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Modeling of fitting of below knee prosthesis with self-suspension socket

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Abstract
In this work, a mathematical model for investigating the fitting of self-suspension Below Knee (BK) prosthesis was attempted. The model is based on linkage kinematic and kinetics analysis of the residual limb segments as well as the stresses analysis at the stamp-socket interfacing. The prosthesis socket and the residual limb are approximated as combined two thick cylinders with shrinkage at interface. Lames equations of hoop and longitudinal stresses are applied with specific boundary conditions. The main parameters investigated in this model are; locomotive speed, swing phase of gait cycle, socket material, skin status and prosthesis weight. A map is constructed to investigate the effect of these parameters on prosthesis suspension fitting. The theoretical results are checked experimentally by using force platform and treadmill tests. A patient wearing BK prosthesis is participate in the experiments. KINOVA video analysis program is used for analyzing gait cycle. MATLAB Rb2013a program is used for solving the theoretical equations and plotting the results. It was found that; the suspension of the socket decrease as the walking speed increase within 1.5 to 12 m/s, skin sweating and dirty reduces the suspension force and increasing the risk of socket slippage, The more dangerous region of slip in socket occurs at mid- swing for the most of walking speeds and increasing the shrinkage fitting increase the suspension action.

Keywords: Socket suspension; Slip; Below knee prosthesis; Gait cycle; Force platform; Treadmill.

1. Introduction

Below knee prosthesis is an external device for assist amputee suffering from losing his part of limbs due to decease or accident. Best fitting of BK prosthesis leads to proper function and as a results improve the locomotive of the patient. The suspension system can be passive or active. Passive or self-suspension depends on the friction at socket-stump interfacing. In active suspension an additional fitting configuration are used such as; sleeve, belt, straps, or cuff and suction method [1]. However for normal uses self-suspension is widely used for fitting the prosthesis socket. Slip of socket can be occurred either due to improper socket design and fabrication or due to the effects of other external parameters related to the patient muscle activities and environment. The prosthesis can loose suspension if slip is just existed, in this case a relative motion between the socket and stump occurs. At this condition the prosthesis can lose its contact or at least some extent of discomfort and abnormal gait is noticed [2]. Many authors investigate suspension phenomena experimentally at rehabilitation period or after using the prosthesis. Francis et.al.

[3] conducted experiments to measure the extent of slip at various locations within above-knee suction sockets. The range of total slip values encountered in four subjects is given as a function of location within the socket, under conditions of level walking. Hossein G.et. al. [4] attempted a method for suspension to prevent slip for walking cycling and running. They made statistical investigation on nine patients at different ages. Charles H [5] investigated the effect of slip of BK prosthesis he tried new suspension method to prevent slip of prosthesis. A comparison study was made between three groups with different suspension types.

While experimental and statistical investigations are satisfactory to describe the suspension phenomena in literature, however theoretical modeling of suspension is seldom attempted. In this study a mathematical model with reasonable assumption is attempted. The model accomplished the effects of many parameters on passive suspension concerning with prosthesis, patient activities and residual limb status.

2. Mathematical model

Figure 1 shows a general position of locomotive of patent wearing below knee prosthesis in sagittal plane, while Figure 2 shows a schematic diagram of the model. The main purpose of the model is to analyze the dynamics of the prosthesis against the slipping between the stump of the residual limb. The present analysis is focused on the swing phase of walking or running gait since; at this period the prosthesis is free from ground contact and the probability of slipping becomes high. In this model the leg segments are considered as rigid body (linkage). The hip and knee joints are of hinge type joints. *CH* is the link between body center and hip joint, *HK* is the link represents the shank and, *KG* is the link between the knee joint and the center of the prosthesis. The coordinate system are located at the hip joint. The horizontal and vertical coordinates unit vectors are *i* and *j* respectively. The angles of hip and knee joints are θ_H and θ_K respectively, and measured from the horizontal axis they are positive at clockwise direction.

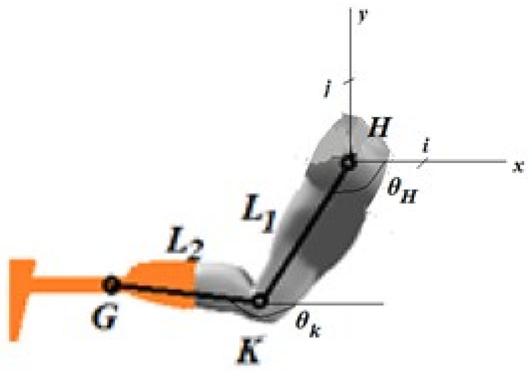


Figure 1. Locomotive motion analysis of BK prosthesis.

