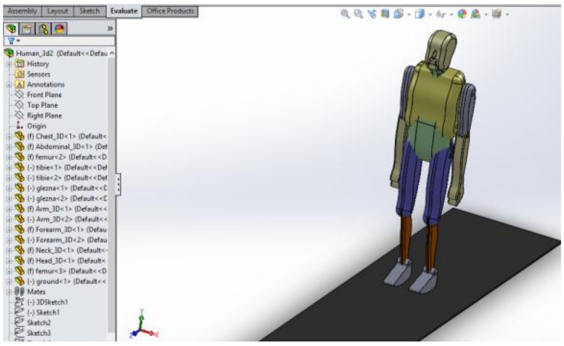


Fig. 4.7. Virtual mannequin in contact with the floor

4.3. Building the multibody model of the mannequin in ADAMS

4.3.1. Defining the kinematic pattern and mass properties

To perform the multibody model of the mannequin in ADAMS, in a first step, the 3D model of the mannequin made in SolidWorks was transferred to the ADAMS database using the parasolid transfer interface.

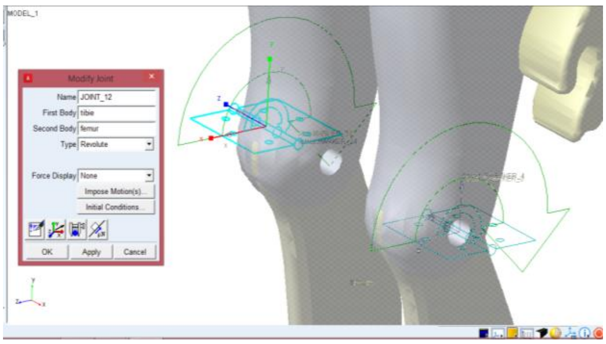


Fig. 4.10. Defining kinematic coupling corresponding to the right knee

In the next step, in each of these 6 couplings, 3 for the right lower limb and 3 for the left lower limb, the motion laws were introduced based on experimentally collected data. Motion laws in the lower limb joints were determined from experimental data collected in tabular form and their introduction into the ADAMS program, followed by their interpolation in the form of SPLINE functions, variation of flexion-extension angle from coupling depending on time, using the Akima method. The variation graphs of the spline function used for defining motion laws in the knee joint are given in Fig. 4.15.

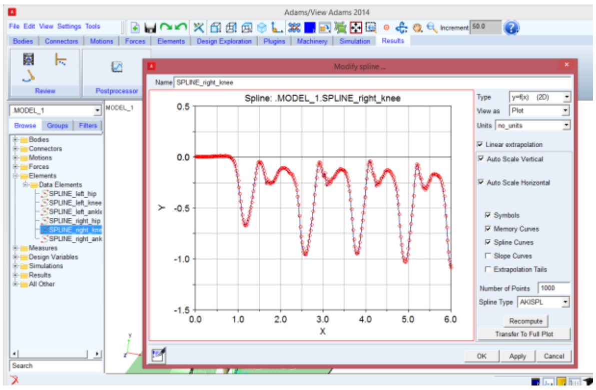


Fig.4.15. Spline variation function of right knee flexion-extension angle [rad], depending on time [s]

Similarly, data for all other joints of the right and left legs were inserted, and the corresponding Spline functions were determined.

4.3.3. Results of the walking simulation

A first part of the results obtained by numerical simulations in ADAMS is the trajectories of some characteristic points, such as, for example, the trajectories made by the mass centers of the mannequin soles (Fig. 19). Fig. 20 shows several successive positions made by the mannequin during walking.

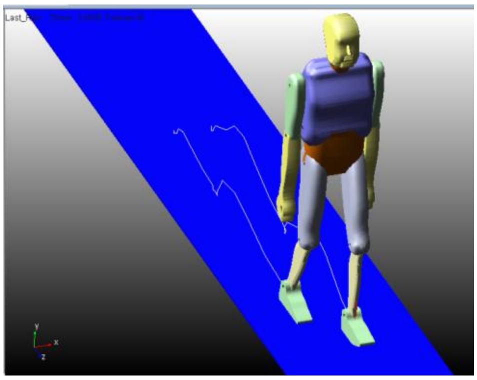


Fig.4.19. Trajectories of mass centers of the soles

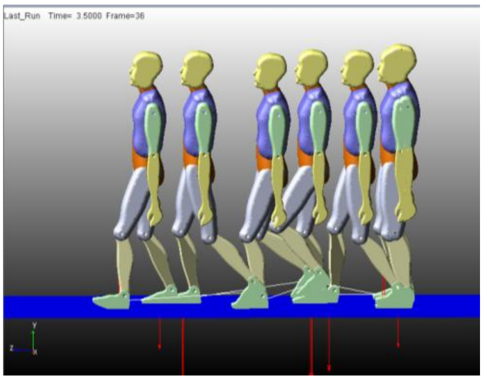


Fig.4.20. Successive positions of the mannequin

The second category of results obtained is the numerical results in the form of graphs of time variation of the kinematic or dynamic parameters of the bipedal mannequin during the walking activity. Fig.4.26 shows the reactions obtained for the right lower limb and the left lower limb. The values of the ground reaction forces and their variation during the walking cycle are useful to compare them with the experimentally obtained results in order to validate the multibody model in ADAMS.

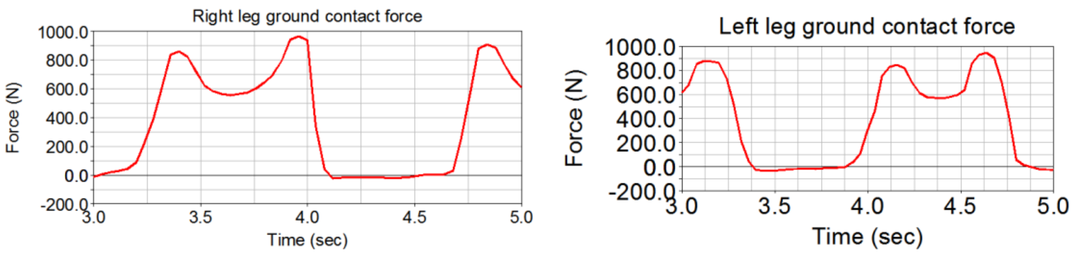


Fig. 4.26. Contact forces (ground reactions) for the right and left legs of the mannequin.

Reaction forces, as well as torsional moments in the lower limb joints: hip, knee and ankle, were also obtained. The vertical reaction forces developed in the right and left knees of the mannequin are shown in Fig.4.28.

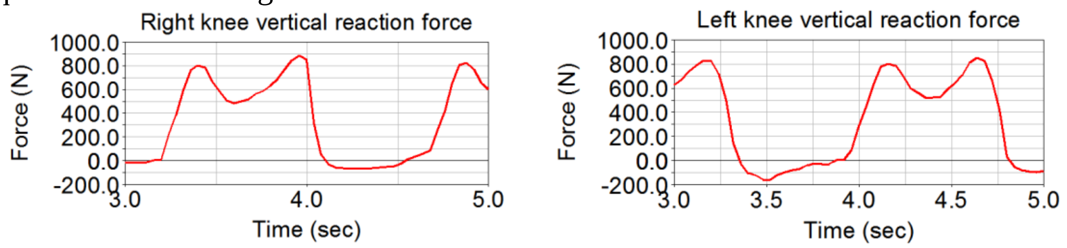


Fig. 4.28. Vertical reaction forces from the right and left knees (on y axis), calculated in ADAMS simulation.

4.4. Modeling and simulation of the walking on stairs of the mannequin

Within this paragraph, we aim to model and simulate the virtual mannequin walking in the stair ascending, starting from experimental biomechanical data gathered and from the mean cycle determined at the sample level for both ascending and descending the stairs by healthy subjects. Thus, experimentally determined motion laws for each of the human leg joints were defined by spline functions, and then introduced as motion laws into the joints of the mannequin.

The motion laws are introduced in the six joints of both legs. The dynamic model obtained in ADAMS is shown in Fig. 4.33.

