IOSUD - Universitatea Politehnica Timişoara
Școala Doctorală de Studii Inginerești

Research on the dynamic behavior of external amputation prostheses and HVOF-deposited layers on titanium substrates

Doctoral thesis - Abstract
for obtaining the scientific title of doctor at
Polytechnic University of Timisoara
in the field of PhD in Mechanical Engineering

author
prof.Mihai Alexandru **SZABO**

Scientific leaders: Prof.univ.dr.ing.**IlareBordeaşu**
Prof.univ.dr.ing**Ion Mitelea**

Timişoara

Month 01 year 2022

Content

1. CURRENT STATE OF THE RESEARCH OF PROSTHESES USED BY PEOPLE WITH LOCOMOTOR DISABILITIES

People with disabilities have always had a certain disadvantage, both socially and economically [76], which often affects them mentally.

Practicing sports gives people with disabilities a purpose in life, they want, through physical activities, to overcome their condition, thus being more motivated than people without disabilities.

Amputated limb prosthesis has a large share in the medical field being an independent part combining elements of biomechanics and therapy. The development of an orthopedic prosthesis involves the collaboration of the surgeon with the orthopedist.

The basic requirements that a prosthesis must have are [1]:

● to be as compliant as possible with the abutment, which must be strong, well vascularized, warm and with toned muscles, so as not to cause discomfort;

● not to obstruct the natural circulation of the blood and to maintain the integrity of the abutment.

● its functionality to help the reintegration of the individual into society

● be aesthetic and light;

● be financially accessible.

In most cases, there are people with prostheses for the thigh and leg (known as segmental prostheses). The main components of these prostheses are: the thigh sleeve, the leg sleeve, the artificial leg and the joint systems. The thigh prosthesis is characterized by the normal length abutment and is with lockable or free knees and with support on the ischium. Moving with it during the support phase is similar to normal. The only exception is the biphasic phase of the knee. The leg prosthesis (fig.1.2) has a short abutment and has specific support that is made on the patellar tendon or tibial tuberosities.

Fig. 1.2 Leg prosthesis [1]

Regardless of the activity or sport that people with disabilities choose to practice, orthopedic prostheses can help them to do it successfully. Orthopedic prostheses adapted to sports can be found in all sports, whether it is shooting, football, athletics, skiing or swimming [151, 152]. Thus, the manufacture of orthopedic prostheses from appropriate materials and with a specialized design for sports leads to the transformation of the lives of people with disabilities for the better [21].

Depending on the sport they want to practice, people with disabilities use orthopedic prostheses specific to the type of physical effort.

The image below, fig.1.8, shows a planar model containing twelve segments of a human body. The dynamics of human movements is directly influenced by the reaction forces in the joints fig.1.9. The approximate determination of the values of these forces of stress on the joints during the movements, lately is done through numerical simulation, as a result of specialized software [14].

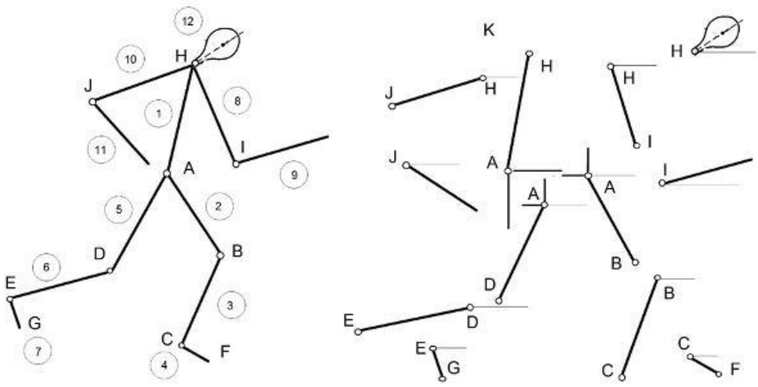


Fig. 1.8 Planar model of the human body used in the study of motion

There are many cases when people, and especially those who do sports, are subjected to difficult operations on the hip, the largest joint of the human body. The cause may be different, due to osteoarthritis (hip osteoarthritis), most often from accidents. Regardless of the cause, prosthesis surgery is required [31].

From the point of view of composition, the hip consists of a concave cavity at the level of the pelvis (acetabulum) and the upper part of the thigh bone (femur), fig. 1.21.a. The bony surface of the hip joint is covered by the articular cartilage (a smooth, shiny layer) that protects the bones and allows movement [31].

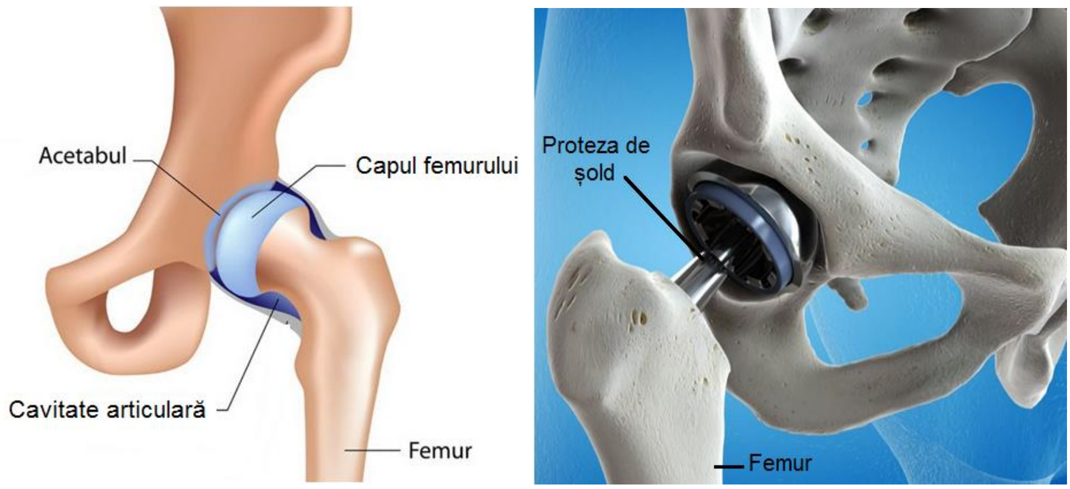


Fig.1.21 Hip-pelvic joint composition (a) and replacement with a hip prosthesis (b) (processing after [115, 116])

Conclusions

The conclusions that emerge from the study are:

- the study of prostheses for people with locomotor disabilities, as a geometric shape, dimensions and manufacturing materials, remains a topical issue, due to the diversity of activities carried out by people with locomotor disabilities, especially those who are engaged in performance sports movements;
- with the development of biomechanics as a science, methods of analysis and experimental study devices were developed;
- the materials used in the manufacture of the components of orthopedic prostheses are of a wide variety, their choice being dictated by ensuring an easy functionality for a long time and without affecting the physical condition of the user or the connecting parts of the prosthesis;
- in order to obtain efficient results in the practical application, especially for prostheses used in sports movement, experimental tests on each type are required, correlated with the numerical analysis to determine the stresses (stresses and strains, force values and moments);
- hip prostheses are very diverse. The choice of type is dictated by the doctor depending on age, daily activity, financial strength of the person and the possibility of repairing the prosthesis in case of wear;
- the prosthesis of the hip joint ensures a movement very close to the one before the appearance of the defect.

The objectives of the doctoral thesis

Based on the documentary study, the objectives that are solved in this doctoral thesis are:

1. Experimental testing of a leg prosthesis using the ZEBRIS platform.
2. Assessment of the state of demand with specific methods of numerical analysis.
3. Performing static and dynamic stress tests on carbon fiber composite materials used to make the abutment.
4. Development and numerical modeling of a composite orthopedic prosthesis model.
5. Reconditioning with a suitable technology of a worn hip prosthesis, made of biocompatible titanium-based alloy.

2. STRUCTURE AND PROPERTIES OF MATERIALS USED TO OBTAIN EXTERNAL AMPUTATION PROSTHESIS

The functions that orthopedic prostheses or artificial organs must perform determine the type of material chosen in their manufacture. Therefore, the materials differ from an athlete's orthopedic prosthesis to an everyday orthopedic prosthesis, or to a golfer's - for example. The demands on the orthopedic prosthesis are still a factor to consider when choosing materials. The orthopedic prosthesis of a speed runner must be harder and more resistant to shock, and because the athletes bend in front of the prosthesis, the prostheses must be light and provide balance [34].

External amputation prostheses for the lower limbs consist of the same elements presented in chapter 1, with some features that I detail below [18]:

- the blunt sleeve;
- prosthesis sleeves that take on the role of the skin of the amputated limb segment;
- the skeleton of the prosthesis with the role of supporting the human body;
- articular elements;
- other elements that take over the role of muscles and nerves: springs, elastic straps, hydraulic mechanisms, pneumatic, with biocurrents, etc. [14].

Composite materials, according to [85], are considered "new materials" and are designed in particular to meet special requirements in terms of:

- mechanical strength and rigidity;
- corrosion resistance;
- resistance to the action of chemical agents;
- low weight;
- dimensional stability;
- resistance to variable stresses, shock and wear;
- isolation and aesthetic properties;

The physical-elastic and mechanical characteristics of the composite material used in the manufacture of prostheses can be estimated from the characteristics of each of the constituents (mixing rule) [13].

The following sizes can be defined for a sheet [87]:

- *”mass percentage of fibers*, M_f, as the ratio of the mass of the fibers contained in a defined volume of composite material to the total mass of the same volume;
- *”the mass percentage of the matrix*: $M_m = 1 - M_f$”;
- *”volume percentage of fibers*, V_f, as the ratio between the volume of fibers contained in a defined volume and that volume”;
- *” volume percentage of the matrix*: $V_m = 1 - V_f$”;
- *” mass of fibers per unit area*, m_{0f} (kg/m^2)”.

Conclusions

• The choice of materials from which the components of the orthopedic prosthesis are made depend on the degree of disability and the activity of the disabled person (only for walking or for a certain form of sport practiced). They must have elasticity and resistance to mechanical stresses determined by the type of gait. These can be: metal alloys, carbon fiber, glass fiber, ceramic and polymer composites.

• For the prosthesis rod, (the most requested component) are used mainly metal alloys (stainless steels, aluminum-based alloys), carbon fiber or glass.

• For the fixing parts to the rest of the foot (the non-amputated part) and for the support part (which replaces the sole) are used: light metal alloys, composites based on polymer mixtures, various forms of composite materials.

• Rapid prototyping (3D printing) is the latest technology used successfully in the manufacture of prosthesis components.

3. EXPERIMENTAL STUDY USING THE ZEBRIS MOVEMENT ANALYSIS SYSTEM

Walking, in general, significantly influences the quality of life of a patient, which can be assessed anamnestically and by observation by a specialized medical staff. Quantifying faults or abnormalities while driving is still quite difficult to quantify. The gait itself consists of several phases, and a correct assessment requires specialized equipment and, more importantly, knowledge and experience in interpreting the data. Next, I will briefly present the description of normal gait, which I will analyze in a study, having as pathology the amputation of the lower limb [1].

To perform the tests, using the Zebris movement analysis system, a 60-year-old former performance athlete was selected with outstanding results in volleyball, football and athletics, who, in 2015, underwent amputation surgery at transtibial level.

Prosthesis 1. Leg prosthesis - modular with vacuum, fig.3.1

The leg of the prosthesis is flexible, made of elastic material with wooden reinforcement, the ankle joint offers a system for adjusting the stiffness of the flexure, and the modular components and tubing are made of stainless steel. The inner sleeve is made of silicone, and the outer sleeve is made of plastic, the fixation of the amputated foot being made through the vacuum valve and the silicone knee pad.

a b c

Fig.3.1 Leg prosthesis - modular, vacuum
a. overview
b. knee
c. silicone inner sleeve

Prosthesis 2. Leg prosthesis - modular with internal silicone sleeve, fig.3.2

Leg prosthesis - modular with internal silicone sleeve is equipped with a flexible leg made of elastic material, reinforced with carbon fiber, the joint is mobile and the modular components and tubing are made of titanium. The inner sleeve is made of silicone, ensuring increased comfort, and the outer sleeve being artificial resin mimics the lost limb.

Fig.3.2 Leg prosthesis - modular with internal silicone sleeve

The analyzes were performed based on the results provided by Zebris' data acquisition systems, as follows:

- the Zebris CMS-HS system was used for the analysis of gait, which uses ultrasound to determine the position of the sensors necessary to establish the time elapsed from the emission to the reception of the ultrasonic pulse;
- the Zebris FDM capacitive system was used to determine the plantar distribution, which is intended to measure the plantar distribution, both static and dynamic.

Zebris FDM system

The Zebris FDM platform allows a quick gait assessment, being a capacitive system for determining the distribution of plantar pressures (forces) in dynamics. The measurement is done by a pre-calibration and then the actual measurement is performed.

