A METHOD OF CONTROL FOR THE ABOVE-KNEE-PROSTHESIS

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Summary

The purpose of our study is to introduce active functions and to show the feasibility of their use in an above-knee prosthesis. The control of a prosthesis is done separately in the stance-phase and the swing-phase during one walking cycle. In the former, the compensation of stability is carried out by locking the knee. In the latter, the sequential control is carried out corresponding to the surrounding conditions such as a flat ground, ascending and descending stairs or ramps. And then, the control signals are given by the switches attached on the sole and the knee mechanism. This prosthesis is operated pneumatically by compressed CO_2 *gas.*

1. Introduction

The conventional above-knee prosthesis poses various problems in that the knee joint has no function of controlling the leg. In practicular, walking on a slope and stairs becomes difficult. In addition, studies on the above-knee prosthesis are presently made principally for walking on level ground, with walking on a slope or a staircase being not studied so much. One of the reasons may be said to be prepossession with the conventional view that it is difficult to provide an above-knee prosthesis with active function. The aim of this paper is to suggest how the function of a prosthesis can be improved by providing it with external power.

To provide an above-knee prosthesis with an external power, one should solve the following three questions:

(i) Control system.

(ii) Control equipment.

(iii) Energy source.

In this paper the questions (i) and (ii), above, will be discussed.

2. Control system

A person equipped with a prosthesis forms a typical man-machine system. Under this system, the prosthesis as the machine should be so designed that is works well as a part of the human body, but to that end it should be studied in conjunction with the movement control mechanism of the human body.

The movement control mechanism of the human lower limb is controlled by a triple hierarchy comprising brainstem-spinal level, limb level, and neocortical level.

The brainstem-spinal level (lower level) is a muscle regulating level to perform reflex movement and in a coordinate movement to be carried out almost unconsciously like walking. The limb level (middle level) also participates. But, in a movement to be conducted fully consciously as in the case of doing fine movements, the neocortical level (upper level) participates and provides adaptation quite well. Of the above mentioned, most closely related to the walking movement is the middle level. The coordinate movement at this middle level is conducted in accordance with the so-called "learned in the course of individual development". Therefore, as for a control system for an externally powered prosthesis, it is most desirable from the aspect of the equipper's mental burden to apply the said learning algorithm without modification. That is, it is desirable to make use of the efferent nerve system and the afferent nerve system corresponding to an amputated region, but a direct use of them is difficult at the present stage. Accordingly, such a control system for an externally powered prosthesis that minimizes effects on the learning algorithm will be employed.

The movement of the lower limb and in particular, the walking movement consists, as can be seen from the analysis of walking, of coordinate movements nearly fixed in form with respect to the environment (level ground. slope, staircase). It can be considered that the following transformation will take place in the human body when reproducing a coordinate movement of an amputated region in using an artificial effector. That is, in the original learning algorithm F of the walking movement, a transformation between the body algorithm G concerning the effector and the artificial algorithm G' will be made:

$$G' = A(G)$$

The transformation A between the body algorithm G and the artificial algorithm G' is made at the upper level based on information via the contact point between the human body and the machine, learned with time, and incorporated into the algorithm of the middle level, with the original learning algorithm F being rendered into a new algorithm F'.

$$G\ F \rightarrow G' \subset F'\ (t \rightarrow t_o)$$

In this case, the time required for transformation from F to F' is related to a mental burden of the equipper and it is considered that the larger the burden, the more the expended time to and the more advanced the conversion from F to F'. As for a control system for an externally powered prosthesis, A and G' should be so selected as to minimize the mental effort. For instance, in A, an exchange of information between the equipper and the prosthesis plays a leading role, to which a positive method of applying electric stimulus, etc. and a passive method of depending solely on stocket equipment information are applicable. Similary, various methods are available for selection of A and G', but the clinical merits and demerits when applying them to a prosthesis are not yet determined. Such being the case, for the present prosthesis, a passive method through socket equipment information only as the conversion A was employed.