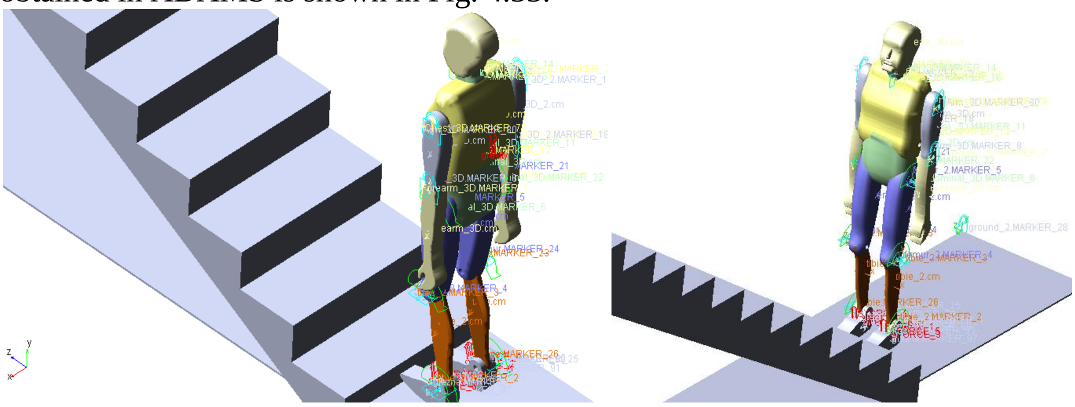


Fig. 4.33. Dynamic model of the virtual mannequin designed in ADAMS.

4.4.1. Results

A first set of results obtained is the law of time variation of ground reaction forces of the lower limbs and the connecting forces of the lower limb joints. The variation of the reaction forces obtained for both legs at the contact between the sole and the ground during the stair ascending test using the ADAMS simulation is shown in Fig.4.34.

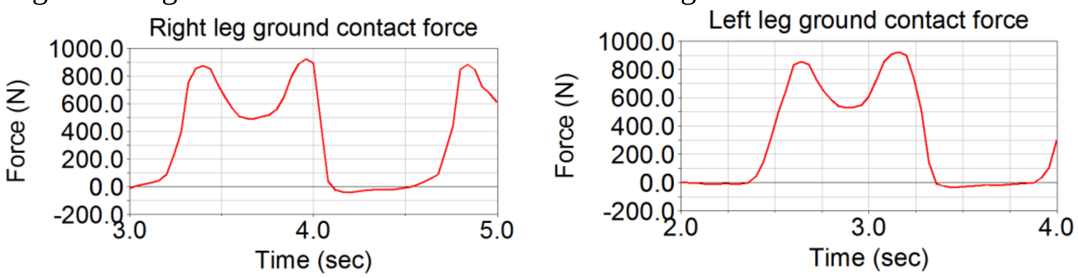


Fig.4.34. Variation laws of ground reaction forces when ascending the stairs.

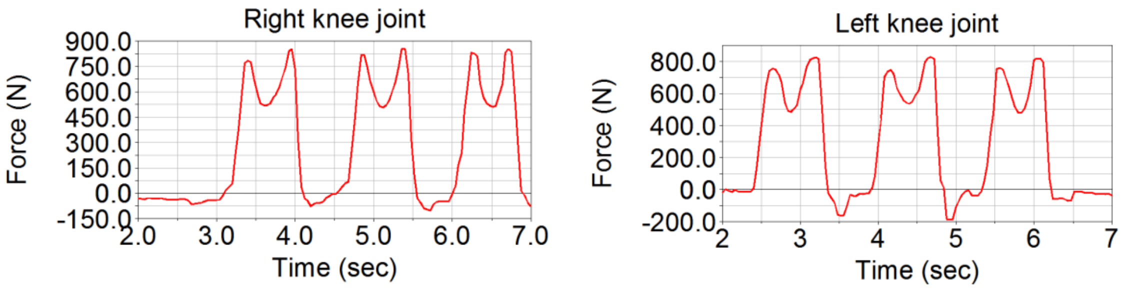


Fig.4.37. Vertical reaction forces in right and left knee joint.

Therefore, the results obtained by numerical simulation in ADAMS for ground reaction forces are comparable to those obtained by experimental measurement. The results obtained are comparable to those obtained by other researchers in their studies [PRO_2007, STA_2005, RIE_2002]. Analyzing the variation of the reaction forces and the moments in the leg joints, it is found that their step is similar to that of the variation of the ground reaction forces determined experimentally. These aspects and conclusions validate the correctness of the virtual modeling of the mannequin and confirm the possibility of taking the forces from the joints for their use in various finite element analyses.

Cap.5. MODELING AND NUMERIC SIMULATIONS OF HUMAN KNEE JOINT WITH CLASSIC PROSTHESIS

5.1. Current studies on virtual modeling of human knee joint

Currently, there are several advanced software programs on the market that can analyze tomographic data and create virtual spatial models. There are many papers dealing with the construction of knee joint components based on images obtained using the Magnetic Resonance Imaging (MRI) method: [PEN_2005, FER_2006], [BAH_2011], [MOH_2011], [YAN_2009], [VID_2008], [CAT_2013].

5.2. Virtual modeling of classic knee prosthesis

To achieve the virtual knee prosthesis model, the DesignModeler application, a preprocessor of the Ansys Workbench 15.07 program, was used. We started from an existing physical model of a classic prosthesis often used in total knee arthroplasty (Fig. 5.16). It consisted of 3 components: a) the femoral component, which is applied to the distal end of the femur, b) the tibial component, which is applied to the proximal end of the tibia, and on the latter c) the polyethylene insert is positioned.

Fig. 5.16. Physical prosthesis for human knee joint.

The virtual modeling of the knee prosthesis was based on the actual sizes of the prosthesis.

Fig. 5.19. Components of knee prosthesis: **A** –Femoral component; **B** –Tibial component; **C** –Polyethylene insert.

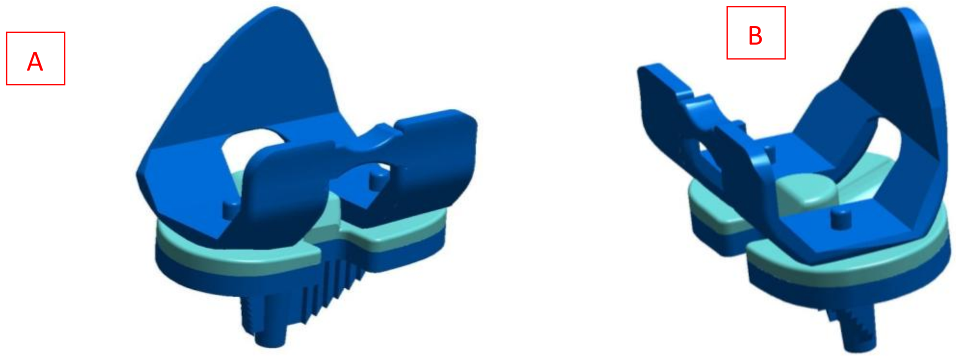


Fig. 5.20. Virtual model of knee prosthesis. **A**–Isometric view to the left.**B** - Isometric view to the right.

5.2.1.Virtual models of prosthetic knee joint with varus inclination.

Six virtual 3D models of prosthetic knee joint assembly have been developed that include the following components: femur, tibia, and the prosthesis with its three components. Each virtual model corresponds to an assembly of prosthetic knee joint, starting from the biological valgus angle between the tibia and the femur, equal to 176^{o}, which is considered normal and increasing by 3^{o}, 6^{o}, 9^{o}, 12^{o} and 15^{o}, that is obtaining the final angles between the axes of: 176^{o}, 179^{o}, 182^{o}, 185^{o}, 188^{o} and 191^{o}.

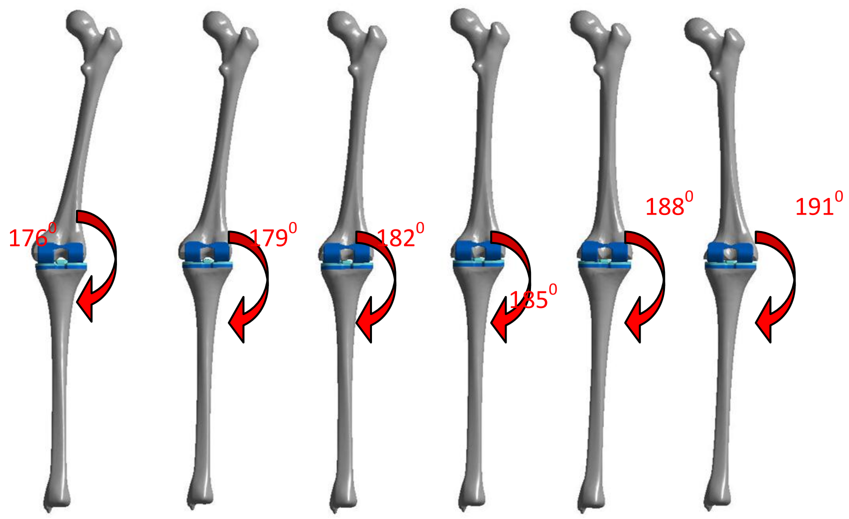


Fig. 5.23. The 7 cases developed for the virtual model of joint-prosthesis assembly with *varus* inclination to 176^{o}, 179^{o}, 182^{o}, 185^{o}, 188^{o} and 191^{o}.

