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COMPARATIVE GAIT ANALYSIS FOR HEALTHY AND LOWER LIMB AMPUTATION SUBJECTS

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Abstract: *Human motion analysis provides essential insights that can enhance the performance of modular lower limb prostheses, support rehabilitation assessment following surgeries such as amputation, and help prevent injuries during the prosthetic process. The human movement is typically described using dynamic and kinematic parameters. Kinematic characteristics include distances, velocities, accelerations, and segment trajectories, while dynamic characteristics refer to internal and external forces and torques. Understanding the forces acting on the body during locomotion is crucial for analyzing the loading of different body segments. At the same time, kinematic data offers valuable information about stability and balance during movement, both in healthy individuals and amputees.*
Key words: *biomechanics, gait analysis, lower limb amputation.*

1. INTRODUCTION

Lower limb prostheses are medical devices designed to replace the function and appearance of the missing limb as much as possible. The development and implementation of passive lower limb prostheses should be guided by scientific research that assess performance, stability, and balance in individuals using prostheses, with results benchmarked against those obtained from healthy control subjects.

Fifteen subjects were analyzed from a comparative perspective (eight healthy individuals, three with TT (transtibial) amputation, and four with TF (transfemoral) amputation to evaluate their kinematic and dynamic characteristics, aiming to improve the prosthetic components used in the development of modular TT and TF prostheses.

2. EXPERIMENTAL STUDIES

The present research outlines two techniques for assessing gait with modular prostheses: (1) gait analysis using the open-source Kinovea software to determine joint angles of the lower limbs throughout a complete gait cycle, and (2) ground reaction force measurement using Tekscan's mat-based system connected to the F-Mat Clinical software. The study involves both healthy individuals and users of modular transtibial (TT) and transfemoral (TF) prostheses.

The configuration includes a ground reaction force analysis device (MatScan – Tekscan) and a video recording equipment. Critical aspects such as subject positioning, camera placement, movement velocity, and capture rate are evaluated to achieve superior video quality and ideal field of view. The camera is positioned on an adjustable-height tripod, 3 metres from the prosthetic subject's examination region and approximately 0.6 meters above the floor. The calibrated field of view distinctly encompassed the subject's lower limb during motion. The camera lens's ideal focus point has been predetermined to produce good photos and video, and a series of red markers positioned on the subject's lower limbs enable visualisation of hip and knee joint angles during the gait cycle. The subject is asked to walk normally so that the self-selected speed is reached before the lower test limb contacts the center of the MatScan plate.

Simultaneously, movement tracking was captured by the Canon EOS 4000D video

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camera for analysis, using Kinovea software to determine the hip and knee joint angles throughout each whole gait cycle.

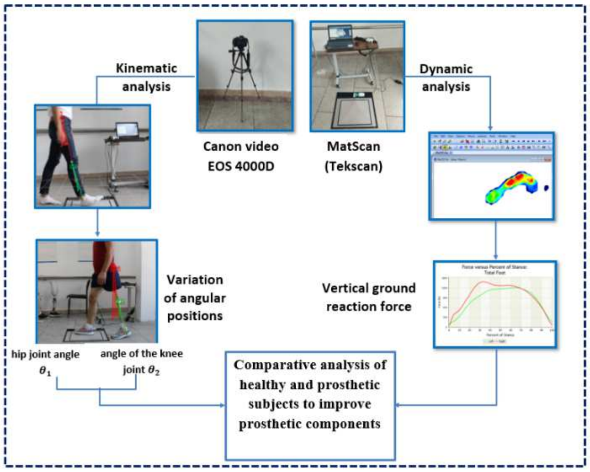

Fig. 1. Diagram of prosthetic gait analysis with Tekscan equipment and Kinovea software for comparative analysis of normal and prosthetic human gait.

The Tekscan system (F-scan and MatScan) has been thoroughly documented in scientific literature, with both the manufacturer and researchers affirming its reliability in accurately quantifying dynamic plantar loading patterns of the foot.

Thus, the examination of plantar pressures and forces, both from a static perspective and during physical activities like walking, becomes essential to the evaluation and analysis of lower limb prosthesis as well as the management of lower limb disorders [1] [2].