From the viewpoint that the human walking action is coordinate movements nearly fixed in form with respect to the respective environments and therefore the coordinate movement of an amputated region can be reproduced by sequence control, the algorithm G' has been determined with the environment as described below.

The human walking action can be divided largely into four patterns, flat, down-hill, up-stairs and down-stairs (Table 1).

Table 1. Changes in knee joint condition during walking

Environment	Swing Phase	Stance Phase
Flat and down hill	Knee joint stretches fully directly before heel contact	Small in angular change
Up-hill	Knee joint cannot stretch fully, with the max. flexion angle being approx. 650.	Stretching appears with an increase in hill gradient.
Up-stairs	Knee joint cannot stretch fully, with the max. flexion angle being approx. 900.	Stretched.
Down-stairs	Knee joint cannot stretch fully, with the max. flexion angle being approx. 900	Flexed.

Up-hill and up-stairs present nearly the same pattern except the degree of knee flexion angle in the swing phase and therefore, for the purpose of the present study the environment was divided in three classifications or flat and down-hill, up-hill and up--stairs, and down-stairs.

Knee flexion in the stance phase must be prevented to preserve stability. The need of compensation therefore applies equally to the respective environments. That is, a control system in the stance phase is so designed as to enable the knee to lock in the direction of flexion and to move freely in the stretching direction. As the control signal, a signal from a switch fitted to the solo is applied. A control system in the swing phase should be varied with the environment. Switchover for that purpose is effected by means of a switch provided on the waist.

For walking on the flat and down-hill, a system of sequence control of knee moment is employed to facilitate adaptation to the walking speed. That is, changes in knee moment of the human body are first determined by simulation with regard to the legs in the swing phase as a double pendulum, and approximated in the form as shown in dotted lines in Figure 1. Because such changes in knee moment depend on the walking speed, the three walking speeds were recognized,high, middle and low (Fig.2). They are compared roughly with typical knee moments at these cadences. In the meantime, for an input signal corresponding to walking speed, stance time is applied.

Because the knee flexion angle is large when walking up-hill and on the stairs,a sequence control of the angle is required. The walking speed at this time is constant. The sequence control signal is given from a limit switch fitted to the knee.

When descending the stairs, it is difficult to successively bring the knee in the stance phase into the direction of flexion and therefore, it is left free as before without any special control, with joints being moved passively.

The control system described above is employed in the present study.

3. Control Equipment

When the weight, output and practicality are considered, compressed gas is found suitable as a power saurce for prostheses.

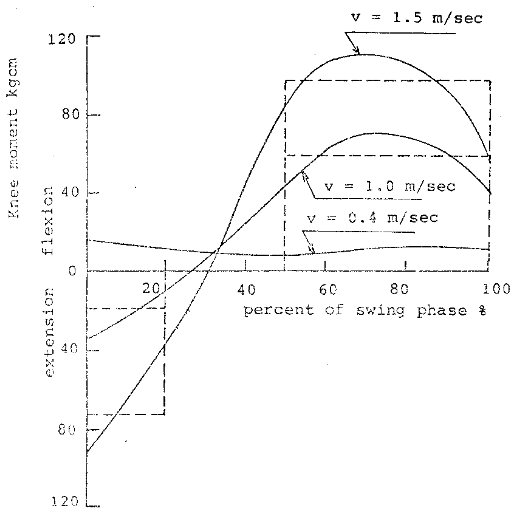

Fig. 1. Knee moment during the swing phase (level walking).