Images for virtual models of prosthetic knee joint with varus inclination are presented in Fig. 5.24.

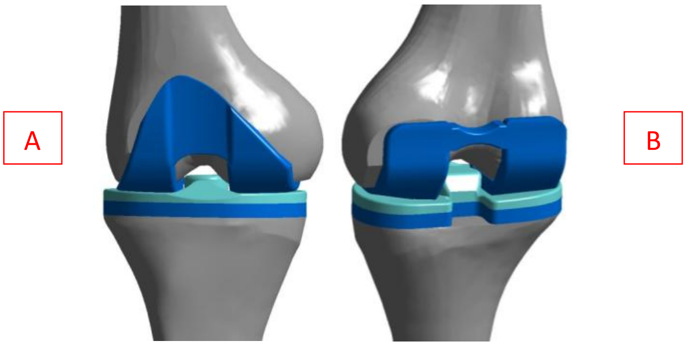


Fig.5.24. Prosthetic knee virtual assembly for varus angle equal to 176^o

5.2.2.Virtual models on prosthetic knee joint with varus inclination and anteroposterior inclination by 5^o.

Virtual knee joint models were analyzed by the finite element method to study tibiofemoral contact area, stress and displacements developed in the human knee joint under various loads. The virtual modeling of the human knee joint was addressed in several articles [KUB_2009, DON_2002, TAR_2014, TAR_2014, YAN_2001, MAT_1999].

Clinical observations show that in the 5° tibial slope it is easier for surgeons to mount knee prosthesis components during total knee arthroplasty and, at the same time, the flexion movement is improved postoperatively (flexion range increases).

The objective of this study is to investigate the effects of this anteroposterior tibial inclination on the values and distribution of contact stresses in the three components of the total knee prosthesis using finite element analysis and numerical simulations on the virtual model. This study aims to compare the values of the contact stresses and the behavior of the three prosthetic components for 24 different situations: for each case of varus inclination, two different loads (800 N and 2400 N) are analyzed in two different sub-situations: anteroposterior inclination angle of 0^o and 5^o. Six virtual 3D models were developed for the prosthetic knee joint with a 5° varus inclination and anteroposterior inclination, and the models include the following components: femur, tibia and prosthesis with the three components.

Compared to previous models, in order to achieve these models, a tibial prosthesis with a 5° inclination in the anteroposterior direction against the knee joint (Fig.5.24).

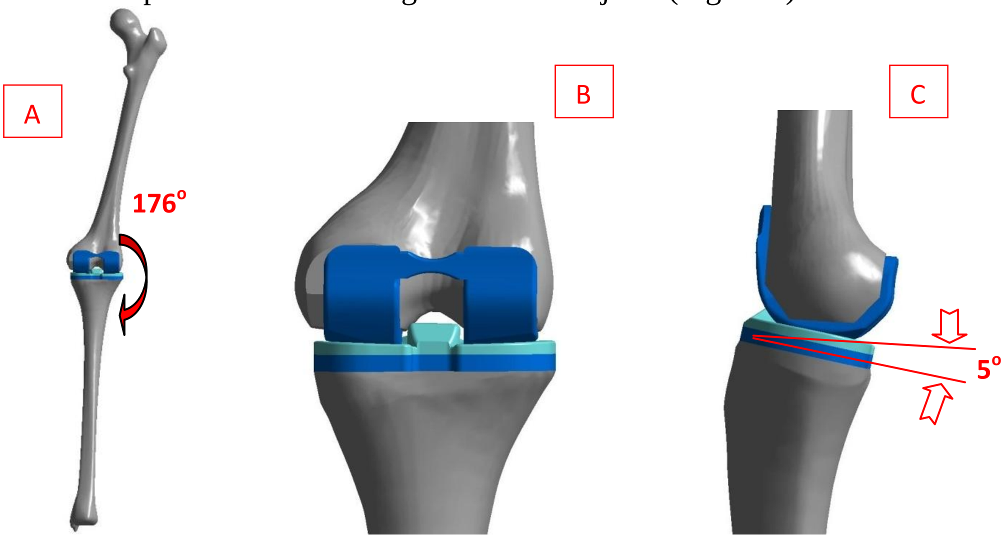


Fig. 5.24. Prosthetic human knee joint at 176^o with anteroposterior inclination of 5^o. A – Joint-prosthesis assembly; **A** – Full assembly; **B** – Frontal image; **C** – Side image;

5.2.3.Numerical simulations of prosthetic human knee behavior with *varus* inclination and anteroposterior inclination of 0^o and 5^o

Using the AnsysWorkbench 15.07 software, numerical simulations and FEM analyses were processed for the eight cases of assemblies of articular prosthesis-prosthesis with anteroposterior slope of 0^o and 5^o, respectively.

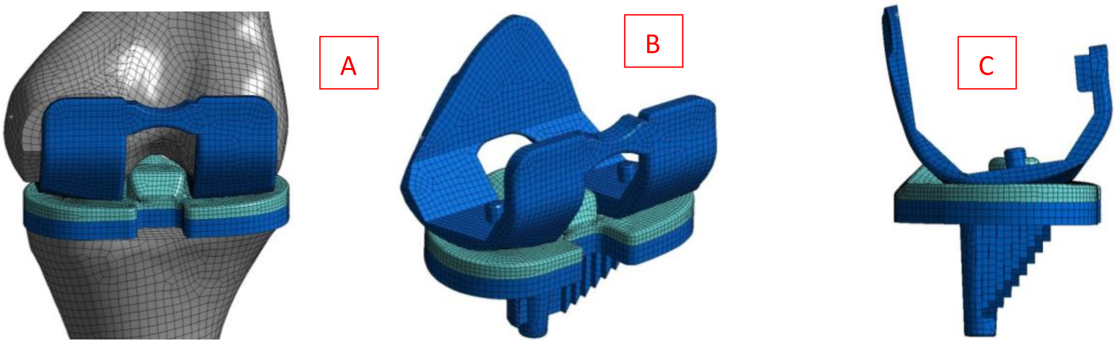


Fig. 5.34. A – Local image of the network of nodes and elements created for the whole model; **B –** Isometric image of the network of nodes and elements created for the prosthesis; **C** – Side image of the network of nodes and elements created for the prosthesis.

The posterior tibial slope that is created during proximal tibial resection in total knee arthroplasty has emerged as an important factor in knee joint mechanics and surgical outcome. But the ideal degree of the posterior tibial slope to recover the knee joint function and prevent complications remains controversial and should vary in different racial groups. The objective of this chapter is to investigate the effects of the posterior tibial slope on the contact stresses in the tibial polyethylene component of total knee prostheses.

5.3.Results

A total of 24 finite element analyses were run, for each case studied, for which the von Mises stress distribution maps and total and axial displacements (after the X-axis) the maximum values of the equivalent von Mises stresses, the maximum values of total and lateral displacements were obtained as results. These sets of results were highlighted at the whole assembly as well as individually at the level of each component, stresses and lateral movements of the prosthesis components being of interest.

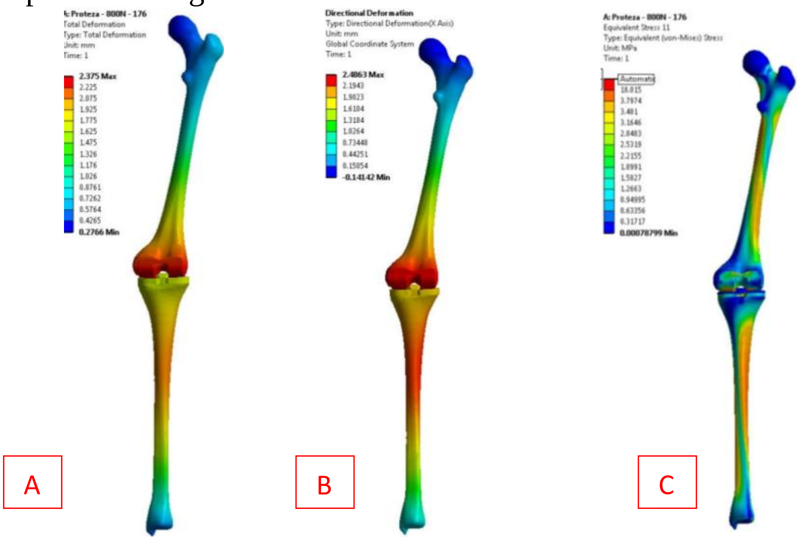


Fig. 5.38. Values of total (A), lateral (B) displacement and maximum equivalent stress for the whole assembly (C) for joint-prosthesis assembly model at 176^o.