3. THE TARGET GROUP

The study was conducted on a group of healthy individuals as well as a group of individuals with TT and TF amputations, who were users of modular calf and thigh prosthesis. To perform a comparative study that would demonstrate the performance of the used prosthetic components, subjects equipped with modular prostheses exhibiting varying degrees of mobility were assessed using MatScan equipment and Kinovea software, with the results compared against data from healthy individuals. Subjective evaluation was also used, calling on the wearer's feedback, which could be freely expressed by completing a questionnaire.

Table 1

Evaluation and registration of healthy subjects.

No.	Gender	Mass/Height	Lifestyle
1.	M	90 kg/1.63 m	Sedentary
2.	M	70 kg/1.72 m	Active
3.	F	68 kg / 1.64 m	Active
4.	M	80 kg/1.78 m	Active
5.	F	125 kg/1.68 m	Sedentary
6.	F	70 kg/1.65 m	Active
7.	M	110 kg/1.82 m	Active
8.	M	66 kg/1.75 m	Active

Table 2

Evaluation and registration of subjects with TT and TF amputation.

No.	Gender	Mass/Height	Amputatio n level	Lifestyle
9.1.	F	78 kg/1,63 m	TT	Semiactive
9.2.	F	78 kg/1,63 m	TF	Semiactive
10.	M	114 kg/1,82 m	TT	Sedentary
11.	M	100 kg/1,74 m	TF	Active
12.	M	78 kg/1,80 m	TF	Active
13.	M	75 kg/1,78 m	TF	Active
14.	M	85 kg/1,68 m	TT	Active
15.	M	112/kg/1,75m	TT	Sedentary

4. EQUIPMENT FOR GAIT ANALYSIS

4.1. The MatScan system

The Tekscan system, which measures the plantar pressure developed by the foot both inside the shoe (F-scan) and freely positioned on a pressure measuring platform (MatScan), is produced by the Tekscan company in Boston, United States of America (US) [3].

Resistive sensors gather pressure-related information from the plantar area of the foot, irrespective of whether Tekscan, F-Scan, or MatScan equipment is being used. This data is accurate and dependable, allowing for the analysis of the foot's functionality and gait. Using the two types of equipment, a thorough examination of plantar pressure is carried out by recording data at various points during the recorded walking cycle, in both static and dynamic mode.

The "FootMat clinical" software was used, with the help of which analyzes can be made on the functionality of the foot, on normal and pathological walking with the evaluation of foot conditions that cause imbalances and defective posture, by investigating the plantar pressure, respectively, the values of the ground reaction force.

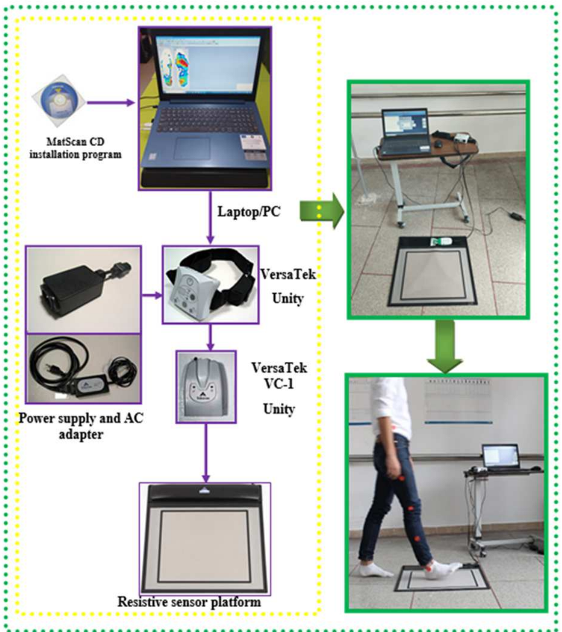

Fig. 2. Hardware components of MatScan (Tekscan) equipment.

The Mat-Scan system allows for real-time screening for plantar pressure distribution of the foot (or artificial limbs – modular lower limb prostheses) while the person is standing or walking on the measurement platform. It can also analyse dynamic weight transfer, local pressure concentrations and identify asymmetries in the plantar pressure profile between the left and right foot.