Determining the plantar distribution with the FDM platform is a non-invasive method, which uses 1 cm2 capacitive sensors, distributed over the entire surface of the platform. The platform connects to a computer via a USB port and can measure forces with values between 1 ÷ 120 N / cm2.

To analyze the gait, the investigated subject must cover a distance of about 1.5 m between the two ultrasound transmitters. This distance can be covered on the ground or on a platform that records the plantar pressure.

Zebris CMS-HS system

The system allows a quick assessment of gait. It starts with the positioning of 4 ultrasound receivers, 3 microphones each: 2 on the outer side of the thighs and 2 on the outer side of the legs, which receive the ultrasound emitted by the two systems consisting of 3 speakers placed on either side of the direction of travel. By measuring the time required for ultrasound to reach from the transmitter to the receiver and based on the triangulation principle, the position and orientation of the sensors attached to the subject can be determined [131].

The static balance represents the oscillation of the pressure center of the foot on the capacitive platform in relation to the support surface, which represents the footprints of the two feet on it. In order to make the determinations, the subject must stand still on the platform surface for 20 seconds, as seen in the image in fig.3.9. This procedure was performed with both prostheses mentioned above.

Fig. 3.9 Determination of the oscillation of the center of pressure on the capacitive platform (equilibrium)

Walking plan, angles and plantar pressures

The recording of the parameters made during the walk by the subject, of the angles of the joints, fig.3.10, is made with the space system of sensors that are attached to it, through a number of successive passages in front of the ultrasound transmitters. For a more accurate determination, a set of passes was made with each prosthesis.

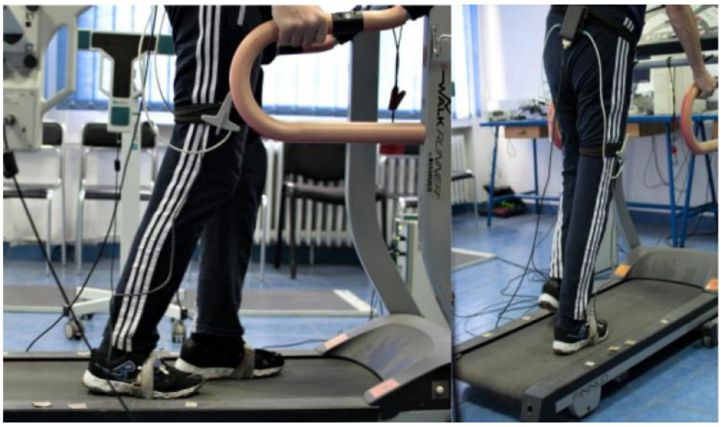


Fig.3.10a Adjusting joint angles when walking

Fig. 3.10b Adjusting joint angles when walking comparison between the two prostheses

The human body has a center of gravity positioned approximately anteriorly by the dorsal vertebra T10, which leads to the observation that it is at an appreciable distance from the ground. The traditional composition of the human body consists of two main categories [1]: the upper part "passenger" and the lower part "locomotive", and the intersection of these is made at the hips. The main functions of the musculoskeletal system are propulsion, maintaining balance in the stance phase, absorbing shocks and, last but not least, conserving energy [77].

The results obtained for the plantar reaction of the subject in the orthostatic position.

Prosthesis 1. Leg prosthesis - modular with internal silicone sleeve

Test 1, fig.3.11 and 3.12

The following can be seen in this test:

- the variation of the plantar reaction forces, fig.3.11, for the prosthetic leg (left) is more accentuated on the back of the drip, the heel, fig. 3.12a;
- the variation of the plantar reaction forces, fig.3.11, for the normal leg (right), fig.3.12b, is more accentuated in the metatarsal part, varying with the passage of time;
- the subject tends to lean on the normal (right) leg, fig. 3.12b;
- the center of gravity of the body is very slightly forward.

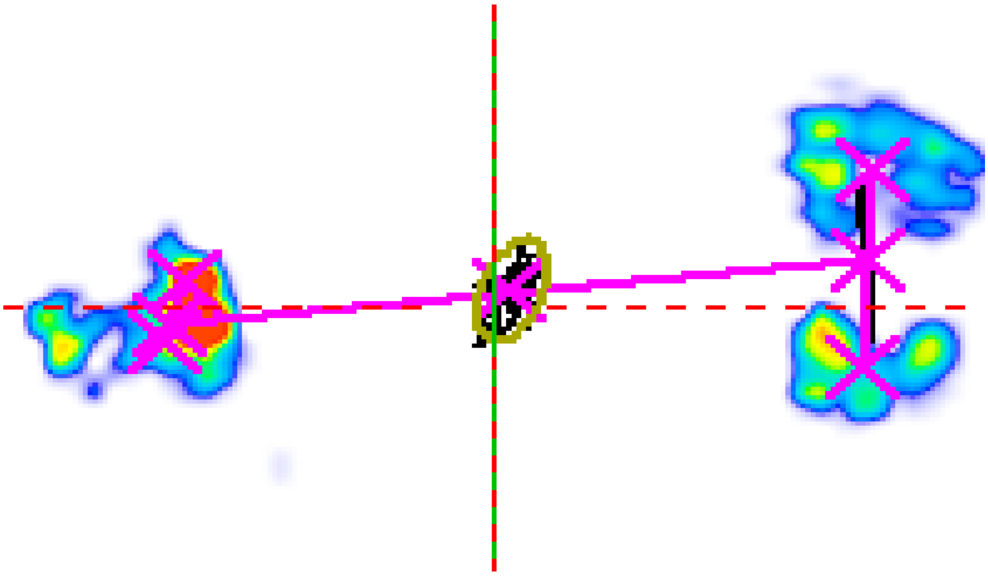
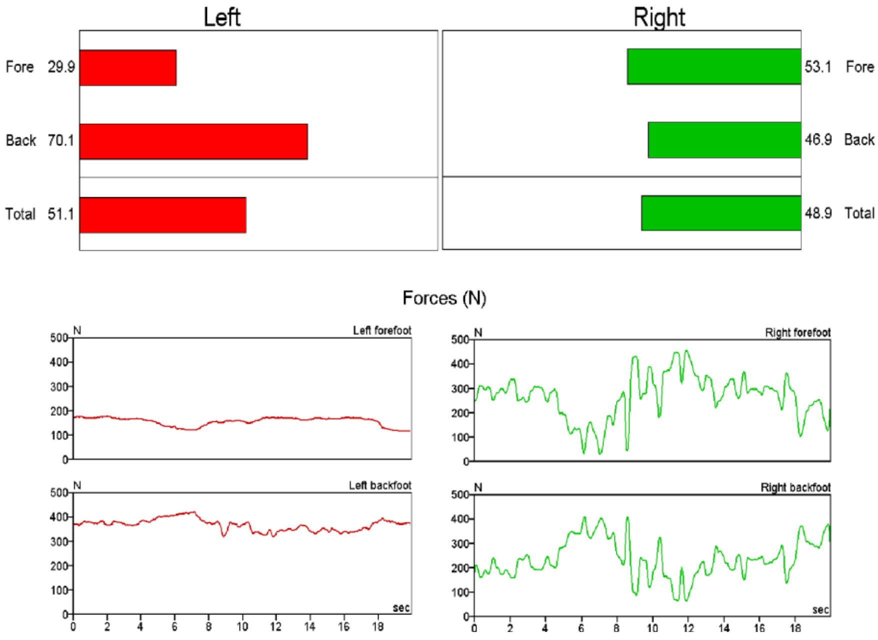


Fig.3.11 Variation of plantar reaction forces



a) b)

Fig.3.12 Force variation diagrams

Prosthesis 2. Leg prosthesis - vacuum modular

Test 1, fig.3.15 and 3.16

After performing the test, the following were observed:

- the variation of the plantar reaction forces, fig.3.15, for the foot with prosthesis (left), fig.3.16a, is more accentuated on the front part of the foot, the area of the metatarsals being approximately constant over the whole duration. It can also be seen that the subject uses only the front area of the foot, and the heel area is not used at all;
- the variation of the plantar reaction forces, fig.3.15, for the normal leg (right), fig.3.16b, is more accentuated in the metatarsal part, having the tendency to become constant with the passage of time;
- the subject tends to lean on the normal (right) leg, fig.3.16b;
- the center of gravity tends to move forward, with a fairly pronounced displacement value.

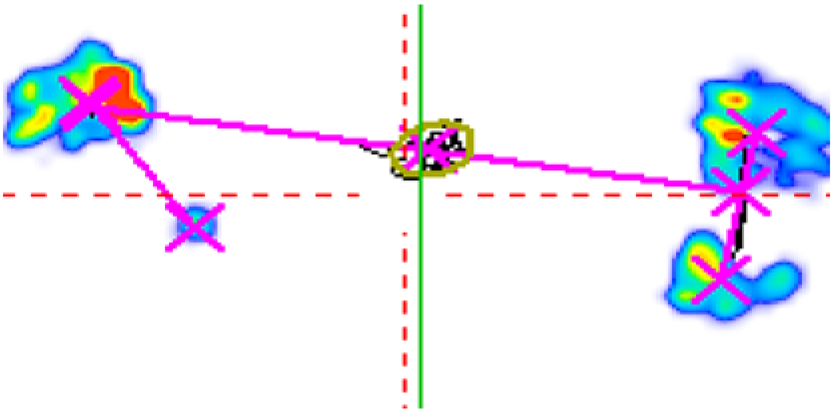
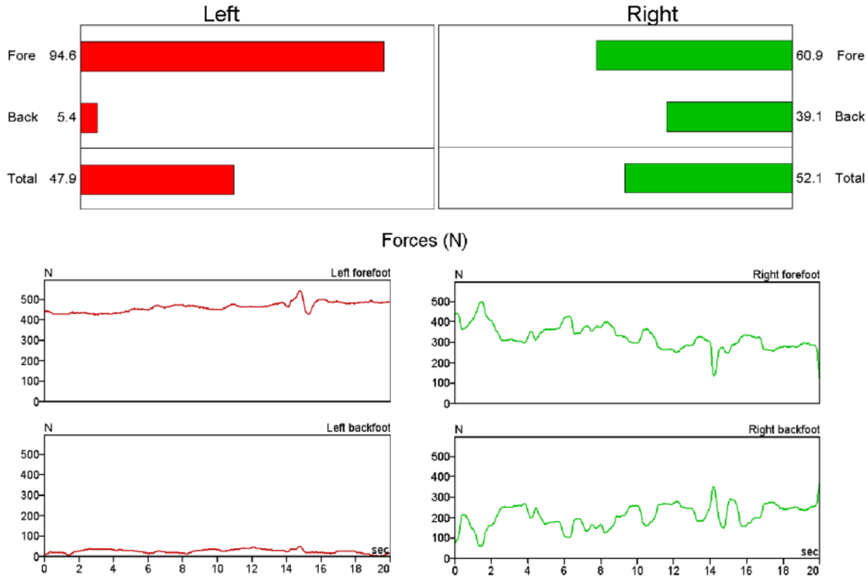


Fig.3.15 Variation of plantar reaction forces



a) *b)*

Fig.3.16 Plantar reaction force values (Variation diagrams)

GRF plantar reaction force profile in case of dynamic analysis

As it is known, while walking the body advances, one leg serves as a support and the other is in the phase of balance or oscillation, to reach the new position where a new phase of support will begin, after which the cycle resumes. There are two moments when walking when both feet are on the ground.

Prosthesis 1. Leg prosthesis - modular with internal silicone sleeve

On the shape of the plantar footprints, taken during the passage over the platform, represented in the figure fig.3.19, an apparently normal gait is observed. The solid black line represents the path of the center of pressure (center of mass of the body). It is observed that this line, in the case of the normal foot (right), is approximately flowing (smooth), angular, but inconsistent and steep under the prosthetic foot (left). This is due to the fact that the joint of the artificial ankle has an improper rolling during the period of support while walking (fig.3.19a).

The graphs below show that the subject has an approximately equal pitch opening for both legs, with an approximately equal time on each step. In the case of the support phase, there is a longer duration of support on the right leg, the healthy one, and in the case of the balance phase, it speeds up the healthy leg. When using the left foot, during the contact phase with the ground, the subject has a tendency to hurry, to reduce the contact time. This finding shows that the subject either has low confidence in the prosthetic leg (left) or that the prosthesis has a deficiency in the ankle area.

The stiff sole is highlighted by longer contact lines and the trembling of the normal foot is probably caused by proprioception (“Proprioception refers to the orientation of the body and our movements in space, the ability of muscles to contract and relax to stabilize body depending on the given situation ”) inaccurate determined by the presence of the foot with prosthesis (left), fig.3.19a, 3.20a, 3.21 and 3.22.