Results for the maximum equivalent stresses (von Mises) for the polyethylene insert (Fig. 5.42A), on the tibial prosthesis (Fig. 5.42B) and the femoral prosthesis (Fig. 5.42C) for the joint-prosthesis assembly at 176^{o} and the maximum equivalent stresses (von Mises) for the polyethylene insert (Fig. 5.43A), on the tibial prosthesis (Fig. 5.43B) and the femoral prosthesis (Fig. 5.43C) for the joint-prosthesis assembly at 176^{o} with a 5^{o} anteroposterior inclination are extracted. The images represent "Top" and "Bottom" views of the results.

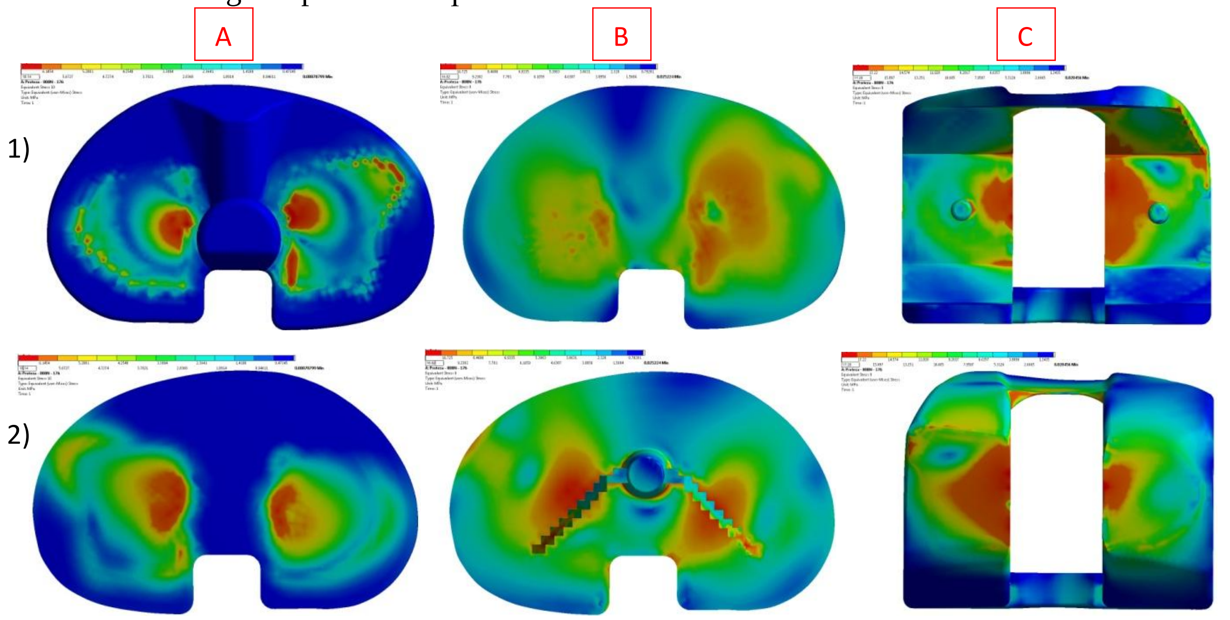


Fig. 5.42. Distribution map of maximum equivalent stresses in the polyethylene insert **(A)**, tibial prosthesis **(B),** femoral prosthesis **(C)** for the joint-prosthesis assembly at 176^{o}; **1**–Top images; **2**–Bottom images

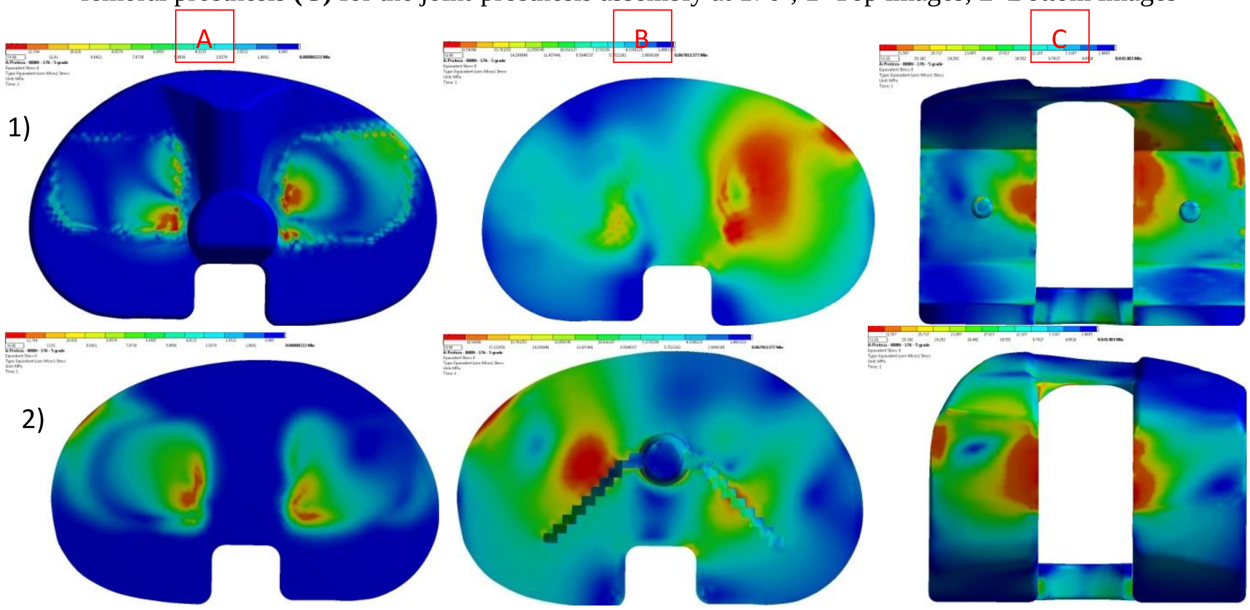


Fig. 5.43. Distribution map of maximum equivalent stresses (Von Mises) obtained in the polyethylene insert **(A)**, tibial prosthesis **(B)** and femoral prosthesis **(C)** for the joint-prosthesis assembly at 176^{o} which shows a 5^{o} anteroposterior inclination; **1** – Top images; **2** – Bottom images.

In the same manner, the stress maps for all **24 cases** with 800N and 2400N load, respectively, have been determined: **12 models** for the joint-prosthesis assembly with varus inclination at 176^{o}, 179^{o}, 182^{o}, 185^{o}, 188^{o} and 191^{o}, 0^{o} anteroposterior inclination and **12 models** for the joint-prosthesis assembly with varus inclination at 176^{o}, 179^{o}, 182^{o}, 185^{o}, 188^{o} and 191^{o} with a 5^{o} anteroposterior inclination. In total, **24 analyses of stresses and deformations states** were performed and stress maps and displacement maps for each component of the prosthetic knee assembly were obtained. The values of the von Mises stresses and of the displacements in the tibia and femur, as well as in the three prosthetic components, *for the 24 models with varus inclination* with 0^o anteroposterior tibial inclination (or practically without anteroposterior inclination) and 5^o respectively for the two external load variants: **800N and 2400N** are determined.

5.4. Disscusions

For a compressive force equal to 2400N and 800N, respectively, as used in our analyses, for a normal, physiological valgus angle of 176° and for a 0° flexion angle, the maximum values obtained for von Mises stresses, are similar to those in [SZI_1996]. Analyzing the results obtained by the numerical simulations, it is observed that as the varus inclination angle increases, von Mises stress values increase for all the components of the prosthesis. In all cases, the stress values on the three components of the prosthesis are similar, with small differences, but we can conclude that higher values are developed on the polyethylene insert, followed by the values developed on the femoral component and on the tibial component, respectively. Based on these results, we can suggest that the tibial cutting should be made at a 5° angle, which confirms the clinical observations and conclusions, and overlaps with the findings in the study [SZI_1996]. This result of the experiment is of major clinical importance, guiding the orthopedic surgeon at the best bone cutting angle.

