Prosthetics and Orthotics International, 1987, 11, 65–70

Bondgraph modelling and simulation of the dynamic behaviour of above-knee prostheses

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Abstract

A mathematical model was used to investigate the dynamic behaviour of an above-knee (AK) prosthesis in the swing phase and to analyse the influence of mass and mass distribution on the maximal stump load and the required energy. The model consists of a bondgraph model of the prosthesis and a "walking" model which predicts the walking velocity, step length and the femoral trajectory. Equipment was developed to measure the inertial properties of the components of the prosthesis.

Through computer simulation, stickdiagrams of the swing phase and graphs of the variation with time of the hip and stump forces were obtained. It was found that for a normal AK prosthesis with a knee-lock mechanism the axial stump load is greatest at the beginning and at the end of the swing phase. At a walking velocity of 5 km/hr the maximum axial stump load amounts to 2.1 times the static weight of the prosthesis.

The maximum axial stump force appeared to be almost directly proportional to the total mass of the prosthesis but independent of the mass distribution. The required energy also increased with the mass of the prosthesis but is dependent on mass distribution.

Because of their comparable weights the influence of the shoe is almost equal to the influence of the prosthetic foot. Thus lightweight shoes should be used with lightweight prosthetic feet in order to add to their advantages.

Introduction

Since the days of Ambroise Paré (1580), the history of the development of lower limb prostheses shows deep concern for the problems of mass and mass distribution. The famous prosthesis of Paré had a total weight of 7 kg and was made of steel. In the 17th century, wood and leather became the most important materials in prosthetics, later followed by aluminium. The success of the these light and strong materials can be concluded from their frequent use until this day. Because of the rapid development of modern plastics in the last half century, it is nowadays possible to make a complete AK prosthesis with a total weight of less than 2 kg. All manufacturers of prosthetic components are developing their lightweight prostheses, usually based on modern materials such as titanium, "aircraft" aluminium alloys and carbon fibre reinforced plastics. All these new designs satisfy some standards for the required strength and stiffness, e.g. the Philadelphia Standards (ISPO, 1978). It is remarkable that there are as yet no standards for the optimal mass of lower limb prostheses. The final mass is the result of the available materials, rather than conforming to well-understood design criteria. The aim of this research is to establish scientific criteria for the optimal mass and mass distribution.

In general there are two approaches:

Experimental It is possible to evaluate walking patterns of amputees wearing prostheses of varying mass. However, in this way it would be difficult to do serious experiments with prostheses with a much lower weight than the currently available types. Therefore, it was decided to approach the problem in a theoretical way.

Theoretical A method, was developed based on bondgraphs, to obtain mathematical models of the dynamic behaviour of lower limb prostheses. At this moment the authors have models for the swing phase of locked and unlocked AK prostheses. A model for the stance phase is in preparation.

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The subject of this paper is the model of the swing phase of a locked knee AK prosthesis. The influence of the mass and mass distribution on the resulting stump load and the required energy was investigated.

The model incorporates two submodels:
1. A bondgraph model of the prosthesis.
2. A model which describes the swing phase.

Bondgraph model of a locked-knee AK prosthesis

For an analysis of the dynamic behaviour the prosthesis was conceived as the physical system represented in Figure 1. The mathematical description is normally presented in differential equations. One of the main disadvantages of this approach is the impossibility of changing the system, e.g. by adding a knee control mechanism, without the laborious derivation of new system equations. Bondgraph description is a way to overcome this problem. In modelling and simulation with bondgraphs it is not necessary to derive equations. The physical system is expressed graphically and this graphical description is fed into a bondgraph-simulation-computerprogram. Because of the unfamiliarity of most readers with bondgraphs, the prosthesis-bondgraph will not be discussed further. Purely as an example of the compactness of this method, Figure 2 represents the bondgraph scheme of a 4-axial hydraulic damped knee mechanism.