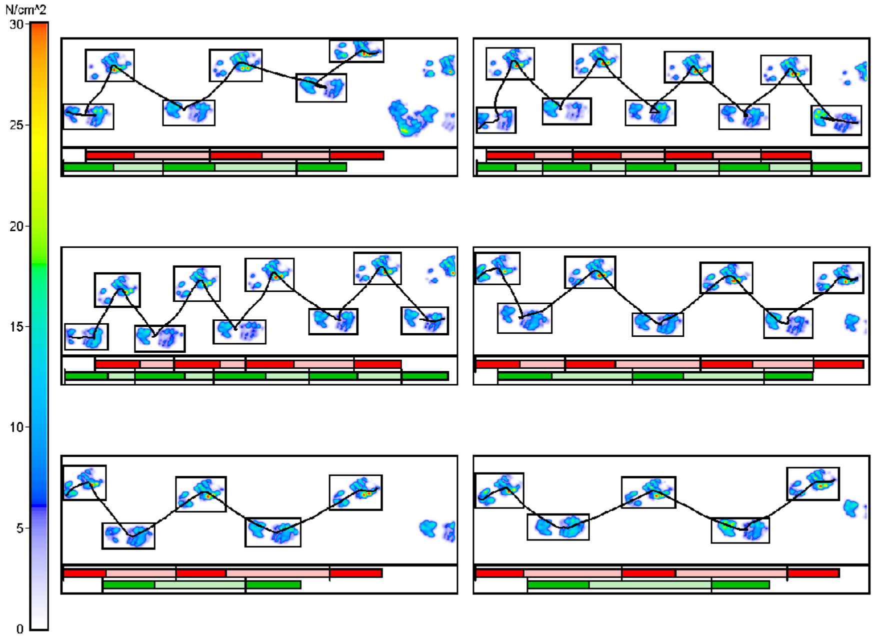
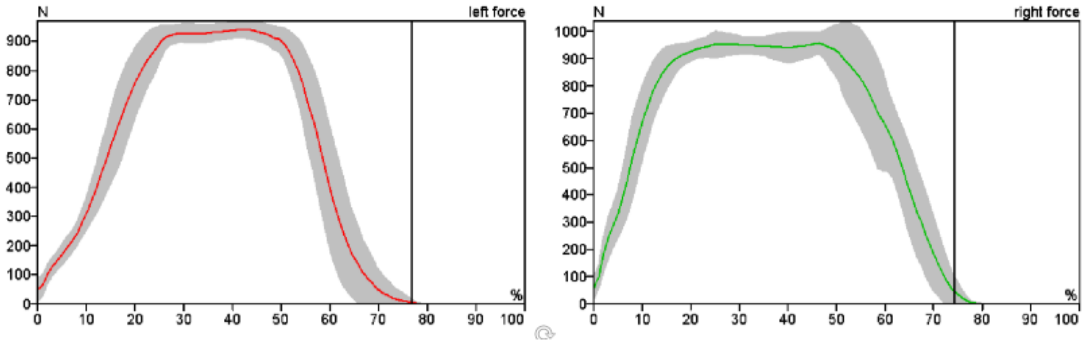
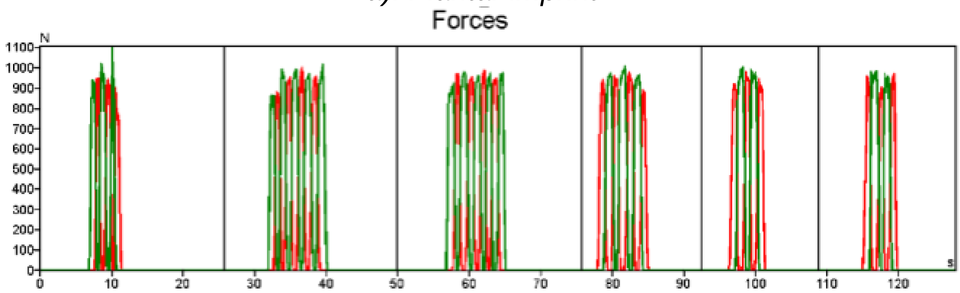


Fig.3.19 The shape of the footprints when walking on the platform



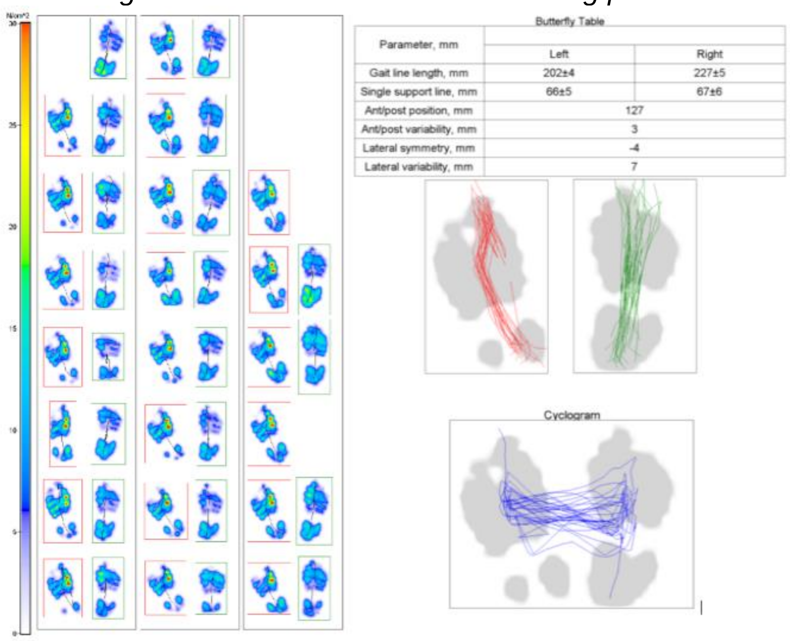
a). Plantar imprint



b). Graph of forces
Fig.3.20 Plantar reaction forces determined by the platform

	left	right
Step time, sec	0.85	0.88
Swing time, %	23.21	25.69
Stance time, %	76.79	74.31
Load response, %	25.55	25.69
Pre-swing, %	25.69	25.55
Single support, %	25.55	23.07
Step length, cm	27	32
Normalized	-	-
Stride length, cm	58	
Normalized	-	
Stride time, sec	1.73	
Cadence, st/min	35	
Velocity, cm/sec	34	
Normalized, 1/sec	-	
Variability of velocity, %	12.67	
Leg length, cm	-	

Fig.3.21 Measurements on the walking phases



Parameter, mm	Left	Right
Gait line length, mm	202±4	227±5
Single support line, mm	66±5	67±6
Ant/post position, mm	127	
Ant/post variability, mm	3	
Lateral symmetry, mm	-4	
Lateral variability, mm	7	

a) b)

Fig.3.22 The center of pressure on the footprints in colored papers (a) and the synthetic numerical table and the butterfly cyclogram (b)

Prosthesis 2. Leg prosthesis - vacuum modular

In the form of plantar imprints, taken during the passage over the capacitive platform, represented in fig. 3.23 a much faster gait is observed than with the previous prosthesis. The black line, which represents the path of the center of pressure (center of mass of the body), in the case of the normal foot (right), is no longer flowing, and under the left foot (with prosthesis) appears a more pronounced angle. This is due to the fact that the subject tends to step on the tip of the prosthesis, as observed in the test on the balance of the center of gravity (fig.3.11-3.12).

From the graphs below (fig.3.23 ..... 3.26) it is observed that the subject has a much smaller pitch opening in the case of the prosthesis member, with a longer duration of support on it, and in the case of the phase The subject uses the healthy foot (right) as a support. As a result, a longer balance period occurs.

These findings show that the subject has very little confidence in the prosthetic foot, but that the prosthesis has a better efficiency. After dispersing the plantar reaction, it is observed that the subject has hesitations in contact with the ground of the prosthesis limb.

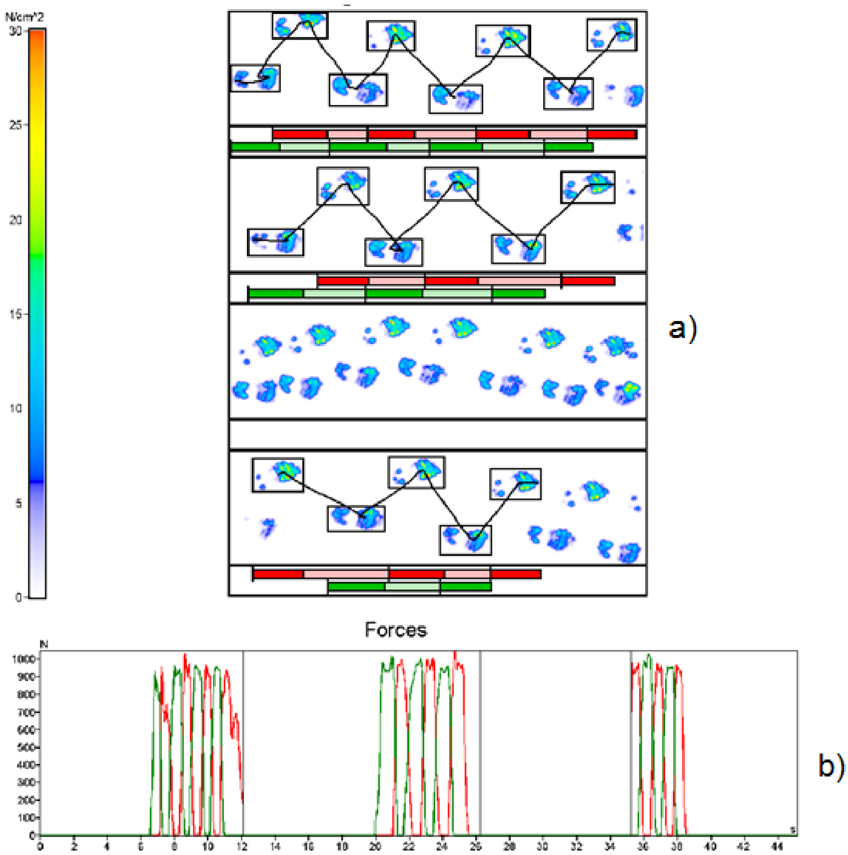


Fig.3.23 The shape of the footprints when walking on the platform a). Plantar imprint, b). Graph of forces

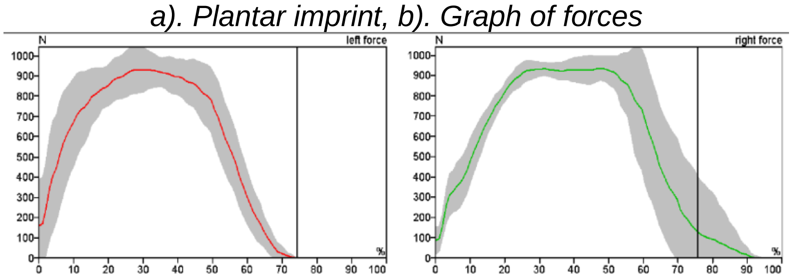
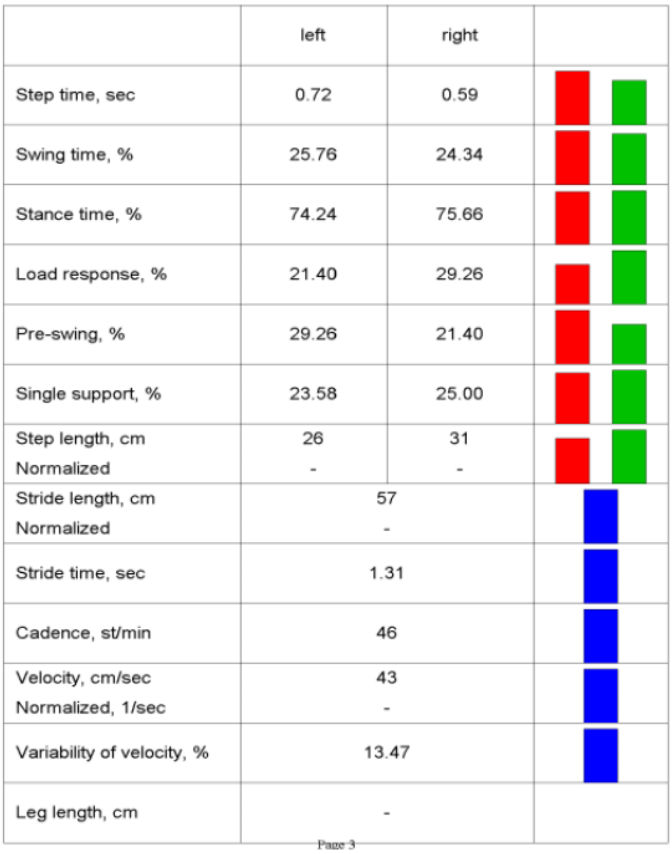