Cap. 6. Modeling, numerical simulations and prototyping the new knee endoprosthesis model

Clinical observations and specialist studies have led to a generally accepted conclusion that current prostheses promoted and marketed by world-renowned companies have only one radius of curvature of the femoral component, and flexion movements of the shank on the thigh are limited to an angle value of about 110-120^o. This inconvenience prevents the patient from benefiting from certain positions and greater amplitudes of the flexion angle, as in the case of the normal knee joint. That is why patients have permanent discontent. The human knee is able to flex up to about 160° [NAG_2002, NAK_2000]. High flexibility of the knee is very important for people working in constructions and agriculture, gardening. Current prostheses and surgical techniques may not meet the needs of patients requiring knee flexion > 120° for daily activities. Most biomechanical studies related to knee arthroplasty focused on knee flexion <120° [DLI_2000, LIG_2002].

6.1. Experimental study

Clinical observations and specialist studies have led to a generally accepted conclusion that current prostheses promoted and marketed by world-renowned companies have only a single radius of curvature of the femoral component, and flexion movements of the shank on the thigh are limited to an angle value of about 110^0. This inconvenience prevents the patient from benefiting from certain positions and greater amplitudes of the flexion angle, as in the case of the normal knee joint. That is why patients have permanent discontent. The studies that have been undertaken over the past few years by both orthopedic surgeons and manufacturing companies have failed to explain the causes of this limitation of the flexion movement of the shank on the thigh.

Based on these observations, we initiated a study that would allow us to find the causes that determine the limitation of the flexion movement in the knee prosthesis.

The proposed clinical technique for the implantation of the new knee prosthesis is based on the technique used for the total knee prosthesis KYN® and on a similar instrument. This is a fixed tibial insert prosthesis, available in the rear-stabilized version and the posterior cruciate ligament retaining version. The use of the same instrumentation and the same implantation method is possible because the internal architecture of the femoral component, that is, the hollow by which it is attached to the bone, of the newly created prosthesis remains similar to that of the KYN prosthesis. The differences between the two prostheses are recorded in the external architecture, that from the joint. In the case of KYN prosthesis, there is only one radius of curvature of the outer surface. This construction does not follow the architecture of the distal femoral epiphysis which has three radii of curvature of the joint surface. Being built with a single radius of curvature, the KYN prosthesis has a number of **disadvantages**. In the first phase, the quadriceps muscle tendon remains too long by the resection of the femoral patellar trochlea. This makes the postoperative patient unable to perform the maximum extension.

Adjustment of the length of the quadriceps for the patient to perform the maximum and even over maximum extension is accomplished over a long time, not by training the muscle but by adapting it to the length of the lever arm that has been reduced by fitting the KYN prosthesis. The retraction of the femoral quadriceps muscle to adapt to the decreased lever arm determines a gradual limitation of the flexion of shank on the femur over 90 degrees. This limitation of the flexion of shank on the thigh is also enhanced by the resection of the posterior part of the femoral condyles. This resection is required in the KYM prosthesis to be able to mount this implant with a single radius of curvature. This resection decreases the distance between the posterior edge of the tibial plateau and the posterior cortical plate of the femur which causes the edge of the tibial plate to reach the posterior cortical of the femur at a much lower angle than the natural angle. Thus, more than 30 degrees of knee flexion are lost, giving the patient the impossibility of achieving the maximum flexion of the prosthetic knee.

The prosthesis proposed in the PhD thesis complies with the three bending radii of the distal femoral epiphysis. Thus, the inconveniences of extension and flexion of the KYM prosthesis are removed. The aforementioned have been demonstrated experimentally.

6.2. Stages of operating technique for the proposed prosthesis

1. Distal femoral resection;
2. Proximal tibial resection;
3. Verification of ligament spaces: The verification of ligamentous spaces is performed by the spacer method and involves the provision of two ligamentous balances: a) Ligamentous balance in extension; b) Ligamentous balance in flexion
4. Selection of the size of femoral component and rotation adjustment – Spacer method;
5. Performance of anterior and posterior femoral resections and facettes;
6. Preparation of intercondylar area and tests:
7. Preparation of trochlea consolidation involves:
8. Intercondylar femoral preparation:
9. Tibial preparation;
10. Patellar preparation;
11. Location of final components.

Each of these stages is presented and the surgical technique is highlighted.

Fig 6.3. Distal femoral resection:

Fig. 6.4. Distal femoral resection: a) mounting the resection guide support; b) view on the resection level using the curved blade.

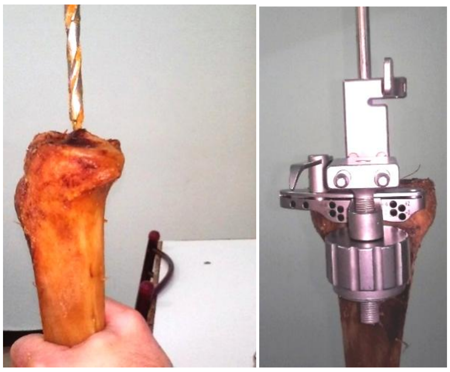


Fig.6.6. Intramedullary guidance system;

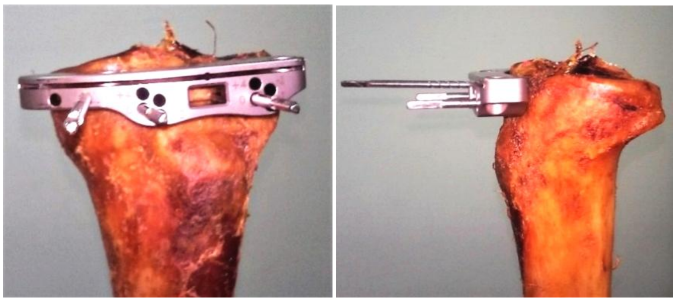


Fig.6.7. Positioning of the feeler

Fig. 6.8. Proximal tibial resection

Fig. 6.10. Performance of anterior and posterior femoral resections and facettes

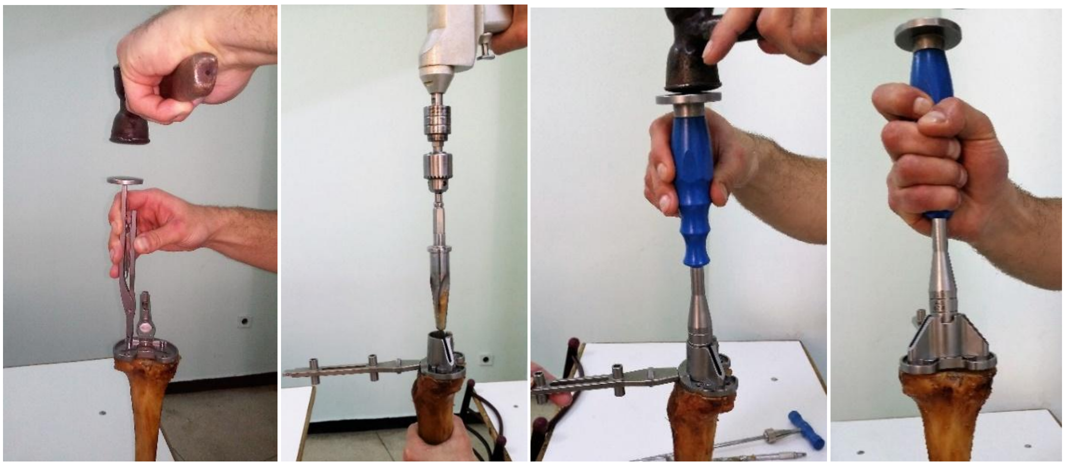


Fig. 6.13. Tibial preparation

The conclusion is that, after fitting the prosthesis on the femur, the lever arm elongates and this makes it difficult for the prosthetic patient to extend the shank on the thigh. Gradually, after a few weeks, the quadriceps muscle retracts, adapting its fiber length to the length of the bone segment. The retraction of the quadriceps muscle makes it difficult to flex the shank on the thigh over 90-100^0 up to total jam. This is one of the reasons why prosthetic patients cannot achieve complete flexion of the shank on the thigh, namely 135^0 degrees. The resection of the femoral trochlea in the surgical technique for fitting the femoral component is responsible for this situation. Continuing experimentally, I found that, at the flexion of over 90 degrees, the edge of the prosthetic tibial component insert is very close to the posterior cortical of the femur, so the two come in contact at 110 degrees preventing further flexion.

