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WALK ON THE PROSTHESES WITH ROTATION DEVICES

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Abstract

A description is presented of the methods based on usage of strain gauges and electronic computers for determination of trajectory of a point in which summary ground-to-foot reaction is applied, and of torque acting to the human foot in locomotion. The appliance of the obtained data to one of the proposed mathematical models of the motions of the human body segments in horizontal plane is considered. In the light of the investigations of amputee's gait on the experimental prosthesis with rotators incorporated, the necessity is discussed of supplying the artificial lower extremities with the devices permitting the rotation of their links in the transversal plane.

Introduction

The working out of the prostheses and orthotics of progressive designs is impossible without detailed knowledge of the motions performed in the space by human extremities during daily activities.

In this paper an attempt is made to investigate the rotation of lower extremities and determine with the help of mathematical model their participation in human locomotion.

The model which was taken as the basis for mathematical description is a 4-link biokinematical chain (fig. 1). Links 1,2 and 3 correspond to the foot, shank and femur, and the 4th link - to the pelvis, trunk, upper extemities and head. The chain links are connected by the means of elastic elements which imitate the main extremity joints and it is supposed that the design of the above elements is similar to helical spring with spring rate $c_i - (i = 1,2,3)$.

The model is subjected to the action of gravity forces and ground forces, muscle forces, inertia forces, but only the motion in horizontal plane is taken into consideration, that is why the gravity forces are not included into the equations. Thus only the torque of components of ground forces round vertical axis is taken into account (M_z).

The following assumptions were adopted: the human body segments are rigid pieces, their masses are not dependent upon the step phases, the constraint between links is stationary and the whole system is heterogeneous. In such a way the given biokinematical chain can be listed as dynamic system. The links move only around vertical axis, that is why it was convenient to choose the angles of segments' rotation (φ_i , $i = 1,2,3,4$) as generalized coordinates.

The equations of the links' motions are given below:

$$
\begin{aligned}
I_1 \ddot{\varphi}_1 + C_1(\varphi_1 - \varphi_2) &= -M_{\text{мыш}_1} - M_z \,, \\
I_2 \ddot{\varphi}_2 - C_1(\varphi_1 - \varphi_2) + C_2(\varphi_2 - \varphi_3) &= M_{\text{мыш}_1} - M_{\text{мыш}_2} ; \\
I_3 \ddot{\varphi}_3 - C_2(\varphi_2 - \varphi_3) + C_3(\varphi_3 - \varphi_4) &= M_{\text{мыш}_2} - M_{\text{мыш}_3} \\
I_4 \ddot{\varphi}_4 - C_3(\varphi_3 - \varphi_4) &= M_{\text{мыш}_3} .
\end{aligned}
$$

where I_1 – I_4 – inertia moments of lower extremity segments relative longitudal axis;
$\varphi_i\ (i=1,2,3,4)$ – angles of segments rotation in horizontal plane
$C_i\ (i=1,2,3)$ – coefficients of spring rate;
M_z – torque of ground-to-foot reaction.

In such a form the equations may be used as formulae for the determination of muscle forces moments applied in the transvesal plane , but for this purpose it is necessary to Know the values of I_i, C_i, φ_i and M_z.

The values of inertia moments were obtained by registering the torsional vibration of cast positives which were made according to the living extremity segments. The value of the 4th inertia moment was calculated, the fourth link was compounded of two truncated cones and a sphere (fig. 1).

The values of c_i which correspond to stiffness of the joints were accepted as equal to 1. It means that friction in the joints was small.

The data about the angles of rotation of lower extremity segments were taken from the paper written by A.S.Levens /1/ Experimentally it has been found that the angle of foot slip relative to the ground is about 4° in normal walking and increases up to 8° in the amputee walking on prosthesis but its variations within the stance phase were not investigated. That is why to express the relation between angle φ_1 and time, function $\varphi_1 = \sin t$ was taken with amplitude which has in the first case the extremum of 4°, and in the second – 8°. It was earlier supposed /2/ that there was a linear relation between the level of amputation and the range of rotation of the remaining portion of the extremity. In the present investigation the same supposition was adopted and the values of angles φ_2 and φ_3 were multiplied by coefficients β_i which correspond to the level of amputation.