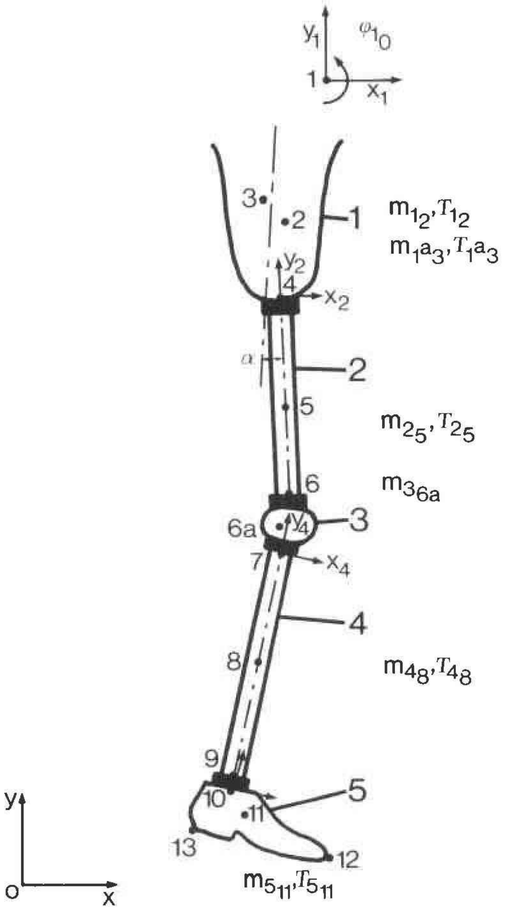


Fig. 1. The prosthesis conceived as a dynamic system.

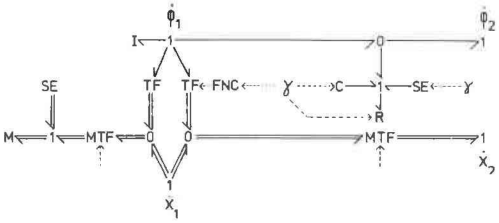


Fig. 2. Bondgraph of a four axial hydraulic knee.

Derived in a simple way, the bondgraph model is in fact identical with the set of Lagrange differential equations of the system. There are two ways to calculate the dynamic behaviour:
1. Using forces as input parameters and calculating the movements of the system. In practice this is rather difficult because it would be necessary to change the input forces until a normal swing phase was obtained.
2. Using the described movements of the prosthesis as input parameters and calculating the required forces, work and energy.

This second way was considered to be very useful. Of course it raises the question as to which movements are desired. The answer is the developed "walking"-model which predicts the optimal movements of a locked-knee AK prosthesis in swing phase.

The "walking" model

It would be possible to record (e.g. with Selspot equipment) the movements and walking patterns of amputees and to use these data as input for the prosthesis model. In this way it would be possible to calculate the required energy and the resulting stump load in particular cases. However, the aim was to investigate the influence of mass and mass distribution as a design criterion. Therefore, it was decided to use a model of "ideal walking". The calculated forces, stump load, work and energy must be conceived as values which are necessary to perform a normal walking pattern with an AK prosthesis. Of course, no amputee has a normal walking pattern and they would not produce these predicted forces, work and energy.

Figure 3 represents the swing phase from toe-off until heel contact.

The model incorporates the following experimental facts (Inman, 1980):

1. Swingtime in relation to the walking cycle frequency N.
2. Thigh angle at toe-off and heel contact in relation to N.
3. Total pelvic rotation.

The following assumptions were made:

4. The walking pattern is symmetrical and regular.
5. The rotation of the thigh and pelvis can be described with goniometric functions.
6. The trajectory of the femur can be approximated by a 3-degree polynome.

The model predicts:

7. The walking velocity, stride and step length in relation to N.
8. The trajectory and velocity of the femur as a function of time and in relation to N.

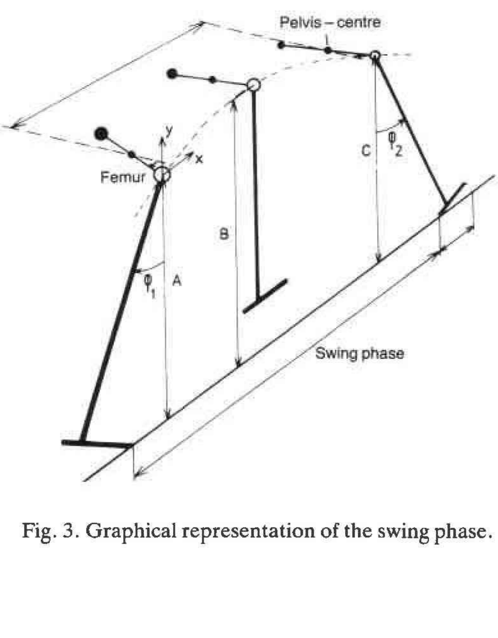


Fig. 3. Graphical representation of the swing phase.