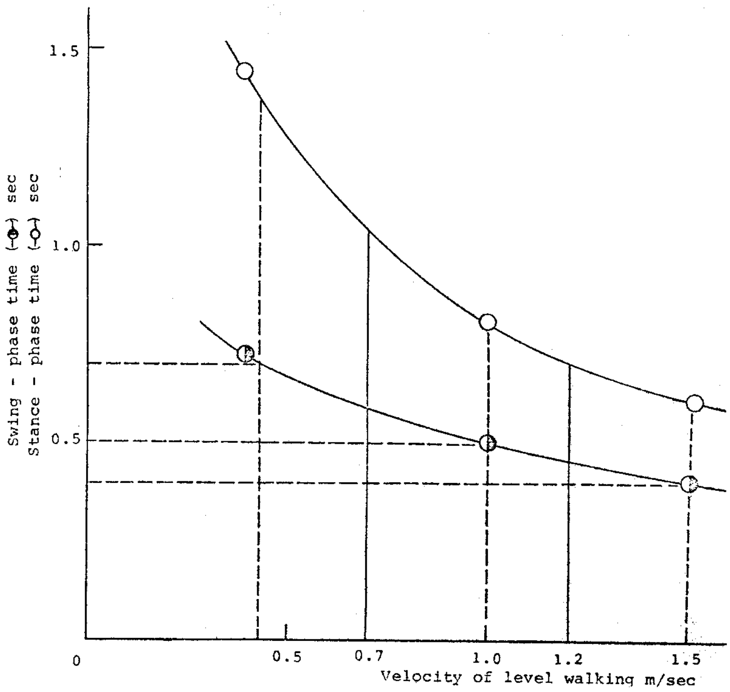

Fig. 2. Relation between velocity of level walking and stance-phase time, and swing phase time.

As for the actuator, in point of performance, a direct-acting cylinder is excellent, but when applied to prosthesis, the one from which direct revolving torque is obtainable a pouch actuator is preferable with respect to fitting, weight, etc. Therefore, a multicell artificial muscle developed in the author's laboratory was employed for the prosthesis (Fig.3). The static charateristics are shown in Figure 4.

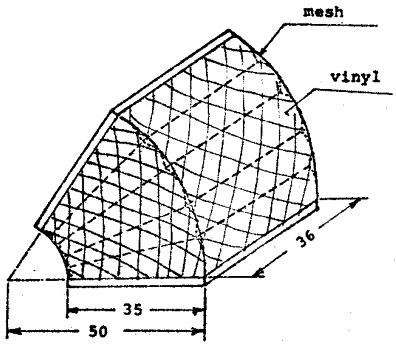

Fig. 3. Multi-level artificial muscle.

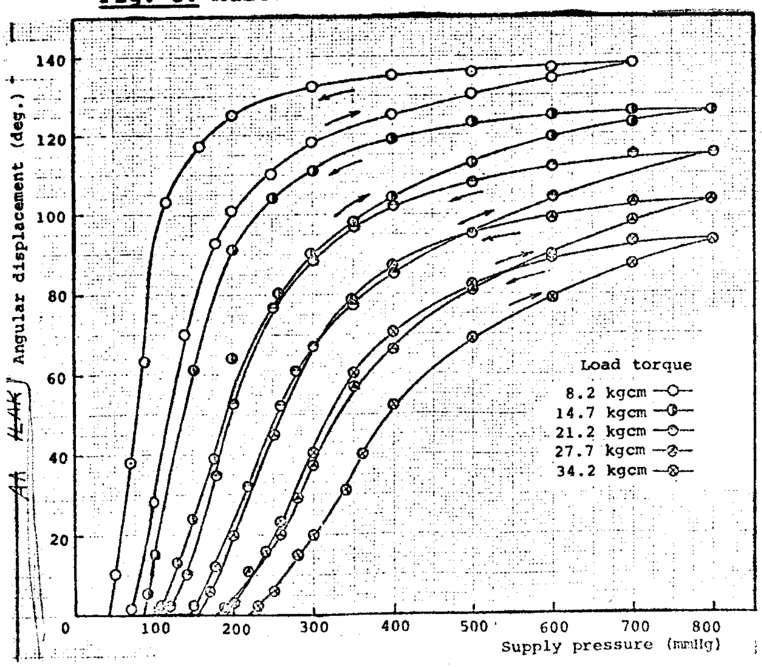

Fig. 4. Angular displacement charateristic curves for the artificial muscle.