4.2. Kinovea

The "Kinovea" software was used for video analysis of the recordings. For the videos created, Kinovea was used to analyze the position of the markers located on the knee and hip joints and track their movement throughout the walking cycle.

The angle of the body segment is defined as the angle formed by the analyzed segment with the vertical axis [4]. It can be calculated by knowing the coordinates of the proximal and distal ends of a body segment, in some known plane.

The angle of an anatomical joint is defined as the angle between the line of the proximal and distal segments of that joint. Figure 3 and 4 shows the hip and knee joint angles (θ_1, θ_2).

The knee joint angle is defined by the long axis of the tibia relative to the long axis of the femur, with full extension defined as zero degrees and flexion motion being positive. The hip joint angle is defined by the long axis of the femur relative to the pelvis, with flexion defined as positive and extension negative.

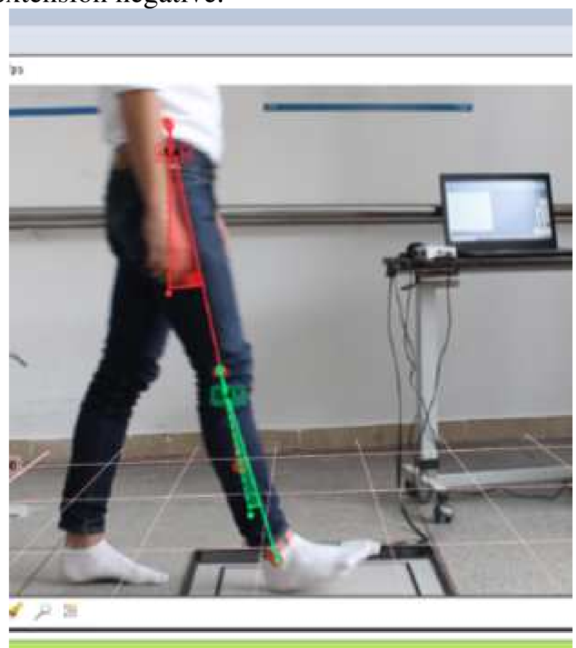

Fig. 3. Placement of landmarks on a healthy subject and measurement of hip and knee angle using Kinovea software.

Fig. 4. Placement of landmarks on a prosthetic subject and measuring the hip and knee angle using Kinovea software.

5. THE RESULTS OBTAINED

To estimate the moments in the hip and knee joints, it is necessary to determine the variation in the joint angles of the lower limbs and the variation in the ground reaction forces during the gait cycle.

The ground reaction force occurs at the contact between the foot and the ground, because of the foot resting on the ground or the impact between them. It has three components that act in three orthogonal directions: anteroposterior (Fx), vertical (Fy) and mediolateral (Fz). In the present study, particular attention is given to the maximum vertical component of the ground reaction force, denoted as Fy.

The maximum ground reaction force value is obtained using the FootMat Clinical software, for healthy subjects and those with modular TT and TF prostheses, the results being centralized in tables 3 and 4. Also, the coefficient with the force of the person's weight was determined, given by the formula:

$$r = \frac{F_y}{G} \qquad (1)$$

Table 3

The data obtained for the maximum reaction force in the vertical direction (Fy) and the determination of the correlation coefficient with the weight of healthy subjects.

Analyzed subject	G [N]	F_y [N]	r
Subject 1	882,9	982	1,112
Subject 2	686,7	815,3	1,187
Subject 3	667,08	790,5	1,185
Subject 4	784,8	985,7	1,192
Subject 5	1226,25	1385,7	1,13
Subject 6	686,7	810,6	1,180
Subject 7	1079,1	1325,5	1,228
Subject 8	647,46	742,3	1,146

The results obtained for the correlation coefficient of the ground reaction force with weight, in the case of healthy subjects, are in the range of 1.112-1.228, resulting in an average value of 1.17. This value is taken as a reference and considered an accepted value within normal limits, later being compared with the values obtained for the force restitution coefficient in the case of prosthetic subjects.

Table 4

The data obtained for the maximum reaction forces in the vertical direction (Fy) and the determination of the correlation coefficient with the weight of the prosthetic subjects.