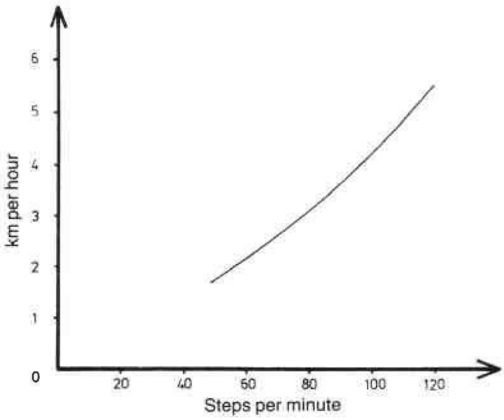


Fig. 4. Predicted walking velocity in relation to the step frequency.

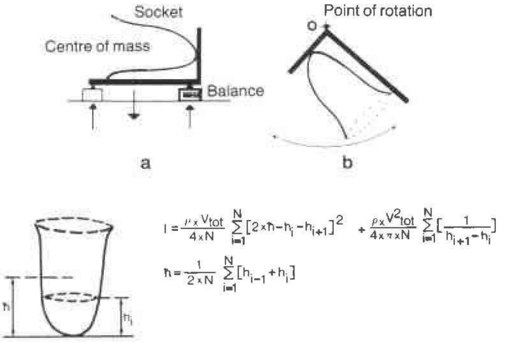
Verification of the "walking" model

Figure 4 represents the predicted velocity in relation to N. The walking velocity is too high at low frequencies and too low at the highest frequencies. However the overall inaccuracy of the model is less than 10%, which is rather good for a not very complex model. Improvements—better values for the parameters and extension from 2-dimensions to 3-dimensions—will raise the accuracy.

Parameters

To calculate the dynamic behaviour of the prosthesis, information is needed about the dynamic properties of the components of the system. Therefore, equipment was designed for measuring the first and second moments of inertia.

The first moment of inertia can be measured with a moment-equilibrium-table, illustrated in Figure 5a. With an oscillation time-measurement, the second moment of inertia can be established, as shown in Figure 5b. With these devices, the dynamic properties of all prosthesis components can be established with an error of less than 5%. Because the stump is part of the dynamic system, moments of inertia must be measured too. Obviously this is not possible with the equipment already mentioned. When it is assumed that the stump fills the socket entirely and an assumption is made for the relative weight of the stump (Drills et al, 1964) the mass and the moments of inertia can be calculated from the level heights when filling the socket with small equal amounts of water (Fig. 5c).



$$I = \frac{\rho \times V_{tot}}{4 \times N} \sum_{i=1}^{N} [2 \times \bar{h} - h_i - h_{i+1}]^2 + \frac{\rho \times V^2_{tot}}{4 \times \pi \times N} \sum_{i=1}^{N} [\frac{1}{h_{i+1} - h_i}]$$

$$\bar{h} = \frac{1}{2 \times N} \sum_{i=1}^{N} [h_{i-1} + h_i]$$

Fig. 5. Equipment for measuring the dynamic properties of the components of the prosthesis (see text).

Results

Figure 6 shows stick diagrams of the calculated swing phase at two different walking velocities. The defined direction of the hip forces is also indicated.

The stick diagrams were obtained by simulation of the combined "walking" model and the bondgraph model, fed with the inertial properties of a normal AK prosthesis. This prosthesis consisted of a rather long socket, a Böck locked knee, pylon and uni-axial foot. The total weight of the prosthesis was 3.9 kg. Figures 7, 8 and 9, show the pattern of the hip forces which must be applied to the prosthesis to obtain this swing phase.

The axial force is particularly interesting; the tangential and rotational hip forces are rather

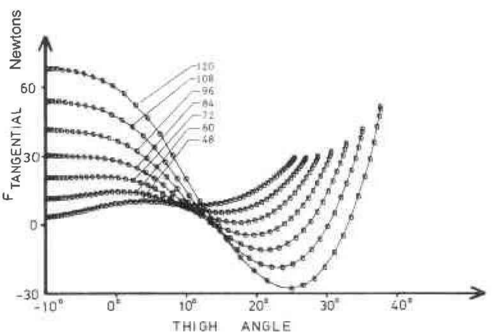


Fig. 7. Tangential hip force in relation to the thigh angle at seven step cycle frequencies.