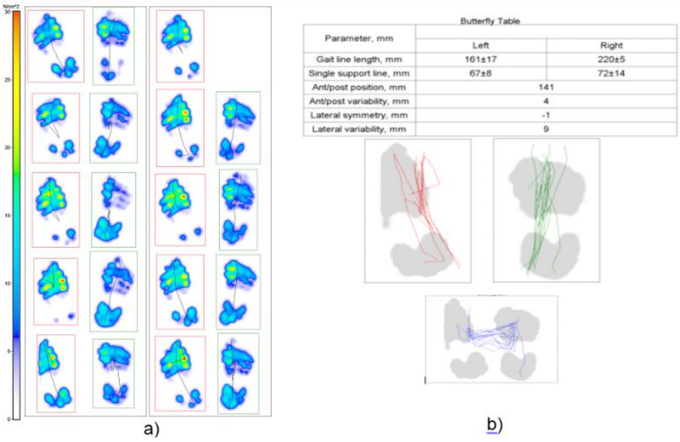
Fig.3.24 Plantar reaction forces determined on the platform



	left	right
Step time, sec	0.72	0.59
Swing time, %	25.76	24.34
Stance time, %	74.24	75.66
Load response, %	21.40	29.26
Pre-swing, %	29.26	21.40
Single support, %	23.58	25.00
Step length, cm Normalized	26 -	31 -
Stride length, cm Normalized	57 -	
Stride time, sec	1.31	
Cadence, st/min	46	
Velocity, cm/sec Normalized, 1/sec	43 -	
Variability of velocity, %	13.47	
Leg length, cm	-	

Page 3

Fig. 3.25 Measurements on the walking phases



Butterfly Table

Parameter, mm	Left	Right
Gait line length, mm	161±17	220±5
Single support line, mm	67±8	72±14
Ant/post position, mm	141	
Ant/post variability, mm	4	
Lateral symmetry, mm	-1	
Lateral variability, mm	9	

a) b)

Fig. 3.26 The center of pressure on the footprints
in colored maps (a) and synthetic numerical table with butterfly cyclogram (b)

Comparing the average values of the angle measurements, for the essential areas between the new prosthesis and a normal gait, the graphs from fig.3.38-3.40 were constructed.

For the hip (fig.3.38), it is observed that the prosthesis imposes a wider flexion movement for the left limb (prosthesis) and, at the same time, a reduction of the amplitude for the healthy limb (right).

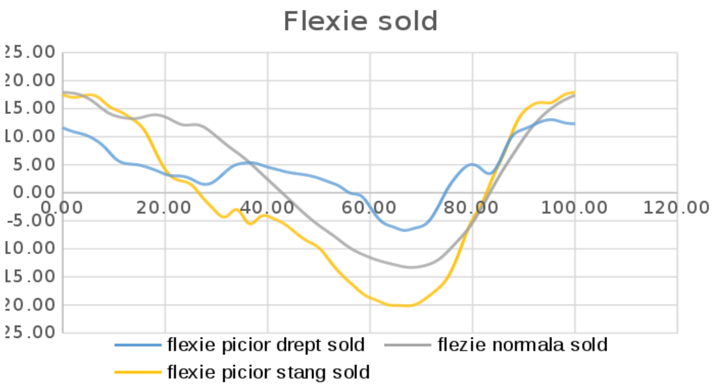


Fig.3.38 Hip flexion comparison

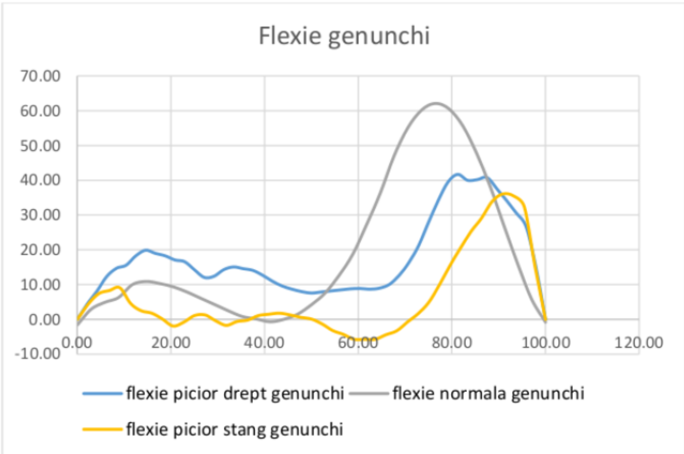


Fig.3.39 Knee flexion comparison

In the case of flexion at the knee joint, fig.3.39, there is a decrease in amplitude compared to a normal movement, both in the case of the left limb (prosthesis) and in the case of the healthy one (right). Also, in the case of a prosthetic foot, the amplitude is lower than in the case of a healthy one (right foot). At the same time, there is a delay in reaching the maximum amplitude.

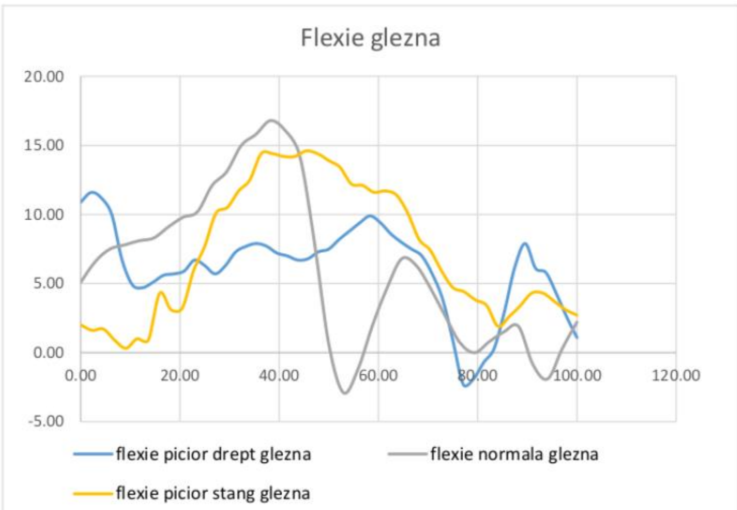


Fig.3.40 Ankle flexion comparison

In the case of the ankle joint, fig.3.40, significant changes are observed for the angular variation in the case of flexion compared to normal. This is because the prosthesis was not fitted with an ankle joint and this causes a change in the kinematic behavior of the ankle joint in the healthy leg.

Conclusions

Both prostheses provide stability, the second prosthesis (leg prosthesis - vacuum modular) is a little more stable but increases the pressure on the metatarsal area.

Comparison of energy consumption

There are significant differences between the values of the parameters that characterize the movement of the subject with the two prostheses, regardless of which one is taken as reference. These differences are the expression of the quality of the prosthesis, the geometric-constructive shape, but also the habit of the subject with the prosthesis.

Symmetry of gait

- the angular variation in the knee joint is an essential geometric quantity in determining the proper opening of the prosthesis;
- the angular variation in the hip joint is decisive for a smooth, linear gait so that the person with a locomotor disability, in this case the one used for the experiment, is not disturbed while walking and does not suffer diseases in other areas of the foot or body;
- a wider flexion movement is required for the left limb (with prosthesis) and a reduction of the amplitude for the healthy limb (right);
- it is necessary to make a mobile joint at the ankle in order not to change the kinematic behavior of the disabled person and not to affect the healthy limb;
- the need for a numerical analysis to certify the resistance of the prosthesis to the stresses while walking;

4. NUMERICAL ANALYSIS OF THE MECHANICAL BEHAVIOR OF THE EXTERNAL AMPUTATION PROSTHESIS USED IN THE EXPERIMENT

One of the modern and efficient methods of numerical analysis of resistance structures is the finite element method (FEM), which consists of breaking down a simple or complex mechanical or biomechanical system, made of isotropic and / or anisotropic materials, into entities called finite elements, interconnected with each other in discrete points called nodes, at which, depending on the nature of the analysis (mechanical, thermal, etc.), unknown parameters (displacements, temperatures, etc.) will be defined. The behavior of the studied structure is approximated by a discrete model of it, called calculation model, obtained after a process of assembling all the finite elements from the composition of the discretized structure.

Geometric model of external foot prosthesis

In order to study the mechanical behavior of the studied prosthesis, first its geometric model was conceived and elaborated, with the help of the SOLID WORKS program.

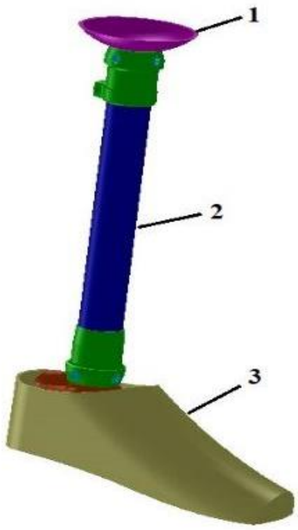


Fig. 4.1 Geometric model of the prosthesis
(1 - capsule, 2 - prosthesis rod, 3 - prosthesis leg)

As can be seen in Fig. 4.1 The geometry of the prosthesis consists of: the capsule (1), which connects the bones of the thigh and the prosthesis, the rod (2), which replaces the bones of the amputated leg, and the leg of the prosthesis (3), which replaces the bones of the leg.

Finite element analysis results

Following the numerical analysis, the desired information was obtained: displacements, specific deformations and stresses, the distribution of displacements (total deformation) being presented in fig. 4.8. The maximum value of the total deformation is **0.278 mm**, a value considered to be very small compared to the size of the prosthesis. Therefore, from this point of view, it can be said that the foot prosthesis was well designed and dimensioned.

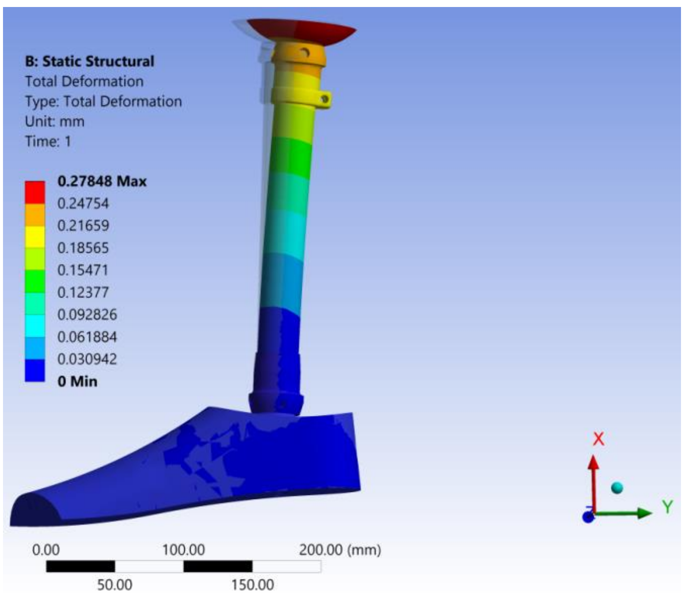


Fig. 4.8 Distribution of total deformations

Finite element analyzes also lead to the determination of the stress state in the studied structures, the distribution of equivalent stresses, calculated according to von Mises resistance theory, being presented in fig. 4.9.

The maximum equivalent stress in the prosthesis due to the application of a static load of 1000 N is **43.66 MPa**. This value appears in the capsule area. Comparing this value with the allowable strength (σa = 92 MPa) of the prosthesis material means that the prosthesis resists, not being in danger of breaking. Therefore, the prosthesis has been well designed and dimensioned, and the material from which the prosthesis was made is a suitable one.

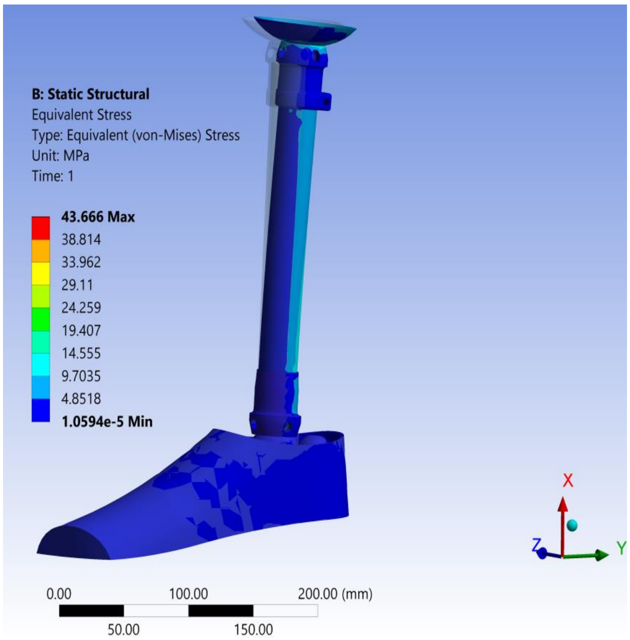


Fig. 4.9 Distribution of von Mises tensile equivalent

Fig. 4.10 shows the distribution of normal stresses in the whole prosthesis, stresses occurring during its use, the values obtained being far from the permissible limit.