Analyzed subject	G [N]	F_y [N]	r
Subject 9_TT prosthesis	765,81	981,13	1,281
Subject 9_TF prosthesis	765,81	846,75	***1,105***
Subject 10_TT prosthesis	1118,34	1382,06	1,235
Subject 10_ healthy collateral limb	1118,34	1498,32	1,339
Subject 11_TF prosthesis	981	1240,55	1.264
Subject 11_ healthy collateral limb	981	1494,42	1,523
Subject 12_TF prosthesis	765,18	929,59	***1,214***
Subject 12_ healthy collateral limb	765,18	1001,8	1.309
Subject 13_TF prosthesis	735,75	972,22	1,321
Subject 13_ healthy collateral limb	735,75	1036,82	1,409
Subject 14_TT prosthesis	833,85	1007,29	***1,208***
Subject 14_ healthy collateral limb	833,85	1090,23	1.307
Subject 15_TT prosthesis	1098,72	1361,31	1,239
Subject 15_ healthy collateral limb	1098,72	1480	1,347

It can be seen that, in the case of prosthetic subjects (except for subject 9, who has a bilateral amputation, on the right leg amputation, and on the left thigh amputation), the healthy collateral limb is affected, regardless of the amputation level, the values being included in the range 1.307-1.523, with a mean value of 1.331 in the case of subjects with TT amputation (subjects 10, 14 and 15 marked in green) and a mean value of 1.413 for subjects with TF amputation (subjects 11, 12 and 13 marked with green color). Thus, it can be concluded that the level of amputation directly influences the loading mode of the healthy limb. In other words, the higher level of the amputation is performed, the higher the value of the ground reaction force for the healthy leg.

On the other hand, in the case of the prosthetic lower limb the correlation coefficient of the reaction force for the transtibial prosthesis is in the range 1.208-1.239 (subjects 10, 14 and 15) with an average value of 1.227, and in the case for the hip prosthesis the range is between 1.214-1.321 with an average value of 1.266 (subjects 11, 12 and 13). Therefore, a small difference is noticeable between the subjects wearing TT prostheses and those with TF prostheses. In this situation, the reaction force is greater in the case of subjects with thigh amputation, caused by the absence of two important joints (the ankle joint and the knee joint), so that the wearer of the prosthesis makes a greater effort during locomotion to ensure balance in the support phase.

In the case of subject no. 9, with bilateral amputation, the supporting leg is the one prosthetic with the modular TT prosthesis, so that for him a higher value of the correlation coefficient of the reaction force with the weight of the subject was recorded.

It should be mentioned that the subjects wore different types of prosthetic components, from the standard ones to the most performing ones, but there was a common criterion for evaluating the way in which the movement is achieved, respectively, the comfort felt at the amputation member-prosthetic socket interface, during the gait cycle. The firmer the contact at the residual limb-silicone sleeve-socket interface, the closer the ground reaction force correlation coefficient is to that of a healthy person. For example, in the case of subjects 9 (TF prosthesis), 12 and 14, the prosthetic socket ensured firm contact with the amputated lower limb, dressed in the silicone sleeve, being made shortly before the recordings were made, at the dimensions appropriate to the amputation limb without volume variation.

For a clearer picture of the importance of the contact between the prosthetic socket, the silicone sleeve and the residual limb, the kinematic variables of the recorded subjects were analyzed during locomotion. Due to the small amplitude, with little impact on the positions of the calf and thigh, the movements of the leg around the ankle joint are neglected.

To determine the angles made by the knee and hip joints in the sagittal plane, the video recordings were processed using the Kinovea software.

The values of the determined angles, which correspond to the subphases of the walking cycle, together with the specified averages, thus, the results for the average values of the angle of the hip joints (θ_1) and knee joints (θ_2), for each situation presented above, are shown in the following graphs.

In figure 5, the average results obtained for the angles made by the hip joint (a) and knee joint (b) expressed in degrees during a complete walking cycle for healthy subjects are represented. In their case, the movement of the hip joint is relatively simple. It begins with a hip flexion that occurs at the beginning of the support phase (heel strike), followed by a maximum extension, reached in the last stage of the support phase (off the ground or propulsion), for which the angular position of joint reaches a minimum value of -13.1° (figure 5, a). Maximum flexion follows (approximately 35°), which prepares the foot for the next initial contact (heel strike).