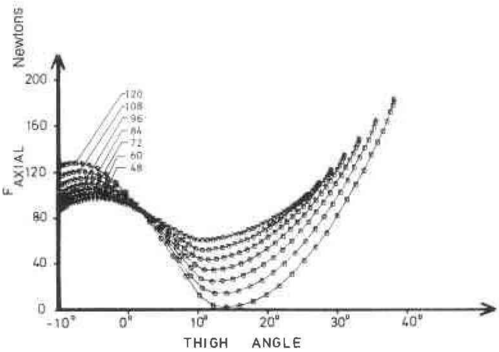


Fig. 8. Axial hip force in relation to the thigh angle at seven step cycle frequencies.

moderate forces, but the axial hip force reaches high values. Therefore, the real axial stump force between stump and socket was investigated, acting as a shear force on the

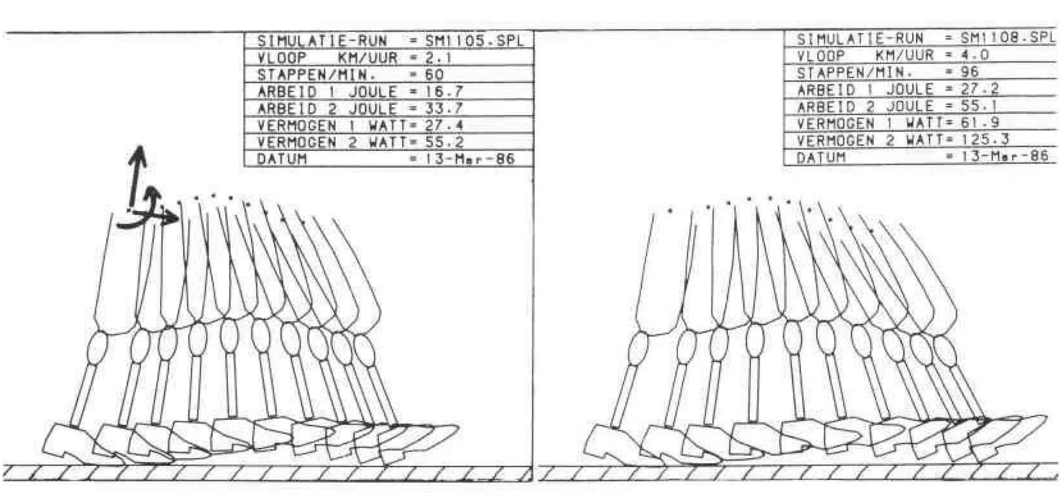


Fig. 6. Stick diagrams of the predicted swing phase at two velocities.

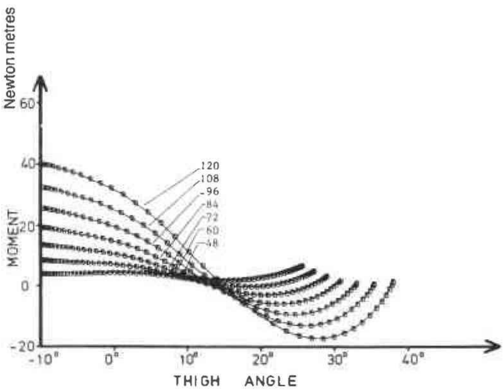


Fig. 9. Moment at the hip in relation to the thigh angle at seven step cycle frequencies.

stump's surface (Fig. 10). At higher frequencies, this force reaches values which are certainly not comfortable and probably impossible to endure.

To investigate the influence of mass and mass distribution on the maximal axial stump force, simulations were made for the already mentioned prosthesis, but also for the same prosthesis with a massless foot, a massless shoe, massless knee mechanism, massless pylon and a massless socket. In this way, the boundaries of "lightweight" prostheses were explored.