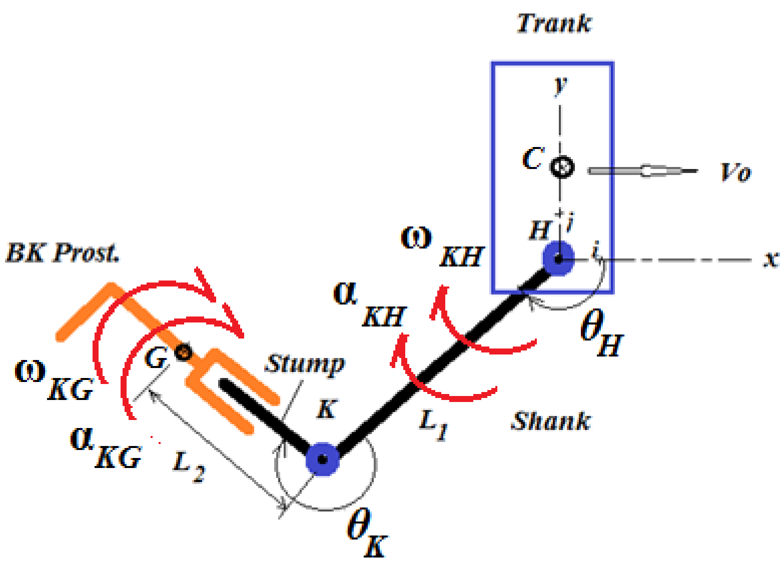


Figure 2. Schematic diagram of the model.

2.1 Kinematic analysis

The acceleration of the shank can be written as [6];

$$a_K = a_H + \alpha_{HK} \times r_{HK} + \omega_K \times (\omega_K \times r_{HK}) \quad (1)$$

where r_{HK}, ω_K and α_{HK} are the displacement, angular velocity and angular acceleration of the segment.

According to the coordinate notation shown in Figure 2, the space vector of the link *HK* take the following form;

$$r_{HK} = -\ell_1 \sin\theta_H\, i - \ell_1 \cos\theta_H\, j \quad (2)$$

Now by making use of equations (1) and (2) and assuming constant locomotive velocity where; $a_H=0$ the following equation is obtained;

$$a_K = \alpha_{Hk} k \times (-\ell_1 \sin\theta_H\, i - \ell_1 \cos\theta_H\, j) + \omega_{HK} k \times [\omega_{HK} k \times (-\ell_1 \sin\theta_H\, i - \ell_1 \cos\theta_H\, j)] \quad (3)$$

Noting that; *k* represent the unit vector of the angular velocity and acceleration which is positive at inward plane direction and can be found by using the right- hand role.

Due to the progress movement of the body center with velocity of V_o an additional angular velocity is added to the shank, hence the total angular velocity ω_t can be written as the follows;

$$\omega_t = \omega_K + \omega_o \quad (4)$$

where ω_o is the angular velocity due to the locomotive velocity of trunk, which can be evaluated as the following;

$$\omega_o = \frac{vo \cos\theta_H}{\ell_1} \quad (5)$$

Referring to Figure 2 the total angular velocity can be written as;

$$\omega_t = \omega_{HK} + \frac{vo \cos\theta_H}{\ell_1} \quad (6)$$

By making the successful cross vector product the acceleration at knee joint is;

$$a_K = -\alpha_{HK} \ell \sin\theta_H\, j + \alpha_{HK} \ell_1 \cos\theta_H\, i + \omega_t^2 \ell_1 \sin\theta_H\, i + \omega_t^2 \ell \sin\theta_H\, j \quad (7)$$

Now, for the lower segment *KG* (from knee joint to the mass center of the prosthesis) one can write for the acceleration at G the following expression;

$$a_G = a_K + (\alpha_{KG} \times r_{KG}) + \omega_{KG} \times (\omega_{KG} \times r_{KG}) \quad (8)$$