In figure 5 are represented the average results obtained for the angles made by the hip joint (a) and knee joint (b) expressed in degrees during a complete walking cycle for healthy subjects.

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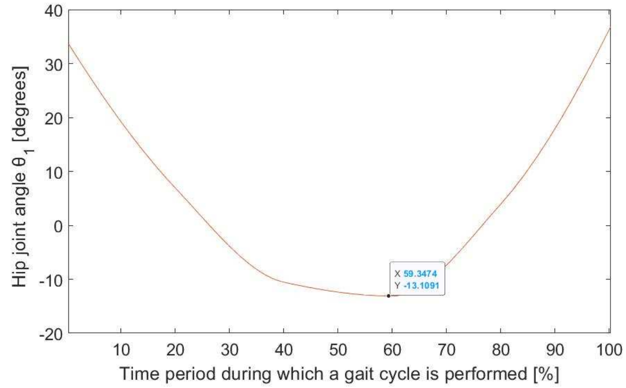
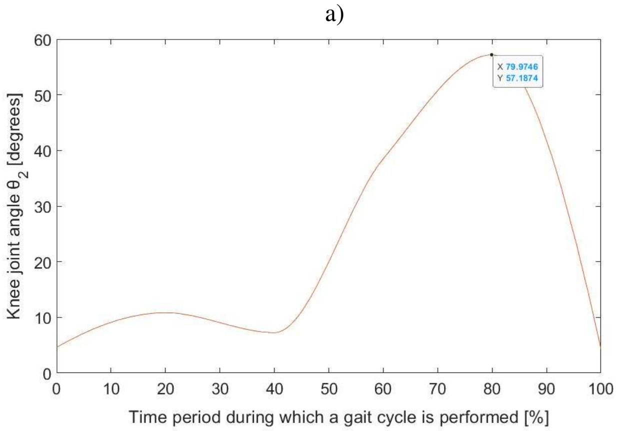
From figure 5, b the first peak of flexion is reached by the knee joint immediately after the attack with the heel, the recorded value being a maximum of 10°, after which the knee performs a slight extension, followed by a maximum flexion movement, which indicates the end of the support phase (take-off or propulsion). The angular amplitude of the knee joint is cyclic and reaches a maximum flexion angle value of 57.187°, although there is some variation in the maximum flexion for each recorded subject.



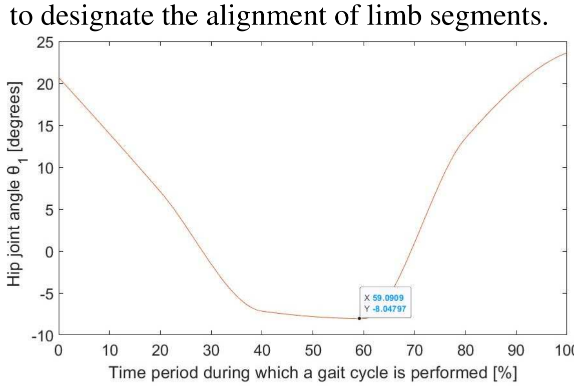
a)



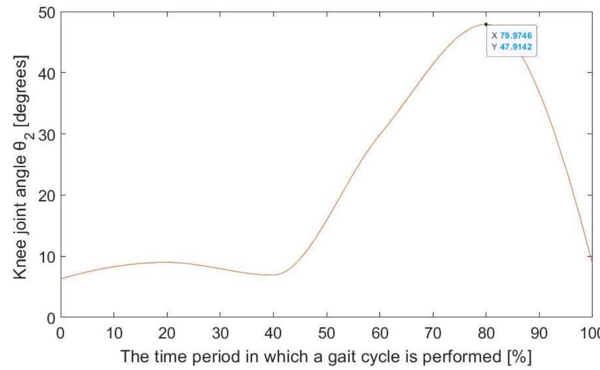
b)

Fig. 5. Plots of the mean angles made by the hip joint (a) and the knee joint (b) during a complete gait cycle for healthy subjects.