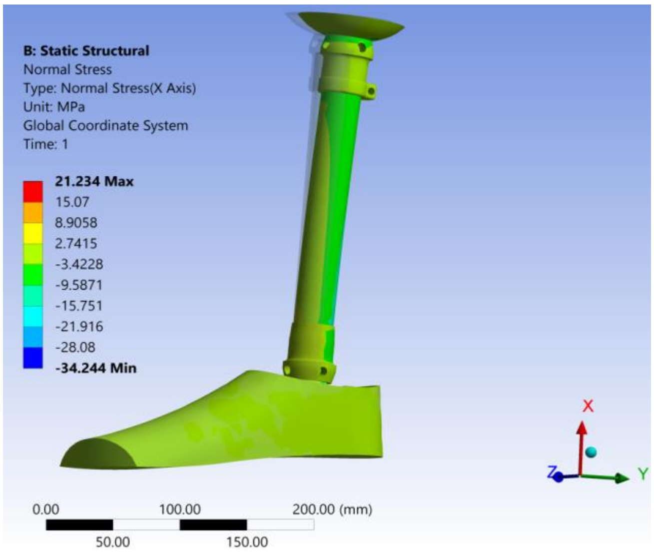


Fig. 4.10 Normal tensile distribution

Conclusions

This chapter briefly presents the basic concepts that underlie the finite element analysis of the designed prosthesis, researched experimentally in the other chapters of the thesis.

Following the conditions (restrictions) determined by the operation of the prosthesis (contacts, slips, deformations), the analysis with finite elements took into account the recommendations and specialized bibliographic indications, for the elaboration of the calculation model using three types of finite elements: SOLID 186, SOLID 187 and CONTA 174, the latter being a contact element between the components of the prosthesis.

The numerical analysis, performed by means of the finite element method of the stress state and the deformation state of the prosthesis, showed that it is not in danger of yielding during use, the deformations being also quite small.

Numerical simulation of the buckling behavior of the prosthesis shows that the orthopedic prosthesis does not present problems in terms of loss of stability**.**

5. DEVELOPMENT OF AN EXTERNAL AMPUTATION PROSTHESIS FROM COMPOSITE MATERIALS

For the development of a new foot prosthesis, a composite material reinforced with bidirectional fabric was used, given the local availability and the advantages of these materials: low weight, the possibility of being designed to optimize strength and rigidity, high fatigue resistance , corrosion resistance [94].

The investigated composite material, supplied by the manufacturer **HexCel Corporation**, has a 5H Satin type woven structure. Carbon fiber yarn (s) are commercially available under the name **HexTow® AS4**, and the matrix, an epoxy resin, is supplied under the trade name **HexPly® 8552S** [124, 126].

The high mechanical strength and high values of elasticity characteristics, as well as the affordable cost, have imposed carbon fibers as the main reinforcing material for polymer matrix composites. By reinforcing with bidirectional fabrics, the highest values of resistance and modulus of elasticity on the yarn directions (warp and weft) are obtained.

Monoaxial tensile test

The mechanical and elastic tensile characteristics of the investigated composite material were determined by experimental tests performed in the CIDUCOS Test Laboratory, Department of Mechanics and Strength of Materials, from the Polytechnic University of Timișoara [145].

The Instron 8874 universal static and dynamic testing machine was used to perform the monoaxial tensile tests (fig. 5.5).

Fig.5.5. Performing monoaxial traction tests
(Instron 8874 static and dynamic testing machine)

The tests were performed at ambient temperature, in controlled travel mode, at a constant speed equal to 1 mm / min.

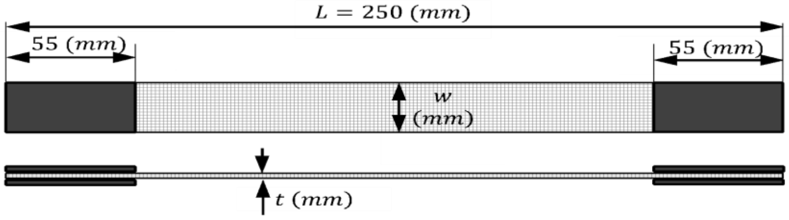
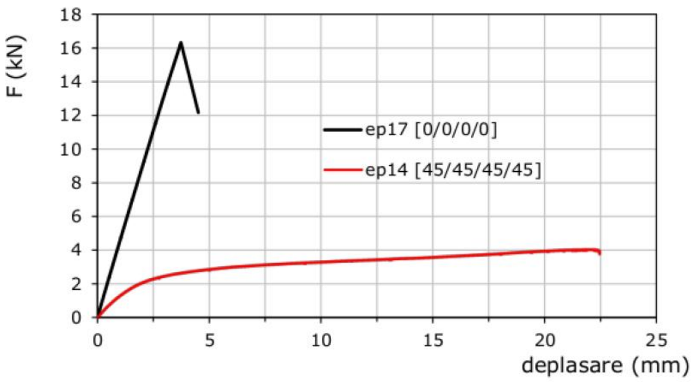


Fig.5.6. Tensile test geometry (ASTM D 3039-00



Force curves - displacement when testing monoaxial traction

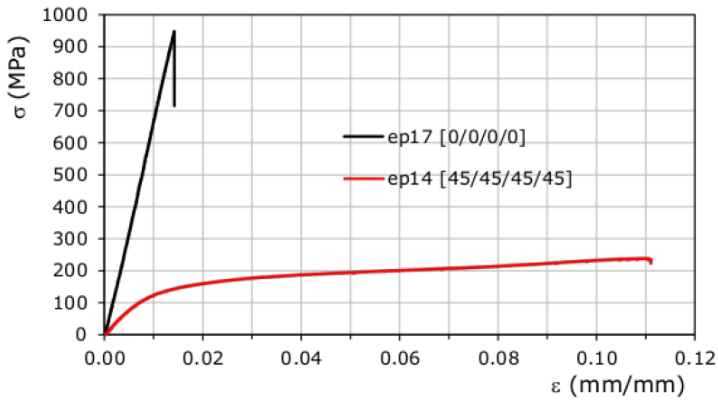


Fig.5.7. Characteristic stress curves σ - specific deformation ε
when testing monoaxial traction

*Tabel 5.1. Experimental results *) in the tensile test of the laminate [0$_4$]*

Laminat	Ep.	t (mm)	w (mm)	F_{max} (N))	σ_{1t} (MPa)	ε_{1t} (%)	E_1 (MPa)	ν_{12} (-)
[0$_4$]	ep15	1,09	15,29	15781	948	1,38	64657	0,109
	ep16	1,09	15,31	15702	939	1,25	67491	0.114
	ep17	1,11	15,48	16324	948	1,42	65974	0,095
	valoare medie				**945**	**1,35**	**66041**	**0,106**

**) σ_{1t} longitudinal tensile strength, main direction 1 of the material;*
ε_{1t} the specific tensile strength in the main direction 1 of the material;
E_1 modulus of elasticity in the main direction 1 of the material;
ν_{12} the transverse contraction coefficient in the reinforcement plane.

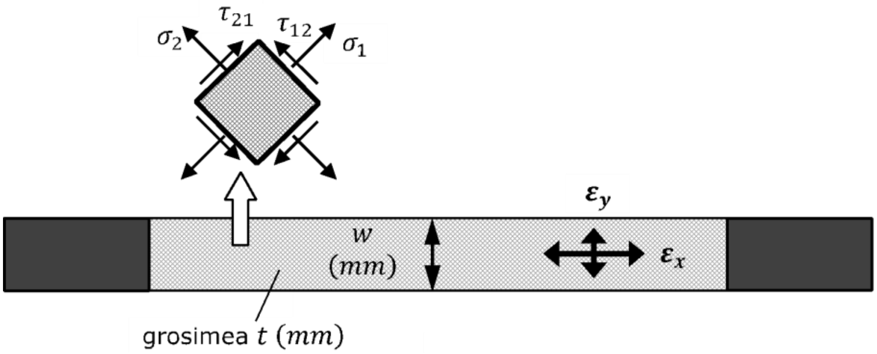


Fig.5.9. The tension state in the 45 ° oriented sheet required for traction

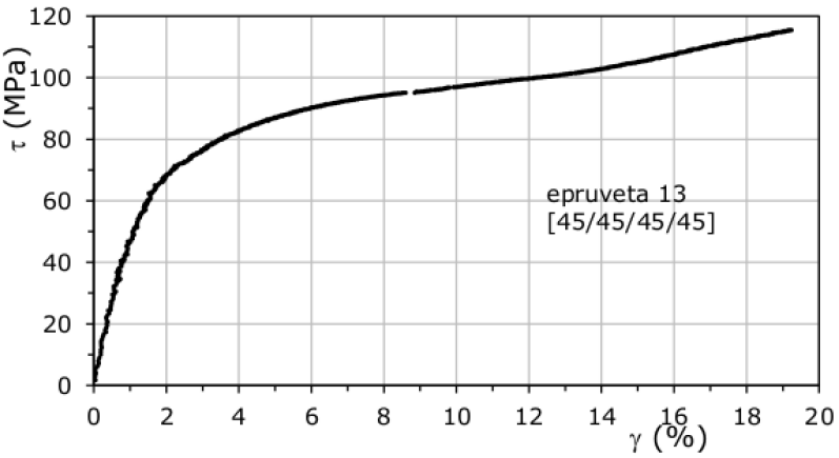


Fig.5.10. Shear stress curve - specific slip $\tau_{12} - \gamma_{12}$

*Tabel 5.2. Experimental results *) in the tensile test of the laminate [45$_4$]*

Laminat	Ep.	t (mm)	w (mm)	F_{max} (N)	σ_{xt} (MPa)	ε_{xt} (%)	E_x (MPa)	τ_{12f} (MPa)	G_{12} (MPa)
[45$_4$]	ep13	1,12	15,42	3933	231	10,82	16552	115,5	7450
	ep14	1,09	15,56	4029	238	10,98	16994	119,0	7025
	ep18	1,09	15,66	4018	235	10,48	16356	117,5	7650
	valoare medie				**235**	**10,76**	**16634**	**117**	**7375**

**) σ_{xt} tensile strength to orient the blade to 45°;*
ε_{xt} the specific elongation at the traction limit for the orientation of the lamina at 45°;
E_x modulus of elasticity in the direction of the load orientation orientation la 45°;
τ_{12f} shear strength in terms of reinforcement;
G_{12} shear modulus in the reinforcement plane.

Tabel 5.6. Estimated elastic characteristics of the studied composite material

	E_1 (MPa)	E_2 (MPa)	E_3 (MPa)	$\nu_{12} = \nu_{21}$ (-)	$\nu_{13} = \nu_{23}$ (-)	$\nu_{31} = \nu_{32}$ (-)	G_{12} (MPa)	G_{13} (MPa)	G_{23} (MPa)
MEF	71181	71121	18109	0,096	0,35	0,089	9098	3679	3679
Exp.	66041	-	-	0,106	-	-	7375	-	-
ER*	+7,7 %	-	-	-9,4%	-		+23,3%		

** relative error*

The experimentally determined mechanical strengths for the studied composite material are presented in Table 5.7. ***Given the similar elastic behavior*** $E_1 \cong E_2$ ***at monoaxial traction on the main directions 1 and 2 of the material the mechanical strengths are considered to be equal in these directions, therefore*** $\sigma_{1t} = \sigma_{2t}$. The experimental results published in the literature [29] highlight this fact.

Tabel 5.7. Mechanical strength of the composite material

	$\sigma_{1t} = \sigma_{2t}$(MPa)	$\sigma_{1c} = \sigma_{2c}$(MPa)	τ_{12f} (MPa)
Exp.	945	634	117

The design of the orthopedic prosthesis had as a starting point a geometry of a prosthesis manufactured by the Ottobock company [129] (fig. 4.28a). It consists of two composite laminates:

- upper plate in the shape of the letter L in order to obtain a flexibility similar to plantar flexion (dorsiflexion);
- horizontal lower plate in order to attenuate the forces resulting from the contact of the back of the sole of the foot (heel) with the ground.

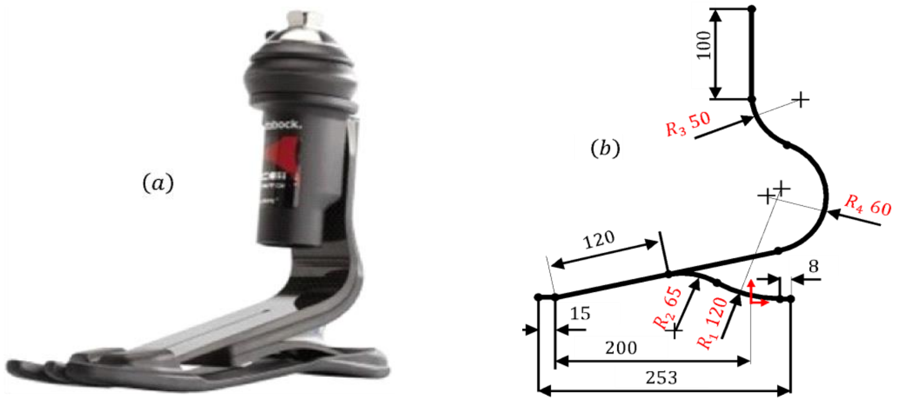


Fig. 5.28. Foot prosthesis geometry (a) Ottobock product (b) geometry proposed for optimization)

The determination of the optimal geometry of the orthopedic prosthesis had as starting point the sketch presented in fig. 5.28b to which a series of dimensional constraints and four variable radii of curvature R_i, have been established, highlighted in the drawing.