To determine the torque of ground-to-foot reactions it was decided to utilize the "force plate", which is schematically shown (fig. 2). The distributed vertical load is applied to foot during walking besides horizontal friction forces. That is why the torque of ground-to-foot reactions is convinient to calculate relative to the point E, in which the resulting vertical component is applied. In such a way the moments about horizontal axis can be excluded from the formulae and the value of torque can be determined on the basis of the formula shown below:

$$M_z = (B_y - C_y)(a-2d) + (A_x - B_x)(b-2d) + x_E \cdot R_y + y_E \cdot R_x \ ;$$

where X_E, Y_E – the coordinates of point E where the vector of ground-to-foot reactions is applied;
a,b and d – the dimensions of the "force plate";
A_i, B_i, C_i, D_i – reaction of the transducer which is proportional to the applied load.

To calculate the M_z values the coordinates of point E should be known, i.e. the distance from the platform edges to the point where the subject put his foot should be determined.

This problem of particular interest was to be solved. To get a solution a sole with incorporated strain gauges was used (fig. 3a).

The distances from the centres of each strain gauge to the axes of coordinate system lying in the plane of sole were determined in advance. The strain gauges were designed in such a way that responded only to the vertically acting load. The trajectory of the point in which the resulting supporting force is applied may be found out according to the algorithms used for the calculation of the coordinates of the centre of gravity:

$$X_E = \frac{\sum f_i x_i}{\sum f_i}; \qquad y_E = \frac{\sum f_i y_i}{\sum f_i};$$

The sole during gait investigations was put into a shoe. During walking the sensors in consequece were subjected to the load bearing, the electrical signals were put into the analogue computer MH-10. On the screen of the oscilloscopeN-6 connected to the output of the analogue computer the curve of the searched function can be seen (fig.3б).

The current coordinates X_E and Y_E of the point to which the vector of ground-to-foot reactions is applied were calculated according to the formulae analogous to those mentioned above. For this purpose four ring transducers located in the vertical plane were used. The algorithm for the definition the values of the torque M_E after substitution in the formulae of X_E and Y_E became as it is shown beneath:

The calculation of the torque values was carried out with the help of electronic computer (БЭСМ-4, Russian variant of IBM). Electric signals from each of the 12 ring transducers were registered on the photo-band of the 12-channel oscilloscope then after the necessary translation to the binary code the collected data were put into the computer. It has been found that alternative and to some extent symmetrical curve of torque corresponds to normal human gait. Amputee's walking on the prosthesis of commonly used type is characterized by the curve located on one side of the abscissa, i.e. it has direction of one sign (fig. 4). All the information collected about inertia and dynamic gait characteristics was used for the definition of moments of muscle forces on the basis of mathematical model given above. For this purpose a programme in the "algol" language has been developed and the processing of the initial information was carried out in the IBM.

The diagrams of moments of muscle forces applied between the joints of lower extremity in the transversal plane are shown on fig. 5 . The values of power developed in motion of living extremity segments as well as the energy spent by the whole extremity in walking (Table 1.) were calculated on the basis of muscle forces moments. The data represented in the table 1 indicates that the living extremity suffered some overloadings while walking on the prosthesis of commonly used designs.

Nowdays there are no prostheses except a few experimental prototypes which can provide transverse rotation of their links. The lack of motion in transversal plane very often caused trauma of the stump surface because in every step the stump should rotate inside the socket of the prosthesis.

The necessity of providing rotation in prostheses has been stated many years ago /2/. The further investigation of the motions in horizontal plane supported this idea /3,4/.