The actuator can be controlled by a combination of an on--off valve and a flow control valve as shown in Figure 5. Figure 6 shows an integrated actuator control circuit employing on--off and flow control valves. The circuits V_1 through V_4 in Figure 5 are integrated. In the meantime, as for the on-off valve

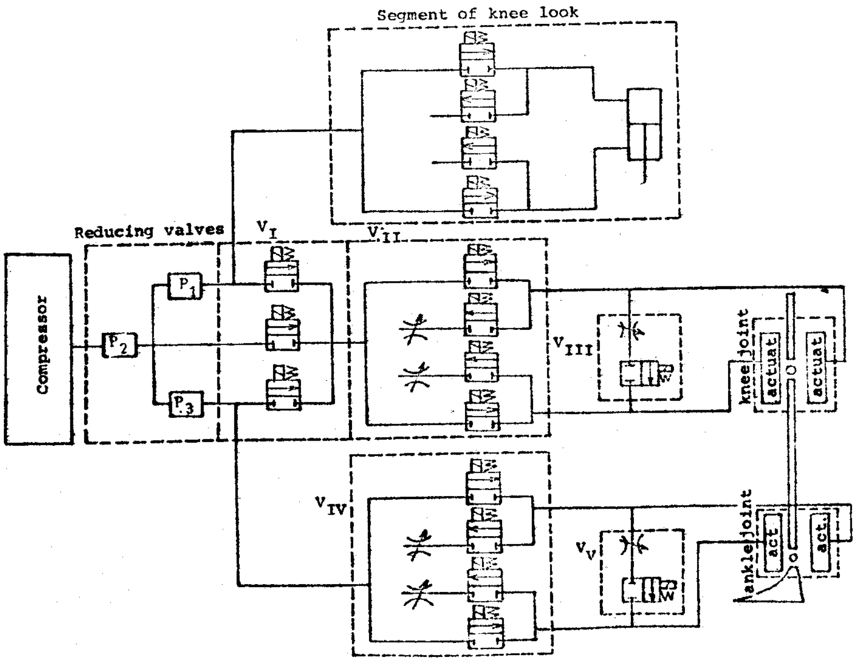

Fig. 5. Pneumatic circuit.

employed here (Fig.7), its operating characteristic is determined by the electromagnet used and its differential pressure flow characteristics, by the thickness of rubber, circuit diameters (a, b, c) and plunger displacement. Figure 8 shows the differental pressure flow characteristic with rubber thickness of 0.5 mm, a = b = 2.6 mmϕ and c = 1 mm. The response time of the electromagnetic valve,solenoid voltage, and power consumption were 20 ms, 6 V D.C. and 0.9 W, respectively.

The overall block diagram of the experimental prosthesis and the overall system diagram are shown in Figures 9 and 10.