Fig. 5.38. Prosthesis modified according to the results of topological optimization

Conclusions

The analysis and processing of experimental results based on diagrams constructed by applying the specific relationships and criteria underlying the theory of stress and strain, presented in the literature on the strength of materials and the use of appropriate software, allowed the characterization of materials in terms of materials. of structure and acceptability.

The monoaxial tensile test of the symmetrical layered composite, consisting of 4 overlapping sheets oriented at 0 ° [symbol 04], in the direction of the warp, shows:

- the force-displacement curve has a linear portion until the maximum force is reached, followed by a sudden drop after reaching the maximum value;
- a linear-elastic behavior until the final failure;
- the rupture is caused by the appearance and propagation of normal cracks in the direction of the load and is completed by the release of carbon fibers, the component of the composite with specific elongation limit.

he monoaxial tensile test of the symmetrical layered composite consisting of 4 overlapping sheets oriented at 45 °, with the symbolization [454] highlights:

- the stress-strain curve shows two zones of variation: the first characterized by a linear behavior, between 0 - 0.01 mm / mm, for which the modulus of elasticity E_x is defined and the second characteristic of a behavior of the “nonlinear ductile” type, in the range 0.01 - 0.03 mm / mm;
- the rupture is caused by the appearance of microcracks in the epoxy resin matrix, as a result of the reorientation of the carbon yarns and completed by their rupture, both the warp and the weft, and the delamination between the fibers and the matrix.

For the composite with the orientation of the sheets at 0 ° (in the main direction 1 of the material), the numerical analysis of the state of stress, shows:

- the force acting on the heel has a linear variation from 0 N to 1000 N at 0.2 sec, followed by a decrease to 0 N at 0.8 sec;
- the force acting on the tip of the pin is zero to 0.4 sec and increases linearly to 850 N at 0.8 sec.
- after reaching the maximum value the force decreases linearly to zero, at the end of the step considered equal to one second

In the time intervals of 0-0.4 sec. (fig. 5.45) and 0.4-0.8 sec. (Fig. 5.46) Numerical analyzes for static stress regime show the prosthesis works, even in the most unfavorable situations, with a sufficiently high safety factor.
So:

- in the time interval 0-0.4 (sec.), in the area of connecting the sole with the upper part of the prosthesis, the main tension reaches the maximum absolute value of 306.34 MPa, and in the ankle area the value is 41.02 MPa; the maximum displacement is equal to 9,588 mm and is recorded in the heel area;
- in the time interval 0.4-0.8 (sec.) the main incision reaches the maximum absolute value of 468.77 MPa in the ankle area, in the sole area the value being equal to 196.19 MPa; the maximum displacement is recorded in the area of the pin and is equal to 77.15 mm.

6. FUNCTIONAL LAYERS DEPOSITED BY HVOF ON TITANIC SUBSTRATES FOR HIP PROSTHESES

People who have an amputation prosthesis may have osteoarthritis of the amputated leg after a certain period. Coxarthrosis is a degenerative disease of the hip joint. The articular cartilage is destroyed with the appearance of pain. The treatment of hip osteoarthritis is done by prosthesis of the hip, with different surgical techniques, using a total hip prosthesis.

A grade 1-symbolized titanium bar Ti6Al4V according to ASTM F1472, characterized by [37], [137] was used as substrate material:

- microstructure after annealing, formed by the α + β phases;
- chemical composition consisting of: 0.03% N ,, 0.1% C, 0.015% H, 0.2% Fe, 0.18% O, residue Ti;
- mechanical properties: mechanical breaking strength Rm = 900 MPa, yield strength Rp0.2 = 830 MPa, elongation at break 18% and modulus of elasticity E = 110-120 GPa.

In principle, the process consisted of passing the additive material in powder form through the plasma jet obtained [61]. Due to the strong compression of the plasma jet inside the nozzle, through thermodynamic effects of cooling and magnetic strangulation, high temperatures and high velocities of the ionized gas jet are obtained. Due to the high temperature, the powder melted and was driven by the gas jet to the surface of the prosthesis. The particles, in the plastic state, adhere to this surface.

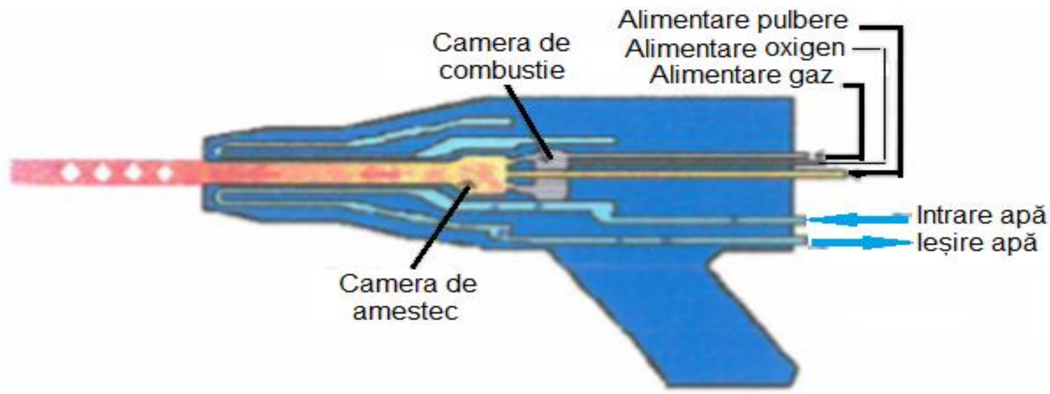
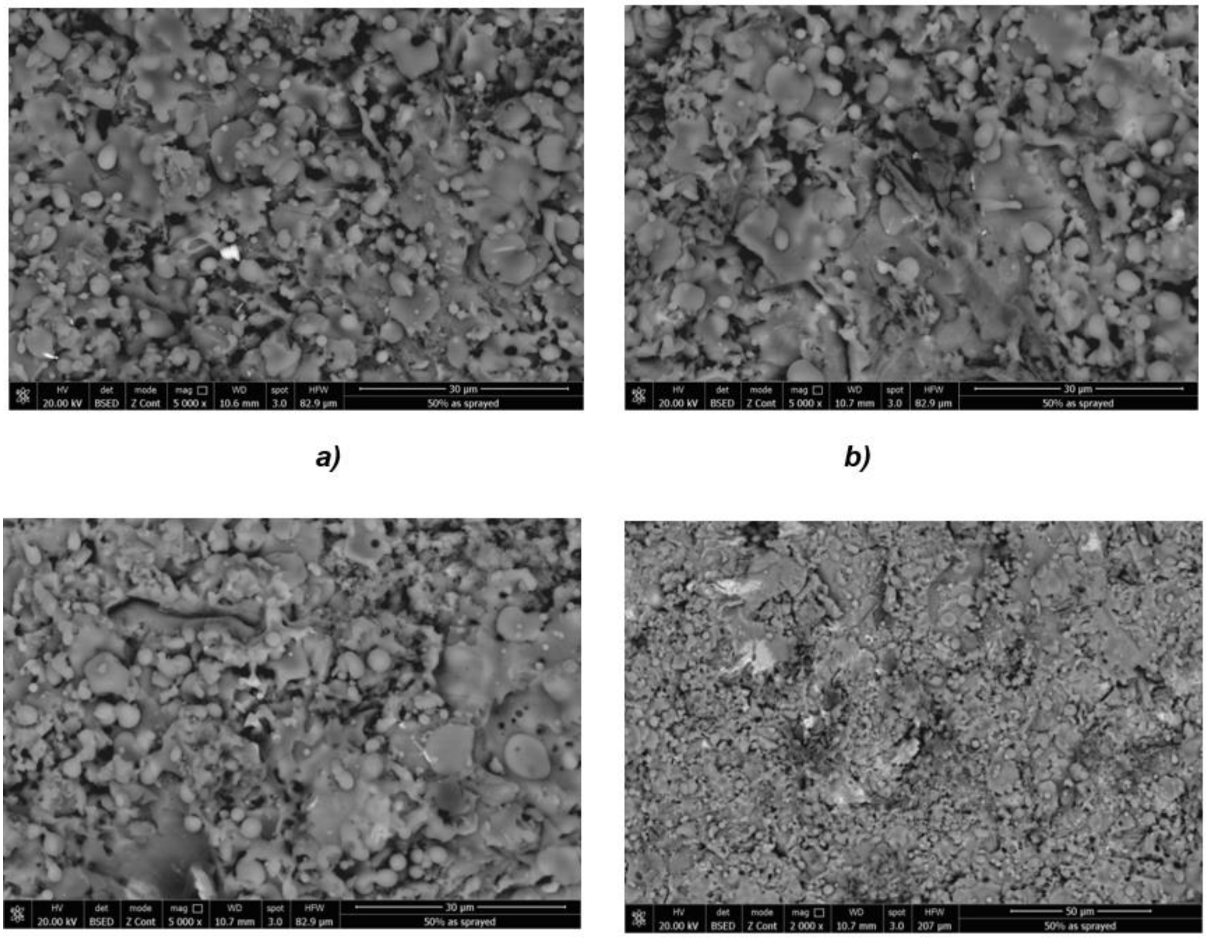


Fig.6.2 Thermal spray gun HVOF [61]

The spray parameters were: 430 l / min oxygen, 60 l / min propane and 470 l /

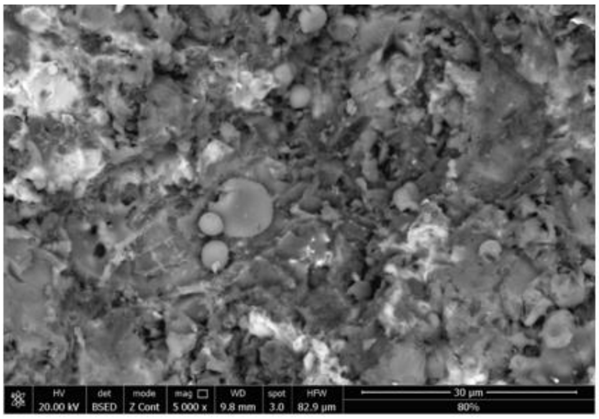
min air, the spray distance being maintained at 170 mm.

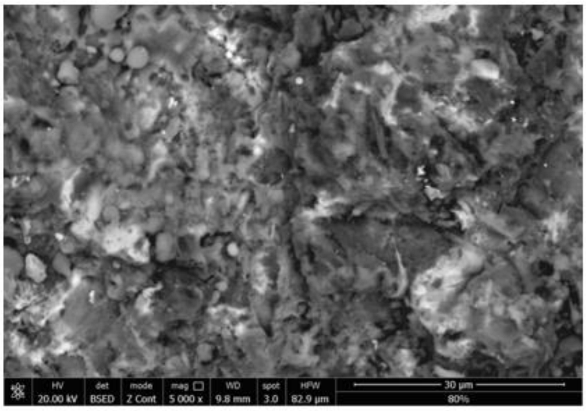
In fig. 6.3 shows the surface morphology of the sprayed layer using Al2O3 / HA powder compositions. Analyzing the SEM images in Fig. 6.3 and 6.4, it is observed that there were no thermal spray defects (cracks, exfoliation, etc.). The EDAX analyzes in Fig. 6.5 confirms the differences between the chemical compositions of the coatings.