Numbering of the prosthesis and its variations:

Nr.	Weight in kg.	Remarks
1	3.9	Normal prosthesis
2	3.6	Massless pylon
3	3.3	Massless shoe
4	3.13	Massless knee mechanism
5	3.10	Massless prosthetic foot
6	2.5	Massless socket

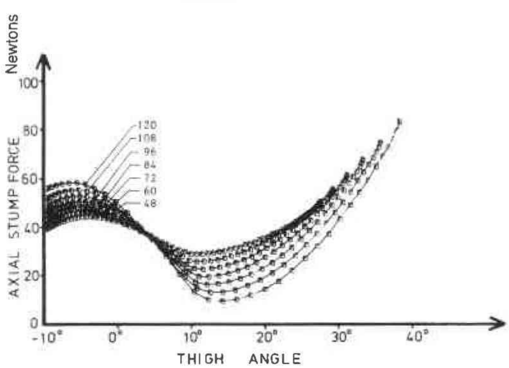


Fig. 10 Axial stump force in relation to the thigh at seven step cycle frequencies.

Figure 11 shows the maximum axial stump force that occurs in relation to the walking velocity for the prosthesis and its variations.

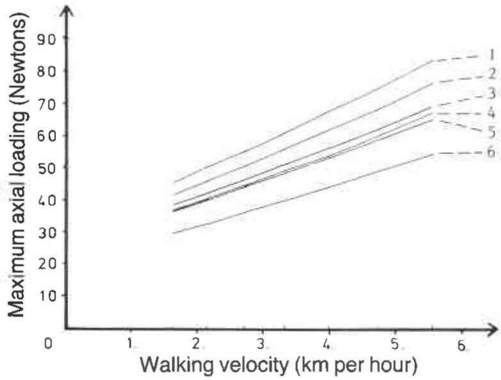


Fig. 11. Maximum axial stump force in relation to the walking velocity for an AK prosthesis with mass variations.

Some interesting results can be seen in this figure:

1. The maximum axial stump force is proportionate to the walking velocity.
2. The maximum axial stump force decreases almost linearly with the total mass of the prosthesis. Therefore, the lowest stump load in the swing phase of a locked knee AK prosthesis will occcur when the total mass is as low as possible.

   (This statement holds only for a locked knee prosthesis, because in the case of an unlocked knee, the femoral trajectory will be different).
3. The influence of the mass of the shoe is quite comparable with the influence of the mass of the knee mechanism and with the influence of the mass of the prosthetic foot. Therefore, lightweight prosthetic feet require lightweight shoes!

Figure 12 represents the total mechanical power put into the prosthesis in the swing phase, calculated for the prosthesis and for its variations, in relation to the walking velocity.

The figure shows the strong influence of the walking velocity on the required power in the swing phase. It can be seen that the required power is not completely linear with the total prosthetic mass, because of rotational effects.

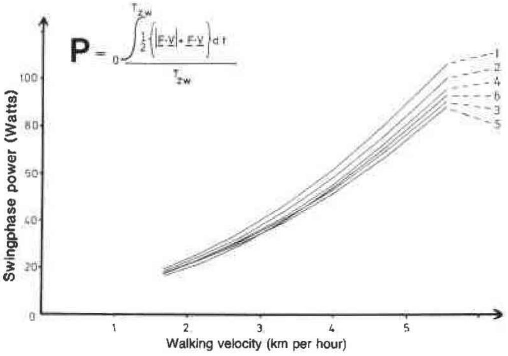


Fig. 12. Required swing phase power in relation to the walking velocity for an AK prosthesis with mass variations.

A massless shoe or a massless foot lead towards a lower required power than a massless knee mechanism, because of their greater distance to the point of rotation.

Discussion

1. The simple "walking" model gives rather good results for the walking velocity and the step length in relation to the step frequency.
2. The required power and the maximum axial stump force occurring in the swing phase of an AK prosthesis increases with the walking velocity.
3. The influence of the shoe is almost equal to the influence of the prosthetic foot on the maximal axial stump force and on the required power.
4. The required power and axial stump load decreases when the total mass of the prosthesis is decreased.
5. This theoretical approach proves to give useful information about the influence of mass and mass distibution on the swing phase of a locked AK prosthesis.

REFERENCES

DRILLIS, R., CONTINI, R., BLUESTEIN, M. (1964). Body segment parameters—a survey of measurement techniques. *Artif. Limbs* **8 (1)**, 44–46.

INMAN, V. T., RALSTON, H. J., TODOL, F. (1981). Human walking. Baltimore: Williams and Wilkins.

INTERNATIONAL SOCIETY FOR PROSTHETICS AND ORTHOTICS. (1978). Standards for lower-limb prostheses—report of a conference 1977. Philadelphia: I.S.P.O.