Substituting value of the component of the space vector into equation (8) giving;

$$a_G = a_K + \alpha_{KG} k \times (-\ell_2 \sin\theta_K\, i - \ell_2 \cos\theta_K\, j) + \omega_{KG} k \times [\omega_{KG} K \times (-\ell_2 \sin\theta_K\, i - \ell_2 \cos\theta_K\, j)] \quad (9)$$

Finally, substituting of equation (7) into equation (9) gives the following vector equation for the absolute acceleration at prosthesis mass center;

$$a_G = \alpha_{HK} \ell_1 \sin\theta_H\, j + \alpha_{HK} \ell_1 \cos\theta_H\, i + \omega_t^2 \ell_1 \sin\theta_H\, i + \omega_t^2 \ell_1 \cos\theta_K\, j - \alpha_{KG} \ell_2 \sin\theta_K\, j + \alpha_2 \ell_2 \cos\theta_K\, i + \omega_{KG}^2 \ell_2 \sin\theta_K\, i + \omega_{KG}^2 \ell_2 \cos\theta_K\, j \quad (10)$$

The horizontal and vertical components of the acceleration can be separated from equation (10) as the follows;

$$a_x = \alpha_{HK}\ell_1 \cos\theta_H + \omega_t^2 \ell_1 \sin\theta_H + \alpha_{KG}\ell_2 \cos\theta_K + \omega_{KG}^2 \ell_2 \sin\theta_K \tag{11}$$

$$a_y = -\alpha_{HK}\ell_1 \sin\theta_H + \omega_t^2 \ell_1 \cos\theta_K + \alpha_{KG}\ell_2 \sin\theta_K + \omega_{KG}^2 \ell_2 \cos\theta_K \tag{12}$$

It is to be stated here that; the angular displacement, velocity and acceleration of the different segments must be evaluated experimentally. In this regard, one of the most powerful tool for this measurement is by using video analysis technique such as KINOVA Analysis which is used in this work.

2.2 Kinetic analysis

The free body diagram of the BK prosthesis is shown in Figure 3: the main internal and external forces are located on the diagram namely; prosthesis weight W, parallel slip force F and the normal force N. From the kinematic analysis the two components of the acceleration a_x and a_y are located at the mass center of the prosthesis G. Now by applying Newton's second low at both x and y direction that is;

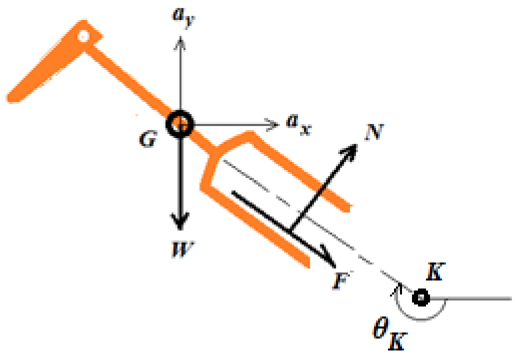


Figure 3. Prosthesis free body diagram.

$$\Sigma F_x = m a_x \tag{13}$$

$$\Sigma F_y = m a_y \tag{14}$$

And considering the free body diagram shown in Figure 3 and making use of equations (13and 14), the following equations will be obtained;

$$N \sin\theta_K + F \cos\theta_K = m_p a_x \tag{15}$$

$$N \cos\theta_K - F \,\mathrm{sins}\, \theta_K - W = m_p a_y \tag{16}$$

Equations (15 and 16) together with equations (11 and 12) can be solved to find the dynamic equilibrium force F which is essential for slip analysis.

2.3 Stress analysis (Approximated analysis)

Although the stump and socket are of irregular cylindrical shape and the pressure is distributed in contour regions at the contact area and the accurate analysis of stress needs to be achieved using finite element method, however for the theoretical purpose in which the effects of various parameters on the stress is the main interest an approximate analysis with reasonable accuracy can be considered. In this analysis it can be assume the socket and stump as two cylinders with average diameters and subjected to average pressure distributed on of socket – stump interface.

In Figure 4 the socket and stump of the residual limb is represented as two cylinders with shrinkage fitting causing pressure of Pi. The socket is treated as a thick cylinder so that Lama's equations for stresses can be applied; [7];

$$\sigma_H = A + B/r^2 \tag{17}$$

$$\sigma_r = A - B/r^2 \tag{18}$$

where A and B are constants depending on the pressure boundary conditions.