These differences may be related to differences in walking speed, the individuality of the subject analyzed, and the landmarks selected to designate the alignment of limb segments.



a)

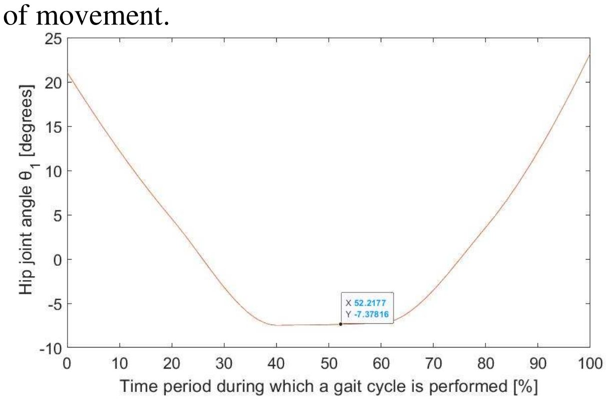


b)

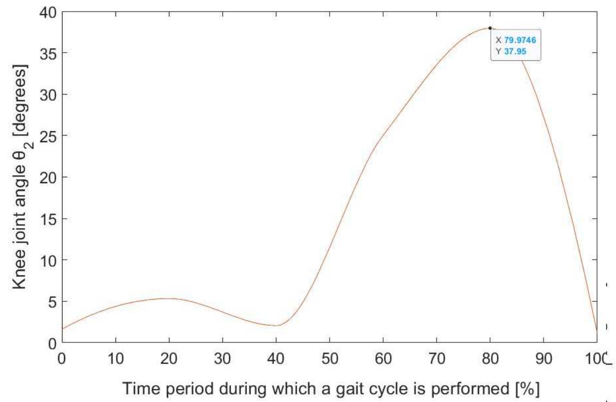
Fig. 6. Plots of the mean angles made by the hip joint (a) and knee joint (b) for the healthy collateral limb during a complete gait cycle (prosthetic subjects).

In the case of prosthetic subjects, the values obtained for the angle of the hip joint and of the knee joint for the healthy collateral limb are lower compared to the values recorded for the healthy subjects. For the hip joint, the maximum angular amplitude is recorded by moving from the initial position defined by an angle value of $+21^{o}$, up to -8.5^{o} (figure 6, a), representing the extension of the hip towards the end of the support phase. For the knee joint, the maximum recorded value of the angular position is 47.9^{o} (figure 6, b), which is decreasing (a difference of about 10^{o}) compared to the situation in which the subject is healthy.

The amputees' insecurity with their traditional prostheses is the main cause of the changes in the angular amplitudes of the two joints, the hip and the knee, which primarily show smaller steps and, as a result, a lower speed of movement.



a)



b)

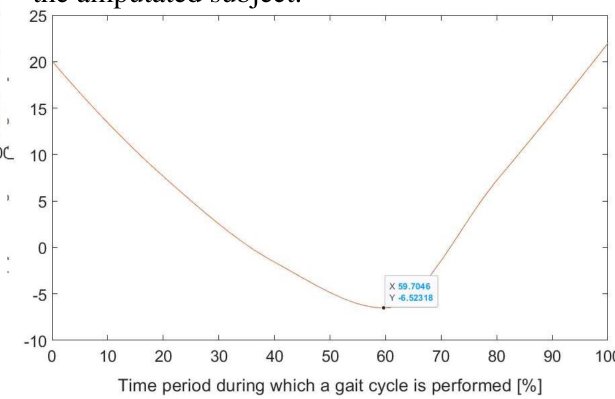
Fig. 7. Plots of mean hip (a) and knee (b) angles for the amputated limb in the middle third of the calf during a complete gait cycle (prosthetic subjects).

In comparison to the graphs obtained for the healthy subjects, as well as those with prostheses, in the case of the analysis of the healthy collateral leg, it is noted that the average angles made by the hip joint (figure 7, a) and the knee (figure 7, b) for the subjects who have had a transtibial amputation respect the same shape.