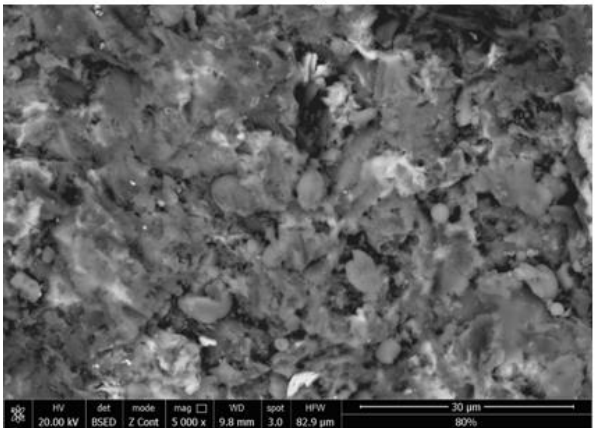
a) ***b)*** ***c)*** ***d)***

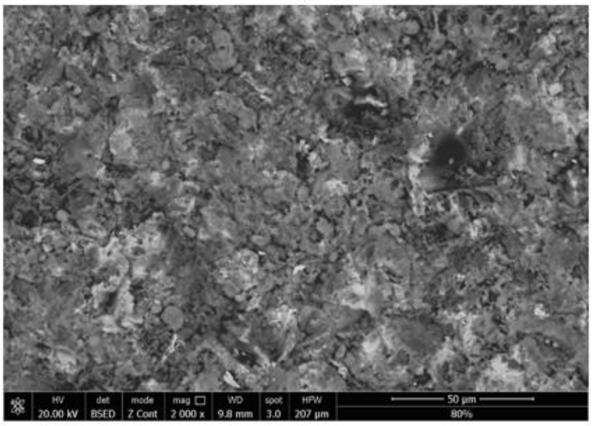
Fig. 6.3. *SEM micrographs of the deposited layers 50 % Al_2O_3+50% HA*





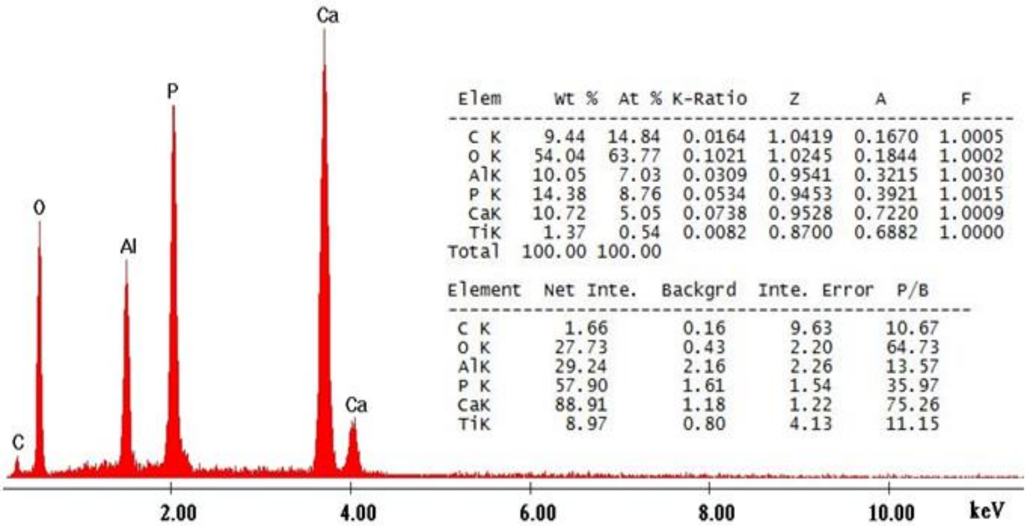
a) ***b)***



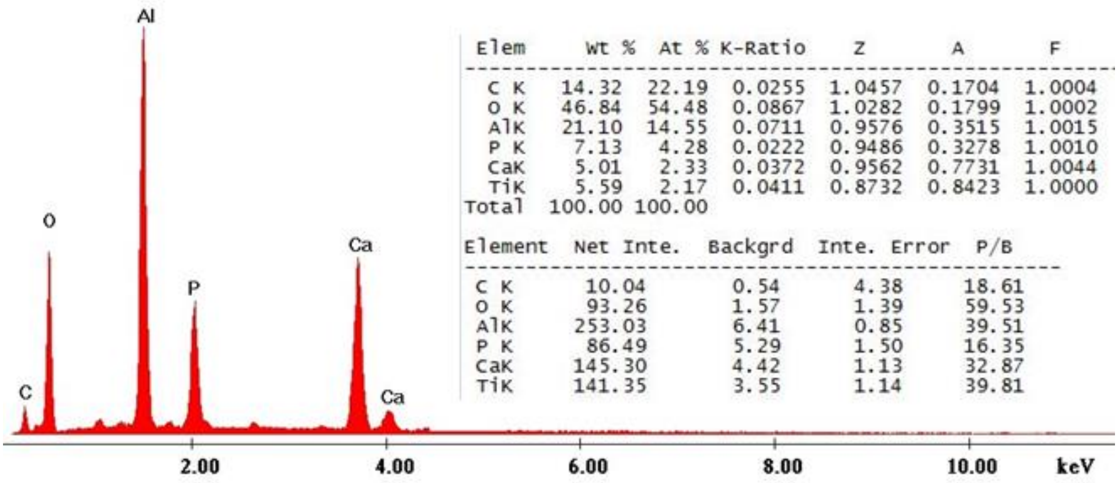


c)d)

Fig.6.4 *SEM micrographs of the deposited layers 80 % Al_2O_3+20% HA*



a)



Elem	Wt %	At %	K-Ratio	Z	A	F
C K	14.32	22.19	0.0255	1.0457	0.1704	1.0004
O K	46.84	54.48	0.0867	1.0282	0.1799	1.0002
AlK	21.10	14.55	0.0711	0.9576	0.3515	1.0015
P K	7.13	4.28	0.0222	0.9486	0.3278	1.0010
CaK	5.01	2.33	0.0372	0.9562	0.7731	1.0044
TiK	5.59	2.17	0.0411	0.8732	0.8423	1.0000
Total	100.00	100.00				

Element	Net Inte.	Backgrd	Inte. Error	P/B
C K	10.04	0.54	4.38	18.61
O K	93.26	1.57	1.39	59.53
AlK	253.03	6.41	0.85	39.51
P K	86.49	5.29	1.50	16.35
CaK	145.30	4.42	1.13	32.87
TiK	141.35	3.55	1.14	39.81

b)

Fig. 6.5. The results of the EDAX analysis of the deposited layers:
a - 50 % Al_2O_3+50% HA, b -80 % Al_2O_3+20% HA

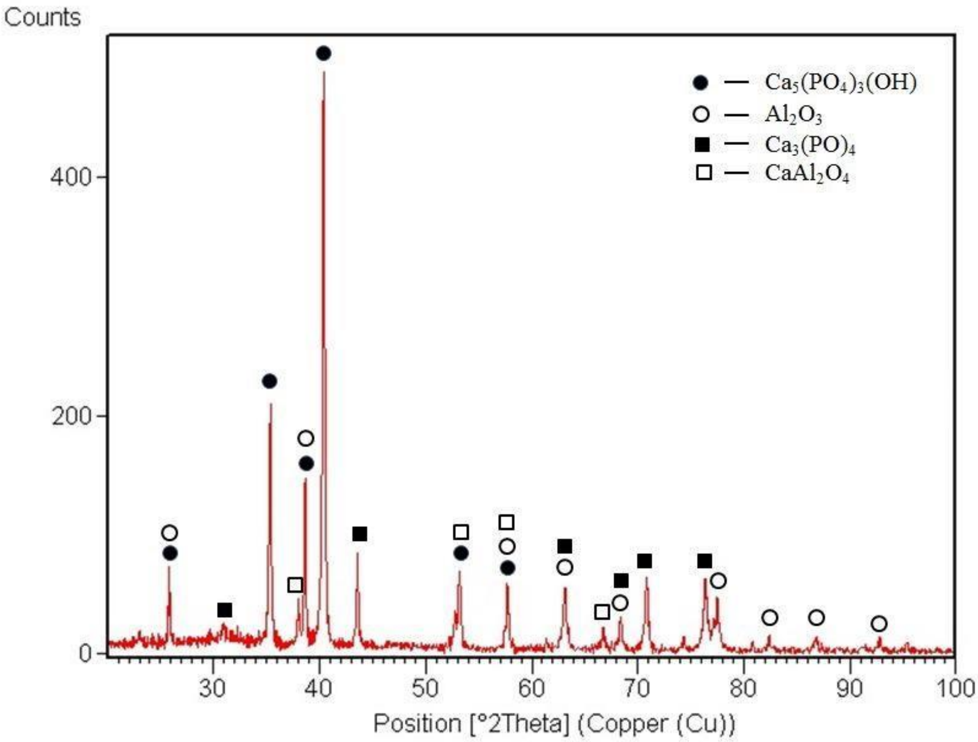


***Fig. 6.6** XRD diffractogram of the deposited layer, 50 % Al_2O_3+50% HA*

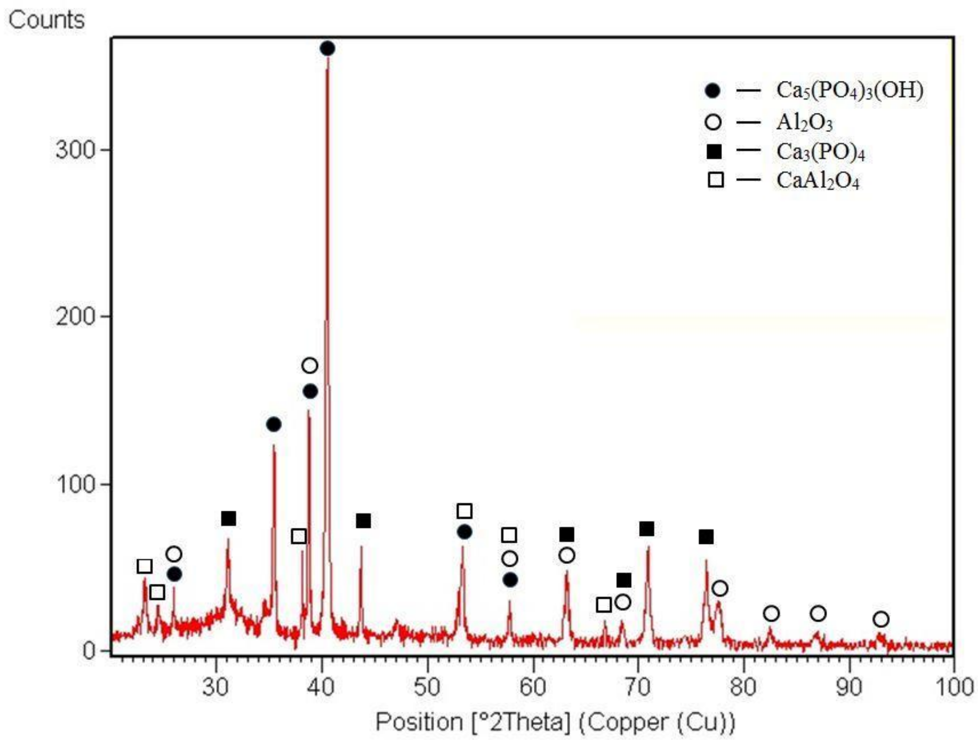
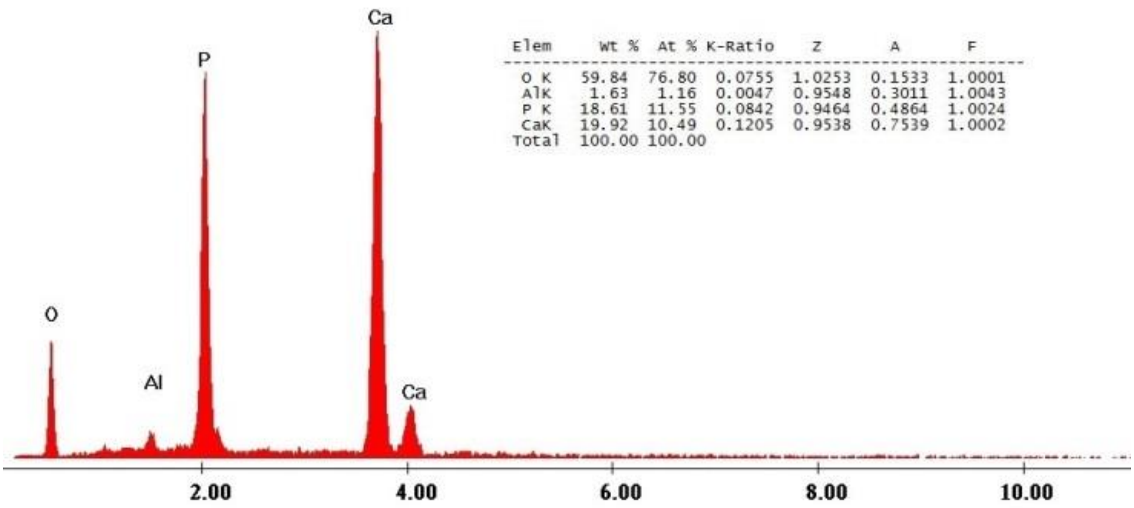
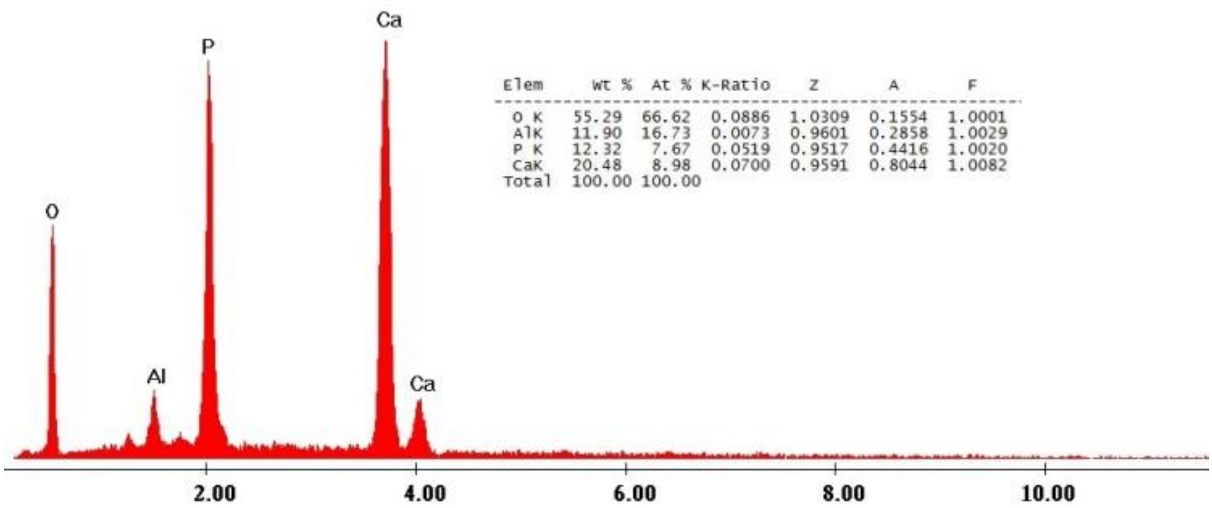