The rotation devices developed earlier have thus far failed to meet all functional criteria, the main disadvantadge was the lack of control of adsorbing mmoment. But as it was written in published materials the implementation of such devices in the prosthesis had positive effect.

This type of devices according to Lamoreux and Radcliff can be listed as devices of "passive rotation". The motions in such mechanisms round vertical axes generated by the inertia forces, are not dependent on the muscle forces and are determined mainly by the style of gait. The springs of different types, rubber tubes of corresponding diameters and rigidity and miscellaneous inserts made of elastic materials are frequently used in the above types of rotators as the moment absorbers.

To carry on the investigation of normal and pathological gait two devices of "passive rotation" were made and tested in our research laboratory. In one of them the elastic torsion made of vacuum brand of rubber was used (fig. 6 a), in the second - stainless plate (fig. 6 b).

To obtain the objective evaluation of gait on the prostheses with rotators the values of torque and the angles between the extremity segments in transversal plane in normal and prostheses walking were compared. The diagrams of torques and angle of rotation were obtained as the result of the experiments.

The curve of torques (fig. 7) comes closely to the M_z curve obtained in normal human walking. As it is shown in fig. 7 the curve has two portions located under the abscissa axis as well as above it, i.e. it has alternate sign character although it is not symmetrical relatively the point where its meaning value is equal to zero. The amplitude of M_z is almost the same as it was in normal human walking.

The degree of shank rotation relative to the prosthetic foot was determined with the help of potentiometer incorporated into the rotation device. Registration of this parameter was carried out by the oscilloscope. The diagram of shank rotation relative to the prosthetic foot is given in fig. 8 .

It is seen that in the moment of heel strike the shank rotates inside and stops its rotation at the moment when the extremity bears the whole weight. Then as the body moves forward the shank of the prosthesis begins to rotate outside and stops its rotation in the swing phase. Comparison of the diagrams of rotation in normal walking with those obtained in walking on the prostheses with rotation devices shows that the diagrams to some extent resemble each other.

Mathematical analysis of walking on the prostheses with rotators which was based on the model described above shows that incorporation of the rotation devices into the prosthetic designs leads to definite positive changes in the dynamics of gait. For example the changes of the moments of muscle forces leads to the fact that living extremity suffers less overloadings than in walking on the prostheses of commonly used designs.

The diagrams of powers generated in walking on the prostheses with incorporated rotators are shown on fig. 9 and the table 1 presents the values of energy losses on the prosthetic and living sides correspondingly. Analysis of the presented data says that the implementation of the devices permitting axial rotation in the artificial legs substantially improves the gait and diminishes tiredness and increases the comfortability. These effects are extremely useful in the below-knee prostheses where due to less amounts of soft tissues the slipping of the stump inside the socket causes very often the injures of skin.

Besides the axial rotation the important effect in gait is knee flexion in early stance phase. As it was stated in the work by M.I. Lapaev / 10 / this functional peculiarity leads to the descend of the centre of gravity by 10-15 mm, which in its turn influences decreasing of energy losses.

But modern types of artificial legs do not permit the above motion in the knee joint and consequently the needed vertical shift of the gravity centre.

It was decided to work out a design of the prosthesis that could provide the axial rotation as well as the descent of the gravity centre of human body in early stance.

The work resulted in the above-the-knee prosthesis in which these two motions were combined by the unit of "forced" rotation. The unit was called so because it provided the force rotation of the socket in early stance phase under the action of gravity forces.

The range of transverse rotation of the socket and the amount of its vertical shift were chosen according to the mean statistical data. So the permitted rotation was within 15° and the vertical shift was equal to 15mm. It was supposed that the shortening of the prosthesis length in early stance corresponded to the transferance of the gravity centre in gait. As it was mentioned above the methods of registration ground-to--foot reactions and muscles electric activity were adopted to get the objective evaluation of gait on the prostheses with rotation devices.