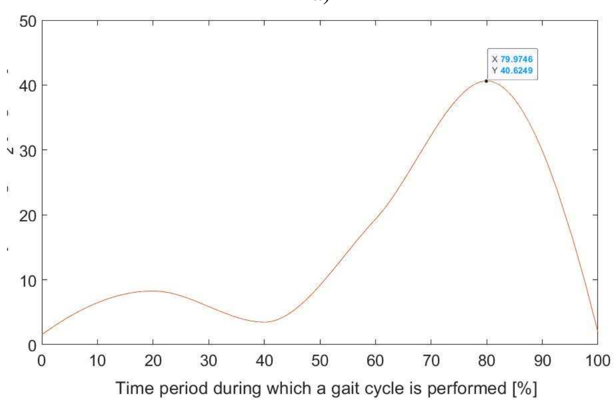
Thus, the extreme value obtained for the hip joint is -7.378^o (figure 7, a), the amplitude having a slight decrease, in relation to the other two analyzed situations. For the knee joint, the maximum value of the flexion angle is 37.95^o (figure 7, b) with a difference of about 10^o, for prosthetic subjects, but for which the healthy limb was analyzed) and respectively a difference of about 20^o, compared to healthy subjects. It is observed that the values have decreased considerably for the flexion movement, recorded for the knee joint of the prosthetic lower limb, with modular TT prosthesis.

This is since the residual limb is covered by the commercially purchased silicone sleeve, which completely covers the knee joint up to the area of the femoral condyles, so that flexion-extension movement of the knee is considerably obstructed by this component, which also includes a part of the thigh. A solution to solve this situation is to make the silicone sleeve, according to the geometric configuration of the residual limb, with the release of the popliteal area located posterior to the knee joint.

Figure 8 shows the graphs of the hip and knee joint angles for the amputated lower limb in the middle third of the thigh. For the hip joint, a maximum value of the extension angle of -6.5^o is recorded, which corresponds to the smallest angular amplitude of the extension movement. In conclusion, it can be stated that the level of amputation directly influences the locomotion of the amputated subject.



a)



b)

Fig. 8. Plots of the angles made by the hip joint (a) and the knee joint (b) for the amputated limb in the middle third of the thigh during a complete gait cycle.

In the case of the knee joint, the mechanical performances of the prosthetic joint are evaluated, for which a maximum value of the flexion angle of 40.6^o was obtained, being a higher value compared to the average recorded for the knee joint of the limb with transtibial amputation, but lower compared to the values obtained in the case of the healthy lower limb, both for amputated and healthy subjects. It should be noted that different types of prosthetic joints were evaluated, with the goal being to evaluate the influence of amputation level on the comfort and balance of the subjects enrolled.

7. CONCLUSION

The small values of the angular displacements of the hip and knee joints for the prosthetic subjects, both those evaluated for the healthy and the prosthetic leg, indicate an instability of the contact between the residual limb and the prosthetic socket connecting it, in the transtibial and transfemoral prostheses. One cause of this instability can be attributed to internal displacements at the residual limb-sleeve-socket interface, which causes distrust and fear, affecting the quality of prosthetic gait. The proposed solution to address this issue involves the use of an adjustable mechanical or pneumatic locking system (liner and prosthetic socket) for lower limb prostheses, which can be operated manually or automatically, to achieve dimensional adaptation to the geometric configuration of the residual limb. Thus, evaluation of the reaction force and the angles achieved by the prosthetic joints used in the construction of the prostheses can provide important information regarding the fit and adaptation of the residual limb in the prosthetic socket.

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Analiza comparativă a mersului pentru subiecți sănătoși și amputați de membre inferioare

Analiza mișcării umane poate oferi informații importante pentru îmbunătățirea performanței protezelor modulare de membru inferior, pentru evaluarea reabilitării după o intervenție chirurgicală de tipul amputației sau pentru prevenirea accidentărilor ulterior procesului de protezare. Mișcarea umană este de obicei descrisă folosind caracteristicile dinamice și cinematice. Caracteristicile cinematice includ distanțe, viteze, accelerații și traiectoriile unor segmente, în timp ce forțele și cuplurile externe și interne reprezintă caracteristicile dinamice. Înțelegerea forțelor exercitate asupra corpului în timpul mișcării este importantă pentru înțelegerea modului în care se încarcă diferitele segmente ale corpului, iar caracteristicile cinematice oferă informații cu privire la stabilitatea și echilibrul în timpul deplasării în cazul subiecților sănătoși, dar și amputați.

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