Fig. 6.7 *XRD diffractogram of the deposited layer, 80 %Al_2O_3+20% HA*



Elem	Wt %	At %	K-Ratio	Z	A	F
O K	59.84	76.80	0.0755	1.0253	0.1533	1.0001
AlK	1.63	1.16	0.0047	0.9548	0.3011	1.0043
P K	18.61	11.55	0.0842	0.9464	0.4864	1.0024
CaK	19.92	10.49	0.1205	0.9538	0.7539	1.0002
Total	100.00	100.00				

Fig.6.10 *The X - ray energy dispersion spectrum of the layer composed of 50%Al_2O_3 + 50%HA after in vitro mineralization*



Elem	Wt %	At %	K-Ratio	Z	A	F
O K	55.29	66.62	0.0886	1.0309	0.1554	1.0001
AlK	11.90	16.73	0.0073	0.9601	0.2858	1.0029
P K	12.32	7.67	0.0519	0.9517	0.4416	1.0020
CaK	20.48	8.98	0.0700	0.9591	0.8044	1.0082
Total	100.00	100.00				

Fig.6.11 *The X - ray energy dispersion spectrum of the layer composed 80%Al_2O_3 + 20%HA after in vitro mineralization*

Conclusions

The immersion of the coating layers, Al2O3-HA, deposited by the method of thermal spraying with high speed flame, HVOF, on the surface of pure commercial titanium, in SBF solution, showed that positive results are obtained in terms of their biocompatibility. Biological hydroxyapatite developed and increased on the surface of the exposed samples, showing a good bioactivity of the deposited coating.

The High Speed Flame Spray (HVOF) method can be used to obtain Al2O3-HA coatings on the surface of a titanium substrate in order to improve wear resistance and biocompatibility.

X-ray diffraction images showed that the structure of the hydroxyapatite did not undergo significant changes, which could decrease the restoration of bone tissue. Moreover, the measured surface roughness showed high values (Ra = 5.5 µm) which ensure a good osseointegration of the implant in the human body due to a larger contact surface.

The presence of alumina in the deposited coatings has improved the wear resistance of titanium by about 2-3 times.

7. GENERAL CONCLUSIONS. PERSONAL CONTRIBUTIONS. FUTURE RESEARCH DIRECTIONS (Selection)

General conclusions

Due to the diversity of activities carried out by people with locomotor disabilities, the experimental study of prostheses, the use of numerical methods of analysis and simulation continues to be a topical issue, due to the increase in the number of people and their involvement in various sports competitions.

The creation of new materials (polymers, metal alloys, ceramics, carbon fiber and glass composites, etc.) through state-of-the-art technologies, and the selection principles for the manufacture of orthopedic prosthesis components lead to the elimination of the concept of disability, as a walk and form of activity.

Experimental tests, followed by numerical simulation, must continue in order to obtain effective prostheses in the daily walk, especially for in the sports movement that stabilizes the psychology of the disabled person and does not lead to his isolation as an individual.

For people with disabilities in one leg, such as the one used in the tests in this thesis, gait is influenced by the type of prosthesis, with observable differences in the movement of the balance - the change between the prosthetic foot and the healthy one.

Energy consumption, when moving with the prosthesis, is dependent on the shape of the prosthesis and the fixing system on the non-amputated side of the foot, due to the angular movements of the hip and knee.

The correction of the kinematic system of displacement, towards a correct shape, which does not affect other parts of the healthy body, requires the realization of mobile joints at the level of the ankle, similar geometrically-functionally to the shape-real.

Experimental stress tests on monoaxial tensile and compression show the dependence, behavior and rupture of the 4-layer laminate composite on the orientation of the sheets, the formation of cracks, their propagation and the way the carbon fibers break.

Numerical modeling and analysis performed on a commercial 5H satin composite prosthesis shows the benefit of using professional software in estimating the value of the parameters that define the state of stresses and strains, topological optimization and geometric shape.

The immersion of the coating layers, Al_2O_3-HA, deposited by the method of thermal spraying with high speed flame, HVOF, on the surface of pure commercial titanium, in SBF solution, showed that positive results are obtained in terms of their biocompatibility.

Reconditioning of internal prostheses, such as hip prostheses, made of biocompatible titanium-based material can be accomplished by using the high-speed flame spray (HVOF) method to ensure the formation of biocompatible Al_2O_3-HA coatings. wear resistant.

Personal contributions

The personal contributions, brought by this doctoral thesis, are:

- analysis of the materials used for the components of the prostheses, depending on the application and the activity of the disabled person, based on the physical-mechanical properties and the structure;

- it has been shown that the choice of materials from which the components of the orthopedic prosthesis are made depend on the degree of disability and the activity of the disabled person, (only for walking or for a certain form of sport practiced) and that it must have elasticity and resistance to the mechanical stresses determined by the type of walking;

- highlighting the difference in energy consumption, made during travel, based on the values of the characteristic parameters, which shows the degree of accommodation and efficiency of the prosthesis on the gait of the person with a locomotor disability;

- based on the results obtained by numerical simulation it was shown that for four values of the factor of multiplication of the compressive / pressing force (44,838 ,; 46,265; 340,24 and 348,16) the maximum deformation does not exceed 1,014, which shows that the orthopedic prosthesis of leg - modular with internal silicone sleeve, is stable from a dynamic and static point of view;

- performing static tensile and monoaxial compression tests on two types of symmetrically layered composite consisting of 4 overlapping sheets oriented at 0 ° and 45 °.

- making Al2O3-HA coatings on titanium alloy, using the HVOF method, which increased the abrasion resistance by about 3 times.

Future research directions:

- experimental research and finite element analyzes on upper limb prostheses
- experimental research and numerical modeling of various prostheses for the lower limbs, intended for athletes with locomotor disabilities
- experimental research and numerical analysis on hip prostheses
- studies and research on various materials used in the manufacture of highly required lower limb prosthesis components
- continued testing with the Zebris system and for people with locomotor prostheses on both legs.

Bibliography (Selection)

1. Aiordachioae, G. A., Implicațiile protezării în dinamica articulară şi în menținerea sanogenezei la amputațiile membrului inferior pentru boala arterială periferică, Universitatea de Vest "VASILE GOLDIS", Teză de doctorat, Arad, 2013.
2. Barela, A., Feritas, P. de, Celestino, M., Camargo, M. și Barela, J., Groundreactionforcesduringlevelgroundwalkingwith body weightunloading, Brazilian Journal of PhysicalTherapy, vol. 18, pp. 572-579, 2014.
3. Begon, M. și Lacouture, P., Modélisationanthropométriquepouruneanalysemécanique du geste sportif. Modèles, leurscaractéristiques et leurvalidation, Partie 1, Science et Motricité, vol. 2, nr. 54, pp. 12-33, 2005.
4. Brăileanu, P. I., Cercetări privind optimizarea protezelor personalizate, Universitatea Politehnica din Bucuresti, Teză de doctorat, București, 2020.
5. Bulancea, V., Materiale Metalice Avansate utilizate in Medicina, Teza de doctorat, Universitatea Tehnica "Gheorghe Asachi" din Iasi , 2018.
6. Constantinescu, I.N., Picu, R.C, Hadăr, A., Gheorghiu, H., Rezistenţa materialelor pentru ingineria mecanică, Editura BREN, Bucureşti, 2006
7. Cristescu, N., Mecanica materialelor compozite, vol. 1, Editura Universitatea din Bucureşti, București, 1983.
8. Dragulescu, D., Apreciere prin metode comparative a refacerii parametrilor biomecanici ai subiecţilor cu deficiențe motrice, Revista de Politica Ştiinţei și Scientometrie, Număr Special, 2005.
9. Filep, F.-R., Cercetări privind optimizarea performanţelor cupă-proteză-bont cu considerarea aspectelor tribologice, , Teză de doctorat, Universitatea Tehnică "Gheorghe Asachi" Iași, 2017.
10. Ghiban, B., Metallicbiomaterials, Ed. Printech, Bucureşti, 1999
11. Hadăr, A., Constantinescu, I.N., Gheorghiu, H., Coteţ, C.E., Modelare şi modele pentru calculi în ingineria mecanică, Editura Printech, Bucureşti, 2007

12. Hadăr, A., Marin, C., Petre, C., Voicu, A., Metode numerice în inginerie, Editura Politehnica Press, Bucureşti, 2005
13. Ioanovici, T. G., Contribuții la sinteza de hidroxiapatita dopată cu magneziu și cercetări asupra proprietăților mecanice în vederea utilizării ei în implanturile osoase, Editura Politehnica, Timisoara, 2012.
14. Ishikawa, T., Matsushima, M., Hayashi, Y., Chou, T. W., Experimental confirmation of thetheory of elastic moduli of fabric composites, Journal of CompositeMaterials, vol. 19, pp. 443-458, 1985.
15. Mareș, M., Materiale compozite, Tipografia Universitatea Tehnică "Gh. Asachi" Iași, 2002.
16. Pasca, O., Studii şi cercetări privind ortezele de glezna și picior, Universitatea Politehnica Timișoara, Teză de doctorat, Timișoara, 2014.
17. Portnoy, S., Siev-Ner, I., Yizhar, Z., Kristal, A., Shabshin, N., Gefen, A., SurgicalandMorphologicalFactorsthatAffectInternalMechanicalLoads in Soft Tissues of theTranstibialResiduum, Annals of Biomedical Engineering, vol.37, nr.12, pp.2583–2605, 2009.
18. Rădulescu, D., Suprafețe bioactive și nanostructurate în endoprotezele articulare, Universitatea "Carol Davila", Teză de doctorat, București, 2018.
19. Țăranu, N., Bejan, L., Cozmanciuc, R. și Hohan, R., Materiale și elemente compozite I. Prelegeri și aplicații, Editura Politehnium, Iași, 2013
20. Tengwall, P., Lundstrom, I., Physio-chemicalconsideration of titanium as a biomaterial, ClinicalMaterials, vol 9, pp.115-134, 1991
21. Toader-Pasti, C., Cercetări privind dezvoltarea sistemelor de testare ale protezelor de gleznă, Teză doctorat, Editura Politehnica, Timișoara, 2014.
22. Utu, I.D., Mitelea, I., SNMS investigations of thermallysprayedcoatings, MaterialsTesting, vol. 57, nr.3, pp. 262-272, 2015
23. *** Ottobock, https://www.ottobockus.com/prosthetics/info-for-new-amputees /prosthetics-101 /finding-the-best-foot-for-you/, 2021.
24. *** ZebrisCMS, ZebrisMeasuringSystem for 3D MotionAnalysis CMS – HS.Technical data andoperatinginstructions. s.l.:s.n., 2006;
25. **** OSSUR, „ossur.com," Ossur Europe, [Interactiv]. Available: https://www.ossur.com/region-selector.
26. **Szabo, A.,**Bordeasu, I., Utu, I. D., et al. In vitro Behaviour of Alumina-Hydroxiapatite Composites Coatings, REVISTA DE CHIMIE Vol. 69, Issue 6, pp. 1416-1418, 2018
27. **Szabo, M. A.,**Atleţii cu handicap locomotor şiperformanţa, A XVI-a Conferință Internațională – multidisciplinară „Profesorul Dorin PAVEL – fondatorul hidroenergeticii românești", SEBES, pp.225-230, 2016