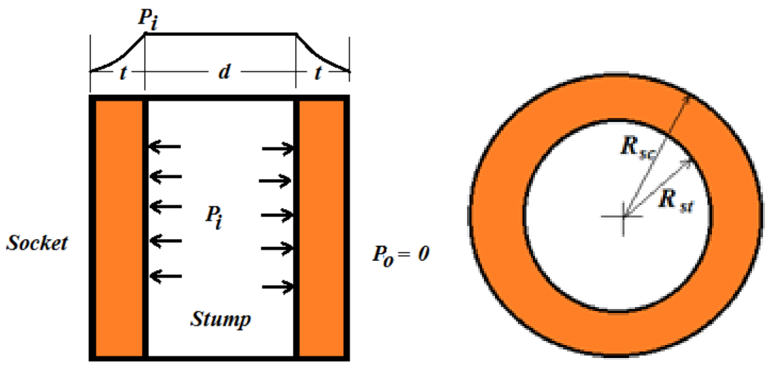


Figure 4. Stress analysis of socket and stump.

For the present case the conditions of the pressure are as the follows;

$$\sigma_r = -P_i \quad \text{at} \quad r = R_{st} \tag{19}$$

$$\sigma_r = 0 \quad \text{at} \quad r = R_{Sc} \tag{20}$$

Applying the pressure boundary conditions given in equations (19 and 20) on equations (17 and 18) lead to the following equations for the radial and hoop stresses;

$$\sigma_r = \frac{PiRsc^2}{(R_{Sc}^2 - R_{St}^2)}\left[\frac{r^2 - R_{Sc}^2}{r^2}\right] \tag{21}$$

$$\sigma_H = \frac{PiRsc^2}{(R_{Sc}^2 - R_{St}^2)}\left[\frac{r^2 + R_{Sc}^2}{r^2}\right] \tag{22}$$

The circumferential strain at radius r of the socket is;

$$\epsilon_{Ho} = -\frac{2\delta_{sc}}{2r} = -\frac{\delta_{sc}}{r} \tag{23}$$

The circumferential strain at radius r of the stump is;

$$\epsilon_{Hi} = -\frac{2\delta_{st}}{2r} = -\frac{\delta_{st}}{r} \tag{24}$$

The minus sign appearing in equation (24) is due to the decreasing in radius of the stump after the shrinking. Considering Hook's laws; the strain for the two cylinders can be written as the follows;

$$\epsilon_{Ho} = \frac{\sigma_{sc}}{E_{sc}} + P_i \frac{v_{sc}}{E_{sc}} \tag{25}$$

$$\epsilon_{Hi} = \frac{\sigma_{st}}{E_{st}} + P_i \frac{v_{st}}{E_{st}} \tag{26}$$

The total interference shrinkage allowance is the sum of the shrinkage of the two cylinders and as the follows;

$$\delta_{sh} = \delta_{sc} + \delta_{st} \tag{27}$$

Combining equations (21 to 27) gives the shrinkage allowance in term of the internal pressure between the socket and stump and as the follows;

$$\delta_{sh} = \frac{1}{E_{st}}(\sigma_{st} + v_{st}P_i) + \frac{1}{E_{sc}}(\sigma_{sc} + v_{sc}P_i) \tag{28}$$

When the internal pressure is known or alternatively if it calculated from the shrinkage allowance as given in equation (28), the friction force F_r between the stump and socket can be evaluated as the follows;

$$F_r = 2\pi\mu P_i r L_{sc} \tag{29}$$

The suspension of the prosthesis is highly depends on the friction force at the stump-socket interface. Based on the above analysis the suspension fitting can be checked by comparing the slip dynamic force of equilibrium F (given in equations 15 and 16) with the friction force F_r (given in equation 29),

An elaborate inspection of these equations tells that; the suspension of prosthesis depends on several parameter which can be summarize as the follows;

1. The state of stump skin; clean, dirty or sweetly as well as the socket contact surface which affect the friction coefficient and hence F_r.
2. The locomotive speed which affect the acceleration components of the prosthesis mass center and hence F.
3. The weight material and dimension of the prosthesis used which affect F
4. The shrinkage fitting of the prosthesis socket which affect the internal pressure and hence F_r.
5. The muscle activity of the stump which affect the interfacing pressure and hence F_r.
6. The activity of hip and knee joint which affect the acceleration and hence F. However, the body weight has no effect on prosthesis slip.