Conclusion:

1. Current knee prostheses whose femoral component has a single radius of curvature, through the resection of the trochlea and femoral condyles, modify the knee biomechanics, hindering the maximal flexion and extension movements, and in particular the flexion movement which, in most cases, remains stuck at 100-110 degrees.
2. The presence of trochlea and femoral condyles is essential in normal knee biomechanics. The two anatomical formations stabilize the joint and allow the knee flexion and extension movements to the maximum.
3. The development of femoral trochlea and femoral condyles is a consequence of phylogenetic and ontogenetic development and adaptation to bipedal walking.

6.3.Virtual model of the proposed prosthesis

To achieve the virtual model of the proposed knee prosthesis, we used DesignModeler, a preprocessor of Ansys Workbench 15.07. The virtual modeling of the proposed knee prosthesis was performed starting from the virtual sizes of the original model using advanced commands available in the DesignModeler preprocessor. The construction of the proposed model (Fig. 6.23.a) was performed starting from the original knee prosthesis model (Fig. 6.23.b).

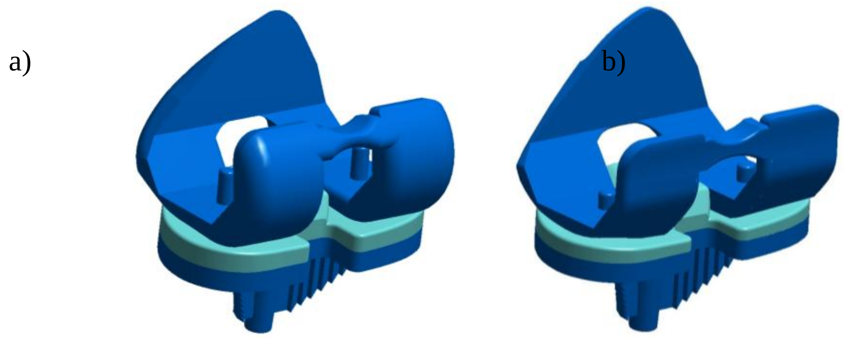


Fig. 6.23.a) Virtual model of proposed prosthesis; **b)** Virtual model of original prosthesis.

Thus, the first model of the proposed prosthesis - joint assembly for a *varus* inclination at 176^o has been made (Fig.6.26).

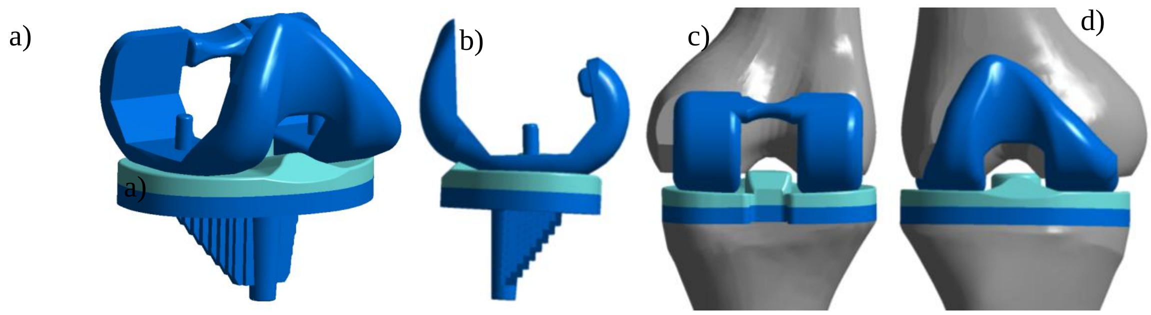


Fig. 6.26. a) Proposed prosthesis – Isometric view; **b)** Proposed prosthesis – Side view; **c)** Back view of proposed prosthesis – knee joint assembly in *varus* at 176^o; d**)** Frontal view of proposed prosthesis – knee joint assembly in *varus* at 176^o.

6.4.Virtual models for the cases of knee joint – proposed prosthesis with *varus* inclination.

For the virtual models of the prosthetic knee joint, six 3D virtual models were developed including the following components: femur, tibia and proposed prosthesis with the 3 components.

In addition to these 6 cases with *varus* inclination, other 12 virtual models of the prosthetic knee joint with 3^o (6 cases) and 5^o (6 cases), respectively, anteroposterior tibial inclination were developed.

For the realization of these models, compared to the previous models, a tibial prosthesis inclination with 3^o and 5^o, respectively, in the anteroposterior direction of the knee joint was made by the virtual inclined resection of the tibia bone in the anteroposterior direction.

Advantages of proposed prosthesis solution compared to existing prostheses

The tibial and femoral components are made of chromium-cobalt alloy whose finished surface provides precise shapes and a minimum friction coefficient. By its construction, by the radius of space curvature, by increasing the contact surface between its components, thus reducing the contact pressure, von Mises stress values, and the wear of its components, the proposed prosthesis allows the restoration of the physiological capabilities of the knee joint affected by gonarthrosis and increases the durability of the prosthesis by reducing wear, but also by restoring the natural mobility of the joint, eliminating joint pain.

- The proposed prosthesis is aimed at young people - in uncemented version, providing a more durable natural fixation, as well as at the elderly - in the cemented version.
- The design of the implant components gives the prosthetic knee an increased flexion, very close to the normal one for a healthy knee (150°)

- In order to significantly reduce the wear of the implant elements and reduce the contact pressures, the design of the proposed prosthesis provides the increase of the contact surfaces, a very important aspect in the case of moderate flexion (walking) but especially in the case of the acute flexion (standing from the chair, ascending/descending the stairs).

6.5.Numerical simulations and finite element analyses of the original knee-prosthesis assembly

For all 18 cases of elaborated prosthetic knees, the controls were used for the finer local discretization required in the contact area and the area of interest.

Table 6.1. Network of nodes and finite elements made for the 18 geometric models.

Case	0^{o} tibial anteroposterior inclination		3^{o} tibial anteroposterior inclination		5^{o} tibial anteroposterior inclination	
	No. of Nodes	No. of Elements	No. of Nodes	No. of Elements	No. of Nodes	No. of Elements
176^{o}	298.959	92.070	302.851	93.988	302.435	93.640
179^{o}	298.696	92.687	300.711	93.226	300.778	93.462
182^{o}	296.696	92.131	298.522	92.476	297.999	92.488
185^{o}	294.352	91.448	295.923	91.844	297.573	92.270
188^{o}	291.291	90.137	293.431	91.076	292.250	90.542
191^{o}	289.620	89.875	290.835	90.283	291.685	90.712

The boundary conditions for the analysis are the same as those used for the initial prosthesis.

6.6. Results

A total of 30 analyses were performed for all models designed throughout the proposed knee joint-prosthesis assembly as follows:
- imposed load of 800N for the 6 models in *varus* without anteroposterior inclination;
- imposed load of 2400N for the 6 models in *varus* without anteroposterior inclination;
- imposed load of 2400N for the 6 models in *varus* with a 3^{o} anteroposterior inclination;
- imposed load of 800N for the 6 models in *varus* with a 5^{o} anteroposterior inclination;
- imposed load of 2400N for the 6 models in *varus* with a 5^{o} anteroposterior inclination;

6.6.1. Results obtained for the analyses of 176^{o} case

The results obtained for the first case analyses (176^{o}) are presented below.

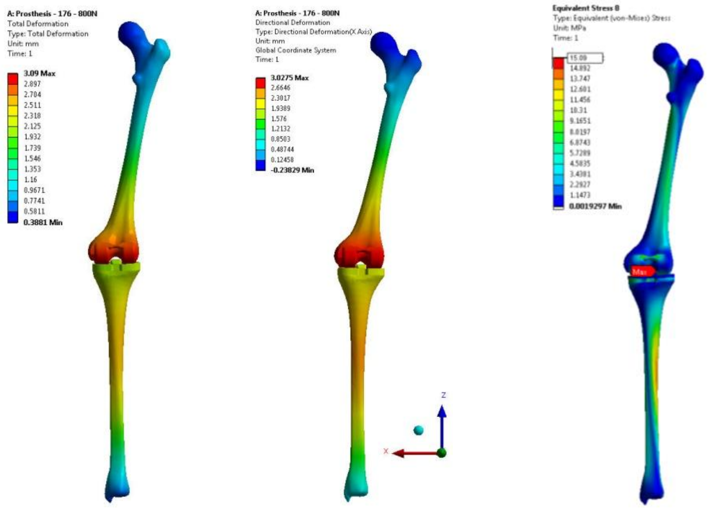


Fig. 6.35. Values of total (a), lateral (b) displacements and maximum equivalent stress for the whole assembly (c) for the model of knee joint-prosthesis assembly proposed at 176^{o} for a load of 800N.