The search of variation of vertical compoment of ground--to-foot reactions presented great interest. From early investigations it is known that this component in normal gait has two extremums. After amputation and prosthesis fitting the biggest changes take place with the vertical component on the side of severed extremity. In that case the curve of resulting vertical force does not sometimes have two humps and if they are there the distance between them is often very small. Improvement of prostheses designs leads to the more normally looking curve of vertical force.

That is why to establish the advantadges of new designs of the prosthesis the comparison of diagrams of vertical force in normal and pathological gait was carried out.

Experimentally it was found that the utilization of the device of "forced" rotation results in the considerable change of vertical force curve: the distance between the extremums increases and the amplitude of the second hump corresponding to the toe-off becomes higher.

The diagram of vertical force on the side of the artificial extremity differs in a lower degree from that of living extremity. Electric activity of the muscles on the living extremity and on the stump was registered. In the first series of tests the gait on the prosthesis with devices of "forced" rotation was investigated, in the second series - that on the prosthesis without rotation. All tests were done under the normal speed of walking. There was studied the electric activity of the following muscles on the stump: tensor fascial latae and gluteus

Table I. Energy expenditures in normal and pathological gait

Lower extremity	Energy expenditures (in joules)		
	Normal gait	Gait on the prostheses without rotators	Gait on the prostheses with rotators
Unsevered	I2,6	I5,4	I3,6
Fitted with prostheses	-	8,9	9,7

Table 2. Integrated electric activity of stump muscles (number of impulses per one step)

Stump muscles	Type of the prosthesis													
	Prosthesis with rotator incorporated							Prosthesis without rotator incorporated						
	Step number						Mean value	Step number						Mean value
	I	2	3	4	5	6		I	2	3	4	5	6	
tensor fascial latae	28	27	28	29	27	34	28,8	I8	20	I4	I8	I9	I8	I8,0
gluteus medius	33	40	35	33	37	33	35,I	22	27	2I	26	27	24	24,5

medius, on the living extremity additionally electric activity of the following muscles was examined: tibialis anterious, gastrocnemius medialis, peroneus longus and soleus.

The electric activity was recorded with the help of 6-channeled oscilloscope. To identify the step phases the electric signals from sole switches were simultaneously recorded on the same paper band.

The tests carried out show that the electric activity of stump muscles substantially increases when the prosthesis with device of "forced" rotation is used. It can be explained by the fact that in prosthesis without axial rotation these muscles have no motion task and they do not act.

As soon as the prosthesis gets additional mobility in horizontal plane the stump muscles begin to take larger part in locomotion of artificial extremity that is a reason why their activity increases. This phenomenon is positive because increasing of the muscle activity lowers the probability of appearance of the blood stagnation zones and influences beneficially the vital activity of the stump as a whole preventing the atrophy of its tissues.

Integrated electric activity of stump muscles within the step period is shown in Table 2 .

Investigations showed that the mean electric activity of all the muscles of living extremity lowers though several muscles show higher activity. The implementation of rotation devices in prostheses does not cause additional energy losses on the side of living extremity, on the contrary - they decrease.

The experiments carried out confirm the assumptions made by previous investigators that the providing the prosthesis of lower extremity with additional mobility in horizontal plane may be a factor of significant normalization of gait.

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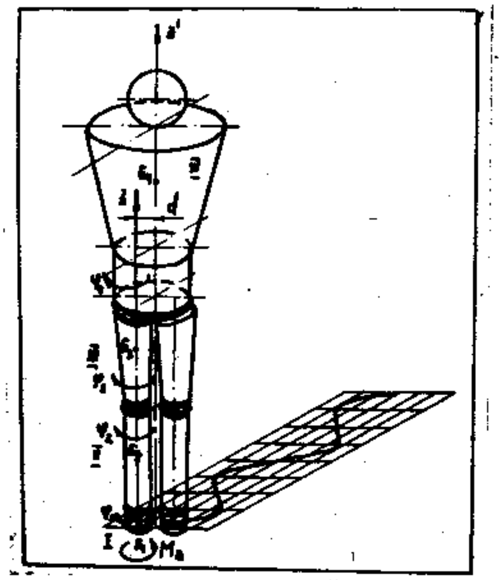


Fig. I. Model of human body segments motions in transversal plane.