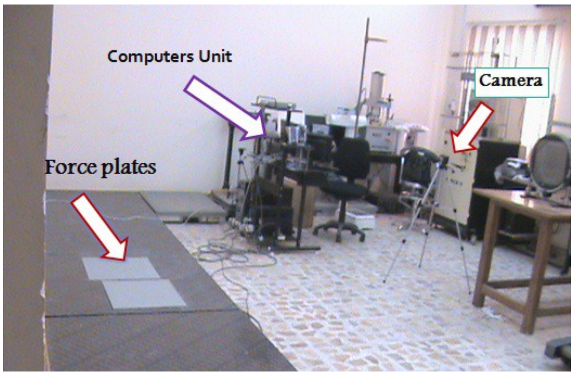
3. Experimental work

A patient (50 year old, 172 cm height and 62 kg mass) is suffering from amputee in his right leg due to accident was participated in the experiments. He use BK prosthesis with passive fitting suspension. The main components of the prosthesis are as shown in Figure 5, [8].

Figure 6 shows the force platform test in which two sensitive plate, high speed camera and computer unit are used. The schematic diagram of the force platform system is shown in Figure 6 also.

To measure the angles of leg joints at complete gait cycle (including stance and swing phases) markers are located at the Hip, Knee, Ankle and foot of the patient. The force platform test was achieved for several walking speeds and repeated three times for every case to allow for comfortable conditions of walking. Figure 7 shows the starting of gait cycle where the heel of the left leg of the patient is just strike the plate. To prepare videos analysis at different walking speeds (low, normal and running) and investigate the effect of walking speed on prosthesis suspension a treadmill is used with high resolution camera as shown in Figure 8. Segment angular positions and segments motion are calculated by using KINOVA program at different locomotive speeds.

Figure 5. BK prosthesis.



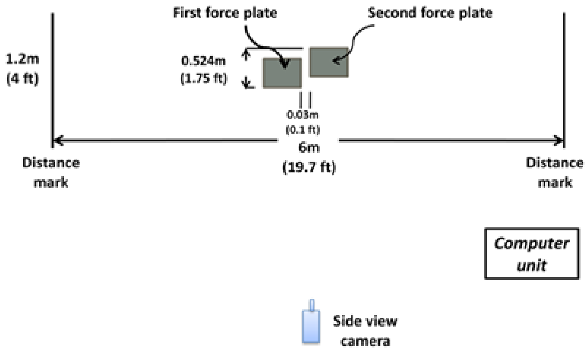


Figure 6. Force platform components and layout.

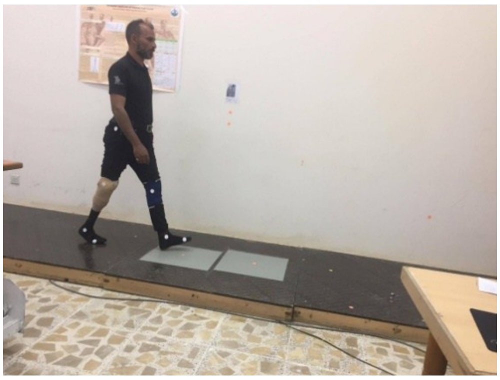


Figure 7. Force platform experiment.

Figure 8. Treadmill test.

4. Results and discussions

The experimental results of the force platform of the prosthesis leg are shown in Figure 9. In these figures the angular position of the hip and knee joints are plotted. In order to supply the kinetic data required for the mathematical model each curve is exported to MATLAB program to estimate the best fitting equation such as shown in Figure 10. As the equation of angular displacement is evaluated as a function of time the other kinematic data such as angular velocity and acceleration for both joints are available from the differentiations.