The maximum equivalent stresses on the polyethylene insert (Fig. 6.39.a), on the tibial prosthesis (Fig 6.39.b) and the femoral prosthesis (Fig 6.39.c) for the joint-prosthesis assembly proposed at 176^{o} with 0° anteroposterior inclination are extracted. Images show top and bottom views.

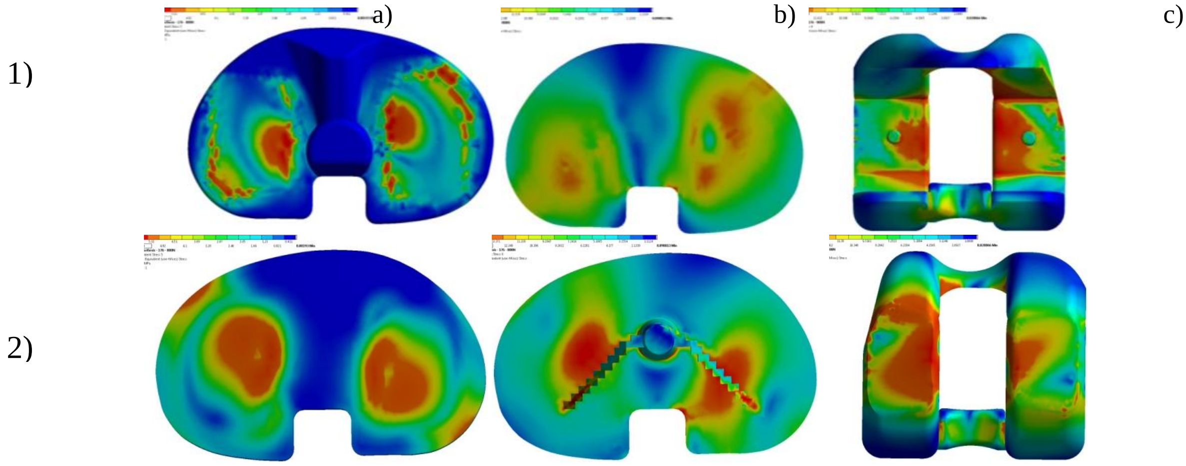


Fig. 6.39. Maximum equivalent stresses (Von Mises) obtained for polyethylene insert **(a)**, tibial prosthesis **(b)** and femoral prosthesis **(c)** for the joint-prosthesis assembly proposed at 176^{o} for a load of 800N; **1** – Top images; **2** – Bottom images

The values obtained for the **<u>30 analyzed cases</u>**, for the 6 angles of inclination in *varus*, for the three anteroposterior tibial inclination angles: 0^{o}, 3^{o} and 5^{o} and for the two external load variants: **800N and 2400N** are presented, in synthesized form, in **Tables 6.3-6.7**.

Table 6.5. Values obtained for *varus* cases with 0^{o} anteroposterior inclination for **2400N**

Inclination 0^{0}	**Cases 2400N**	***POLI. Stress [MPa]***	***P.T. Stress [MPa]***	***P.F. Stress [MPa]***	***Femur Stress [MPa]***	***Tibia Stress [Mpa]***	***Total displacement [mm]***	***Lateral displacement [mm]***
1	**176^{0}**	45.12	41.33	43.79	25.03	32.11	9.27	9.08
2	**179^{0}**	47.01	42.97	45.67	26.32	33.07	3.74	3.20
3	**182^{0}**	48.68	44.36	47.12	27.73	34.21	2.07	0.49
4	**185^{0}**	50.07	45.98	48.77	29.17	35.43	4.97	0.77
5	**188^{0}**	51.63	47.43	50.15	30.83	36.71	10.55	1.37
6	**191^{0}**	53.17	49.89	51.73	32.43	38.02	14.79	1.81

Table 6.6. Values obtained for *varus* cases with 3^{o} anteroposterior inclination for **2400N**

Inclination 3^{0}	**Cases 2400N**	***POLI. Stress [MPa]***	***P.T. Stress [MPa]***	***P.F. Stress [MPa]***	***Femur Stress [MPa]***	***Tibia Stress [Mpa]***	***Total displacement [mm]***	***Lateral displacement [mm]***
1	**176^{0}**	37.87	33.96	34.88	18.59	25.62	10.83	9.96
2	**179^{0}**	39.11	35.43	36.01	19.96	26.49	5.82	3.95
3	**182^{0}**	40.76	37.03	37.68	21.32	27.83	3.88	0.37
4	**185^{0}**	42.15	38.47	39.33	22.78	28.79	6.32	0.81
5	**188^{0}**	43.67	39.98	40.37	24.13	29.82	11.87	1.44
6	**191^{0}**	45.07	41.18	41.64	25.42	31.03	16.41	1.91

Table 6.7. Values obtained for *varus* cases with 5^{o} anteroposterior inclination for **2400N**

Inclination 5^{0}	**Cases 2400N**	***POLI. Stress [MPa]***	***P.T. Stress [MPa]***	***P.F. Stress [MPa]***	***Femur Stress [MPa]***	***Tibia Stress [Mpa]***	***Total displacement [mm]***	***Lateral displacement [mm]***
1	**176^{0}**	31.86	28.11	28.93	14.75	21.15	11.05	9.96
2	**179^{0}**	33.17	29.78	30.43	16.09	22.08	6.10	3.90
3	**182^{0}**	34.63	31.11	31.99	17.45	23.13	4.43	0.39

4	**185^0**	36.09	32.48	33.41	18.87	24.25	6.71	0.82
5	**188^0**	37.61	33.99	34.86	20.11	25.31	12.41	1.46
6	**191^0**	39.07	35.42	36.23	21.47	26.47	16.97	1.94

Compared to the results obtained from the analysis of the original prosthesis, the proposed prosthesis improves (relieves) the load on the proposed prosthesis components, and also on the femur and tibia by 15% to 25%. The results obtained show good behavior in the proposed prosthetic model and also the effectiveness of the materials used. The proposed prosthetic device reduces the stresses and pressures in the prosthetic human knee joint.

Cap.7.PUBLISHED PAPERS, OWN CONTRIBUTIONS AND FUTURE WORK

7.1. Published papers

Chapter in book edit in International Publishing House:

1. Daniela Tarnita**,** D. Popa, C. Boborelu, N. Dumitru, **D. Calafeteanu**, D.N. Tarnita**, Experimental Bench Used to Test Human Elbow Endoprosthesis,** New Trends in Mechanism and Machine Science, Vol 24 (2015), pp. 669-677, **Springer Publishing House,** https://link.springer.com/chapter/10.1007%2F978-3-319-09411-3_71**.**

Papers in ISI Journals with IF

2. DN Tarniţă, Daniela Tarniţă, D Grecu, **D Calafeteanu**, B Căpitănescu, **New technical procedure involving Achilles tendon rupture treatment through transcutaneous suture,** Rom J Morphol Embryol, 2016, 57(1):211–214 http://www.rjme.ro/RJME/resources/ files/570116211214.pdf
3. TARNITA Daniela, **CALAFETEANU Dan**, GEONEA Ionut, TARNITA Danut-Nicolae Effects of malalignment angle on the contact stress of knee prosthesis components, using Finite element method, Rom J Morphol Embryol, 2017 (acceptata, in curs de publicare)

Papers in ISI Journals, BDI Journals and ISI Proceedingsuri

4. Tarnita Daniela**, Calafeteanu D**., Matei I, Tarnita D.N, **Experimental Measurement of Flexion-Extension in Normal and Osteoarthritic Knee During Sit-to-Stand Movement,** Applied Mechanics and Materials Vol. 658 (2014) pp 520-525, www.scientific.net/ AMM.658.520 **(Scopus)**.
5. Petcu Alin**, Calafetenu Dan**, Tarnita Danut Nicolae, **Numerical simulations and experimental flexion-extension measurements of human leg joints during squat exercises**, Medicina sportiva - Journal of the Romanian Sports Medicine Society, jun, 2017,
6. Daniela Tarnita, **Dan Calafeteanu**, Marius Catana, Ionut Geonea, Dan Tarnita, **Development of a Three-Dimensional Finite Element Knee Prosthesis Mode**l, Applied Mechanics and Materials. Vol. 822, 150-155, 2016, www.scientific.net/AMM.822.150.
7. **Calafeteanu, D.,** Tarnita, D., Catana, M., and D.N. Tarnita, **Influences of Antero-Posterior Tibial Slope on the Prosthetic Knee Contact Stresses,** Applied Mechanics and Materials, Vol. 823, 137-142, 2016, www.scientific.net/AMM.823.137.
8. **D. Calafeteanu**, Daniela Tarnita, M. Catana and D. N. Tarnita, **Influences of Varus Tilt on the Stresses in Human Prosthetic Knee Joint,** Applied Mechanics and Materials, Vol. 823, 143-148, 2016, www.scientific.net/AMM.823.143