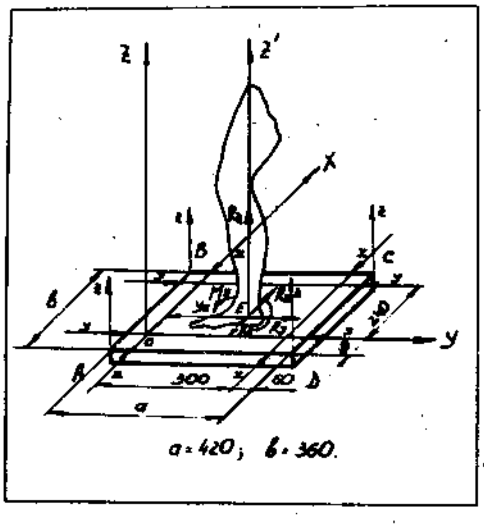


Fig. 2. Scheme of "force plate".

Fig. 3. Trajectory of the point in which the vector of ground-to-foot reaction is applied.

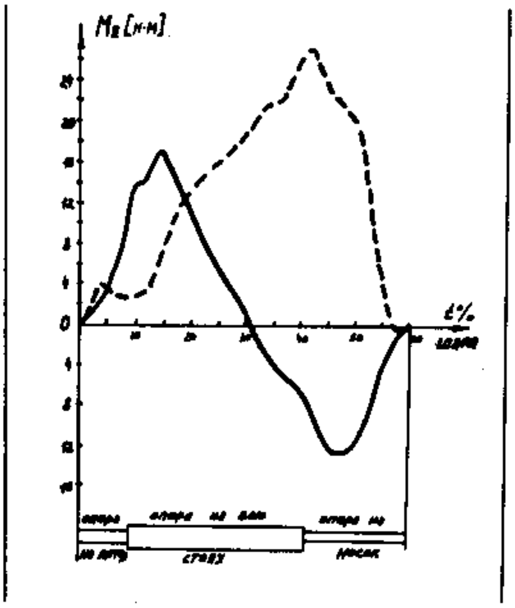


Fig. 4. Torque. Continuous line - normal gait, dashed line - amputee's gait.

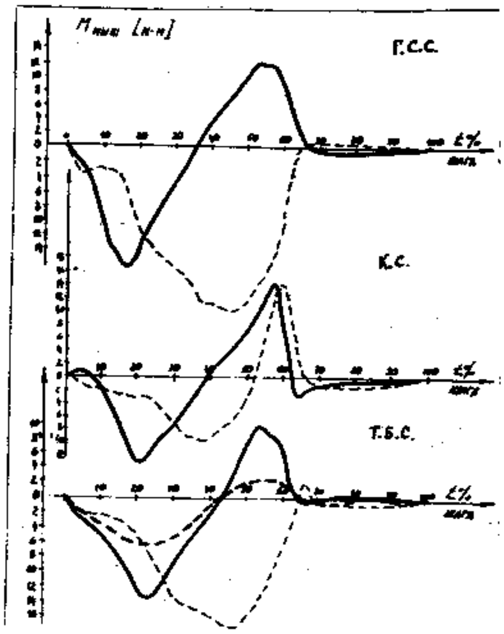


Fig. 5. Moments of muscle forces applied to the segments of lower extremity in transversal plane (up to down: foot, shank, hip).