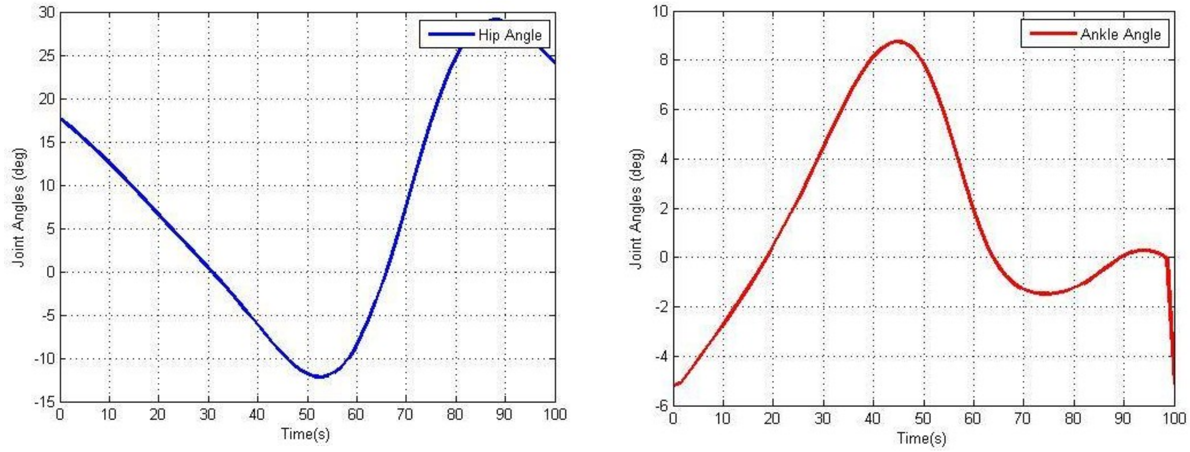


Figure 9. Experimental results of Hip and Knee (force platform test).

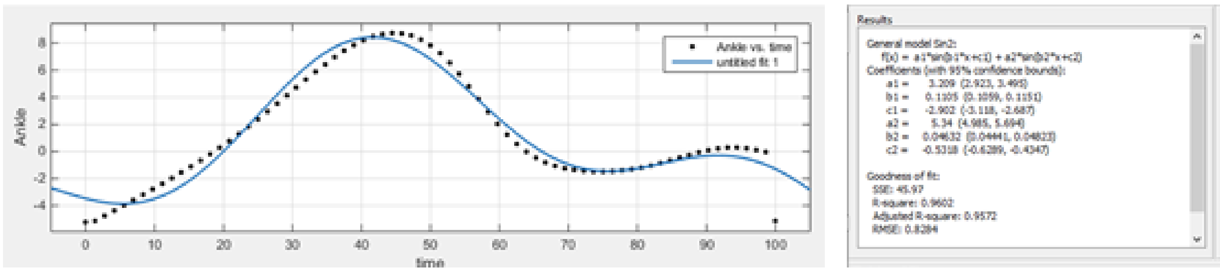


Figure 10. MATLAB Fitting equations of knee joint.

For investigating the effect of locomotive speeds on slip of residual limb, several videos are captures. Each video is analyzed by using KINOVA program in which many key images are extracted for sequential times periods and the angles are calculated. The results are collected in Excel tables as shown in Table 1.

Table 1. Excel tables for KINOVA analysis of Hip and Knee joints for residual limb.

Angles		
Key Image	Time	Value
1	1	0
3	3	0
5	5	0
6	6	1
8	8	1
10	10	0
12	12	0
15	15	0
17	17	7
20	20	11
22	22	14
25	25	22
27	27	30

Knee

Angles		
Key Image	Time	Value
1	1	62
3	3	64
5	5	72
6	6	77
8	8	78
10	10	83
12	12	84
15	15	90
17	17	95
20	20	100
22	22	99
25	25	110
27	27	104

Hip

By using the kinematic data and plotting of equations (15,16 and 29) via MATLAB a design chart is constructed for different values of locomotive speeds (low, normal and running) and different skin status (wet, dry and clean) for swing phase of gait cycle as shown in Figure 11. To examine the suspension state at any position of swing phase the skin status is specified first from which the red line is selected. The portions of blue curves under this line indicate that the prosthesis is able to suspend without slipping while at the above portion the prosthesis lose suspension and tends to slip. Since all the swing phase curves has

maximum peak at the mid swing and at these points the curves approach the regions of slip hence at mid swing represents the most dangerous region.
Examining the chart of Figure 11 indicates that; at low locomotive speed (1.5 m/s) the swing blue curve is totally laid under the red lines whatever the skin status is. So that the slip cannot occur at this speed. However with increasing in the walking speed from above 1.5 till 12 m/s reduce the suspension in a manner depending on the akin status.
To check the theoretical results of the model, some points are investigated experimentally as shown in Figure 12. In these experiments the speed of the treadmill are increased gradually until the socket is just begin to slip. The patient is asked to alarm at this condition. The skin is tested for three different cases namely: dry clean, coated with dust and exposed to water spray (wet).