9. TARNITA Daniela, ROSCA A., GEONEA I, **CALAFETENU D**., **Experimental measurements of the human knee flexion angle during squat exercises,** Applied Mechanics and Materials. Vol. 823, 113-118, 2016, www.scientific.net/AMM.823.113
10. D. Calin, Daniela Tarnita, D. Popa, **D. Calafeteanu** and D.N. Tarnita **Virtual Model and Simulation of the Normal and Affected Human Hip Joint,** Vol. 823, 167-172, 2016, www.scientific.net/AMM.823.167.
11. **D. Calafeteanu**, Daniela Tarnita, D. N. Tarnita, **Numerical Simulations of 3D Model of Knee-prosthesis Assembly with Antero-posterior Tibial Slope,** IFToMM Congres, Taipei, 2015, oct, DOI Number: 10.6567/IFToMM.14TH.WC.OS1.008

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Participare la Conferințe internaționale și naționale in domeniul ortopediei și al ingineriei mecanice

- Al XIV-lea Congres National SOROT, Timișoara, oct. 2011.
- Conferinta Națională de Ortopedie și Traumatologie, SOROT 2012, Craiova, oct., 2012.
- Al XV-lea Congres Național de Ortopedie și Traumatologie, Cluj-Napoca, oct. 2013.
- Consfătuirea Ortopezilor si Traumatologilor, Calimănești-Căciulata, oct. 2013.
- Al XV-lea Congres National al Societații Române de Anatomie, Craiova, mai 2014.
- Conferinta Națională de Ortopedie și Traumatologie, Constanța, sept. 2014.
- Conferința Regională de Ortopedie și Traumatologie cu tema: Patologia gleznei și piciorului, Băile Herculane, oct. 2014.
- 3rd International Congress - SMAT 2014, Science and Management of Automotive and Transportation Engineering, oct. 2014.
- International Seminar on Biomaterials and Regenerative Medicine, BIOREMED 2015, Baile Felix, sept. 2015.
- Al XVI-lea Congres Național de Ortopedie și Traumatologie, București, oct. 2015.
- International Conference of Mechanical Engineering, ICOME 2015 Craiova, oct. 2015.
- Conferința Societății de Ortopedie și Traumatologie din Oltenia, Rânca, nov. 2015.
- Simpozionul “Tehnici Moderne în Artroplastie Primara și de Revizie”, Baile Felix, sept. 2016.
- Conferința Societății de Ortopedie și Traumatologie din Oltenia, Ediția a XIV-a, oct. 2016.

7.2. Own contributions

1. The protocol of the 6 experimental tests was established.
2. The two homogeneous samples were selected: a sample consisting of 7 healthy subjects and a sample consisting of 5 patients suffering from advanced phase knee osteoarthritis (gonarthrosis).
3. Experimental data were obtained for the 6 experimental tests established for each of the 7 subjects in the healthy sample and the 5 patients in the sample affected by gonarthrosis, before and after prosthesis, with a total of 252 data files for healthy subjects and a total of 360 data files for patients.
4. Cycle diagrames of flexion-extension movements were obtained and normalized by interpolation in the SIMIMotion software, in order to obtain average cycles; For all subjects and for all patients (before and 4 months after surgery), corresponding to each of the 6 tests, the mean sample cycles of the flexion-extension angle of the knees from both lower limbs

were obtained. For healthy subjects, the average sampled cycles were obtained for both ankles and the two hips, which are needed to simulate the virtual mannequin.

5. For all tests, the maximum values of flexion-extension angles of the right and left knees were obtained and synthetized in tabular form for each subject and for each patient.
6. The charts of ground reaction forces for the 3 walking tests and for stair ascending and descending tests were obtained for each subject and for each patient; By normalization and interpolation in SIMIMotion, the mean sample-level cycles of the reaction forces were obtained for each of the two samples.
7. 3D parametrized modeling of a virtual mannequin in Solidworks based on mean anthropometric data of the sample of healthy subjects.
8. Determination, by interpolation of the collected experimental data, of the motion laws for the six joints (ankle, knee and hip) of the two lower limbs for normal walking and for ascending stairs.
9. Simulation of walking of the virtual mannequin in ADAMS multibody simulation environment in several situations: normal walking, ascending and descending on stairs;
10. Obtaining the variation laws of ground reaction forces function of time and the reaction forces and torques of the lower limb joints of the virtual mannequin.
11. Elaboration of the virtual 3D model of existing knee prosthesis, often used in orthopedics, using the ANSYS parameterized modeling environment.
12. Elaboration of twelve 3D virtual models of knee joint-prosthesis assembly, including the following components: femur, tibia and the three components of the prosthesis: femoral component, tibial component, and polyethylene insert. Each virtual model corresponds to a prosthetic knee joint assembly with anteroposterior tibial inclination angle equal to 0^o, with anteroposterior tibial inclination angle of 5^o, respectively, and different angles between the tibia and the femur, measured in frontal plane between the femoral mechanical axis and the tibial mechanical axis: : 176°, 179°, 182°, 185°, 188°, 191°.
13. Achievement of the separate discretization of the components of knee joint-prosthesis assembly for all 12 virtual models developed and obtaining a number of 12 node-element networks, the differences being determined by different case-by-case resections and different femoral inclinations, in accordance with the actual, clinical, appropriate cases. The discretization of geometric patterns in the network of nodes and elements was made using Solid186 hexahedral and Solid 187 tetrahedral elements.
14. A number of 24 static analysis with finite elements (12 analyses for a load of 800 N and 12 analyses for a load of 2400 N) were run.
15. For all 24 cases analyzed with MEF in ANSYS, the stress maps and the displacement maps were obtained for each of the three components of the prosthetic knee joint assembly.
16. The comparative charts of the maximum stresses in the prosthesis components were drawn and analyzed for the 24 studied cases.
17. The 3D virtual optimized model of prosthetic knee was developed.
18. A number of 18 models for the joint-prosthesis assembly proposed were developed: 6 geometric models for the proposed prosthesis – knee joint assembly (*varus* of 176^o, 179^o, 182^o, 185^o, 188^o, 191^o), 6 models with anteroposterior inclination of 3^o and 6 models with anteroposterior inclination of 5^o for the same set of 6 *varus* angles, respectively.
19. A nonlinear static analysis (nonlinearity of contact) was performed for each geometric model. The analysis conditions in which the simulations took place, limit conditions and material characteristics, as well as the realization of a network of nodes and good quality elements for each of 18 individual models are presented.
20. Running a total of 30 static analyses with finite elements on all of the models presented above under the application of a load of 800 N (cases without anteroposterior inclination, cases with 5° anteroposterior inclination, respectively) and 2400N for all developed models.

21. Tabular synthesis of all extreme values of stresses and displacements obtained in the 30 distinct cases for 800N and 2400N forces.
22. Obtaining the total and lateral (X-axis) displacement maps for the whole assembly, as well as the maximum equivalent stresses (von Mises) for the components of the proposed prosthesis - knee joint assembly: femur, tibia, femoral prosthesis, tibial prosthesis, and polyethylene insertion for all the 30 analyses performed.
23. Determination of the stages of surgical prosthesis implantation technique
24. Performance of experimental corpse study of the influence of proposed prosthesis geometry
25. Obtaining the 3D physical prototype of prosthesis proposed by Rapid Prototyping Technology and resuming the experiment.

7.3. Future work

Future research directions address the following:

- Dynamic analysis, using finite elements method, of the prosthetic knee joint's behavior during a complete walking cycle.
- Biomechanical studies of the wear behavior of prosthetic components for different anterior-posterior tibial slopes.
- Biomechanical studies for the knee prosthesis fatigue.
- Designing a biomechanical stand of the prosthetic knee joint for the experimental study of its flexion-extension.

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