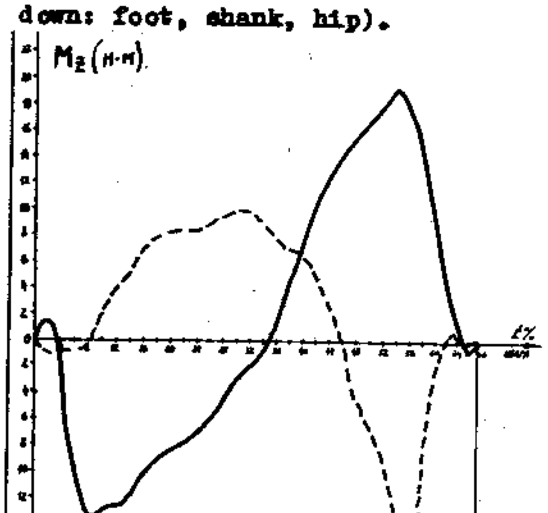


Fig. 7. Torque. Continuous line - unsevered extremity, dashed line - prostheses with rotator incorporated.

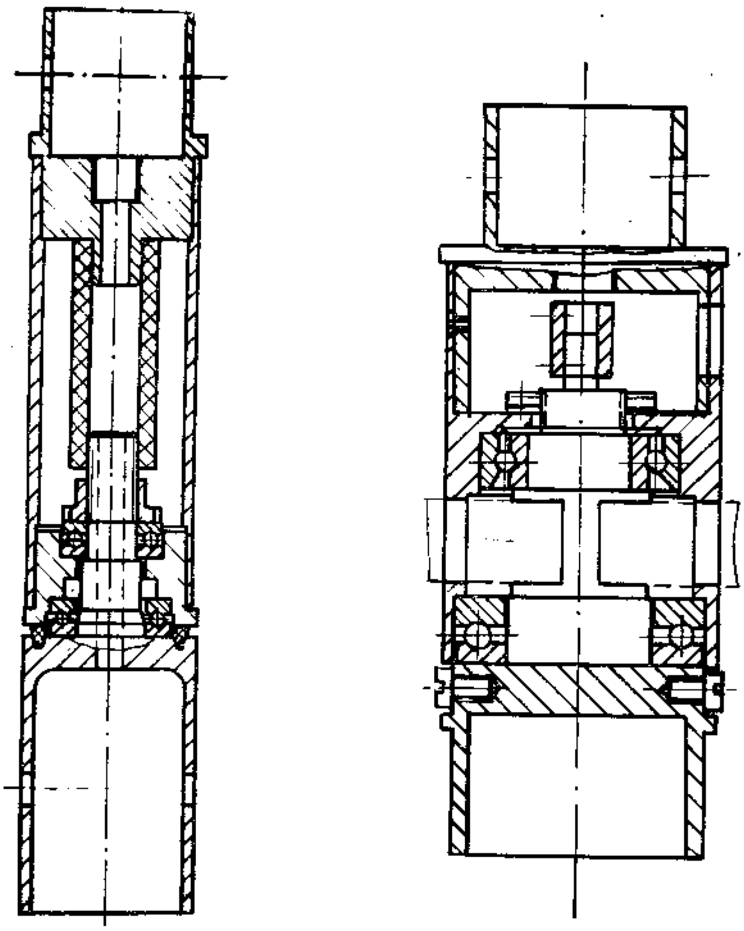


Fig.6. Devices with " passive rotation".
a) with rubber torsion, b) with metal plate.

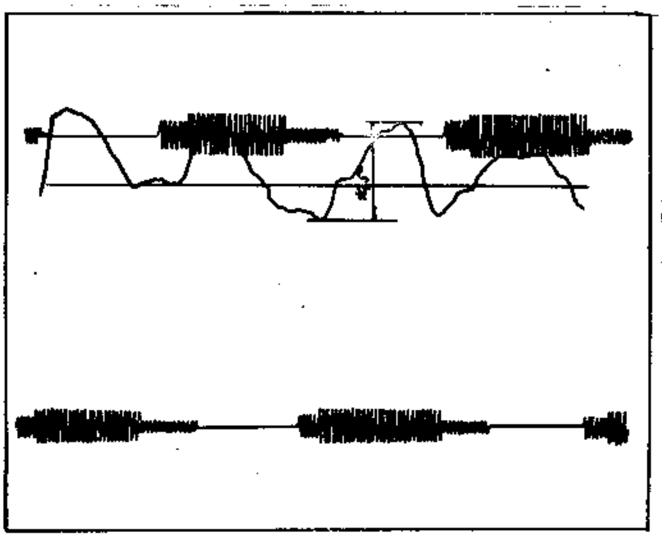


Fig.8. Shank rotation relative to the foot of the prosthesis.