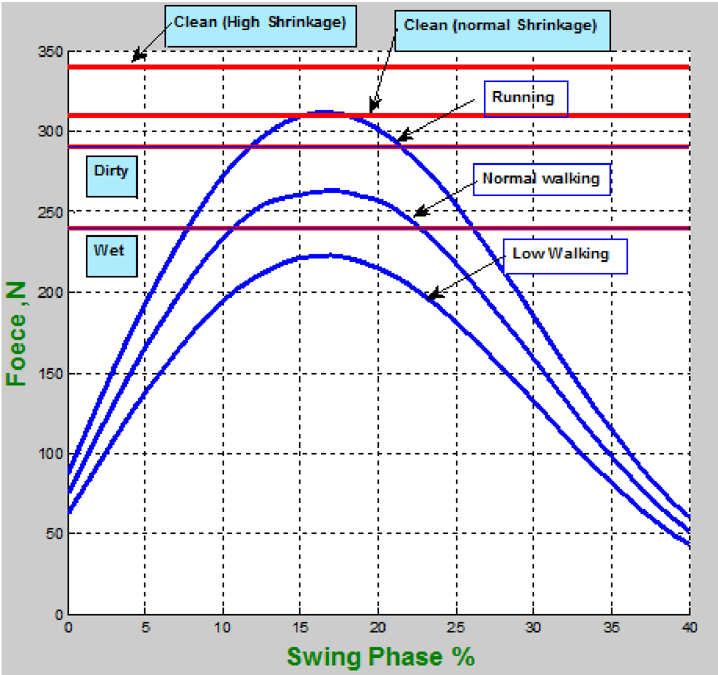


Figure 11. Suspension chart of BK prosthesis at different locomotive and skin status.

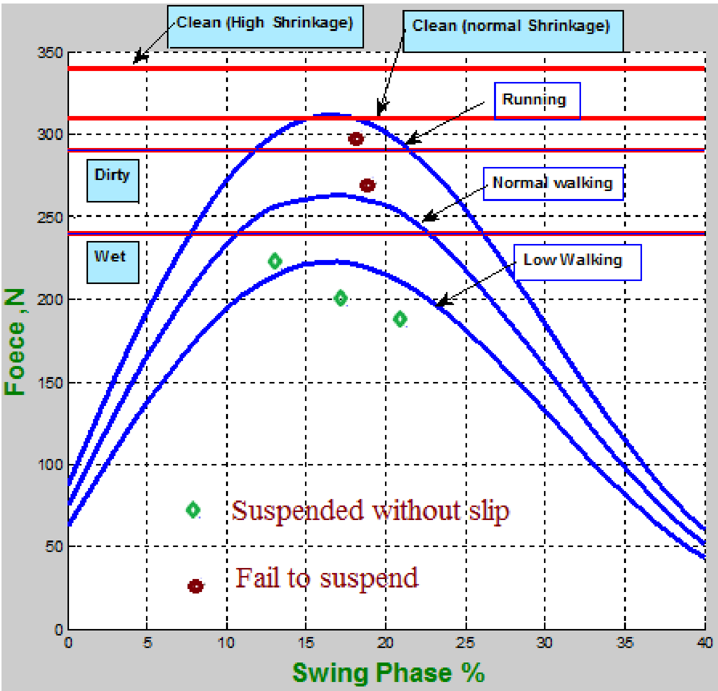


Figure 12. Experimental results of suspension.

5. Conclusions

From the theoretical and experimental results the following conclusions can be stated;

1. Slip of socket increases as walking speed increase from 1.5 to 12 m/s in a manner depending on skin status.
2. At mid- swing the more dangerous region of socket slip occurs for all walking speed
3. Skin sweating and dirty increase the risk of socket slippage
4. Increasing the shrinkage fitting make the prosthesis more suspended, however it is limited to the patient comfort.
5. From comparing the theoretical and experimental results it is clear that the presented mathematical model is succeeded to describe the suspension fitting and the effect of main parameters on it.

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