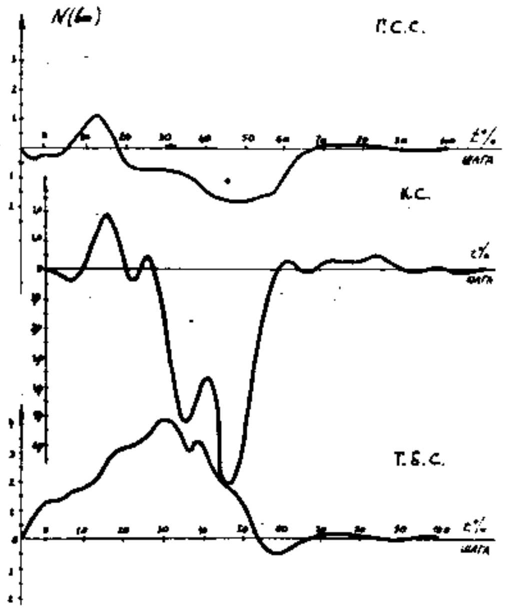


Fig. 9. Powers in walking on the prostheses with rotators incorporated (up to down: foot, shank, hip).

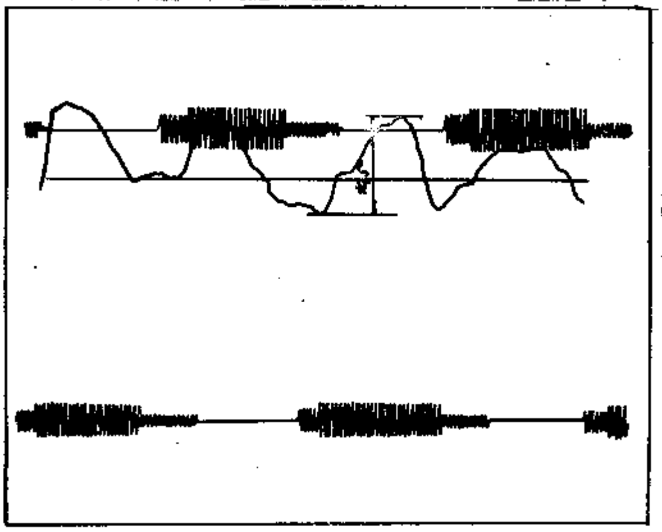
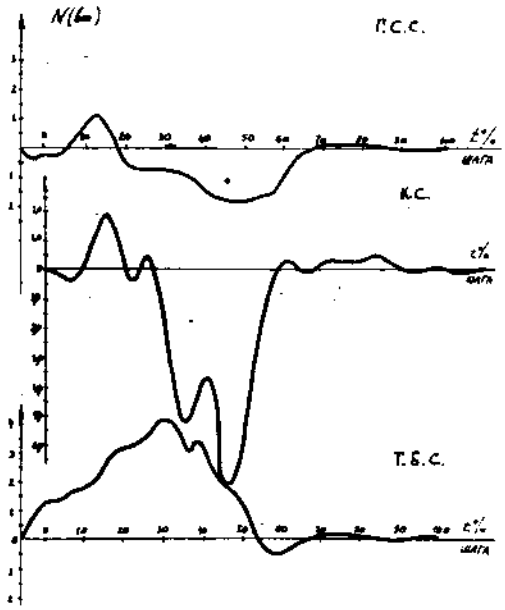


Fig.8. Shank rotation relative to the foot of the prosthesis.



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