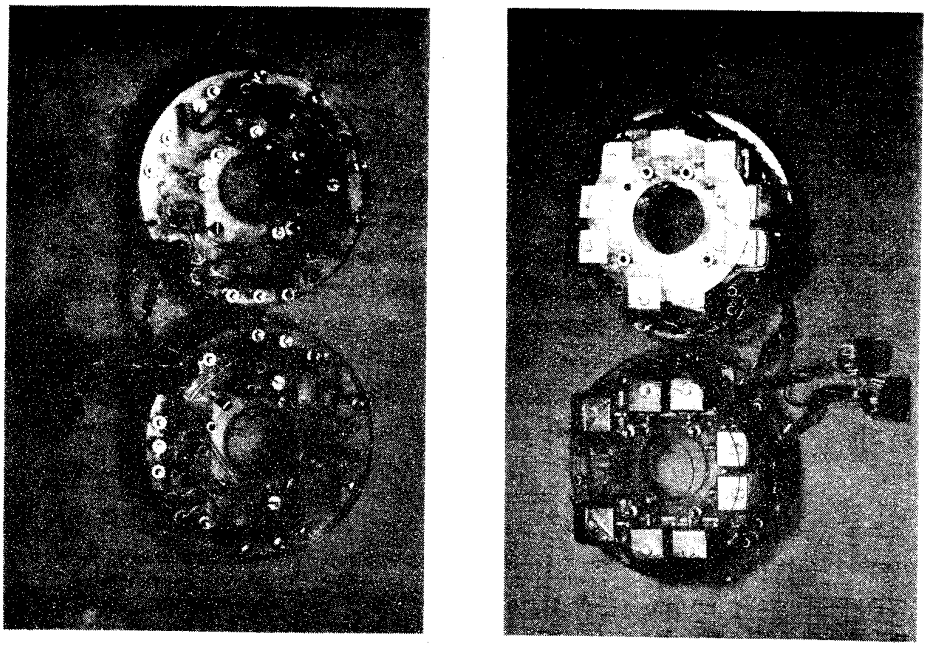

Fig. 6. Integral pneumatic circuit.

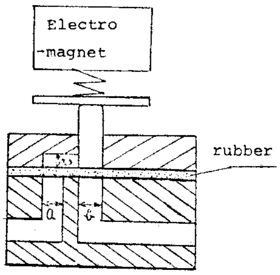

Fig. 7. On-off valve.

4. Clinical Experiment

As a result of the clinical experiment on level ground using this externally powered prosthesis, the gait was almost similar to walking with a pylon prosthesis. The reason for the expected gaitsnot being obtained is deemed to consist principally in that stretching moment was given to the knee at the acceleration period in the swing phase and the time constant of the entire prosthesis system was large (Fig.11), but no definite

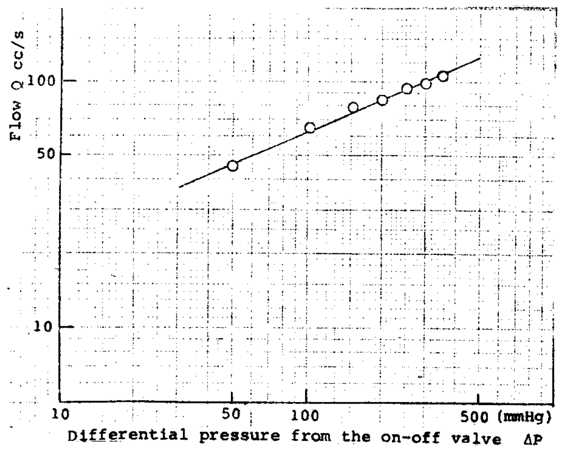

Fig. 8. Differential-pressure-flow charateristics of the on-off valve.

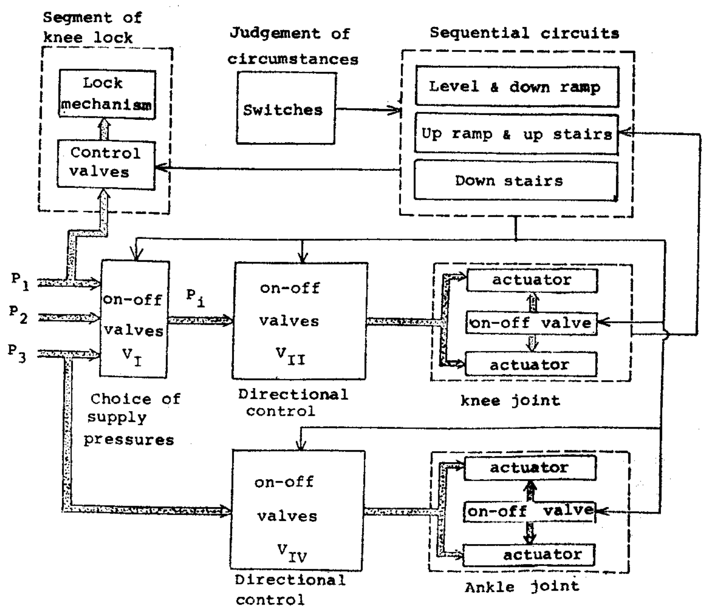

Fig. 9. Control block diagram of the above-knee prosthesis.

conclusion can be drawn until further experiments are conducted.

Compared with the conventional method of preventing knee flexion in walking by alignment adjustment, the compensation as realized in this prosthesis for knee flexion in the stance phase can be said to be more excellent and desirable in that it gives the sense of safety to the subject.

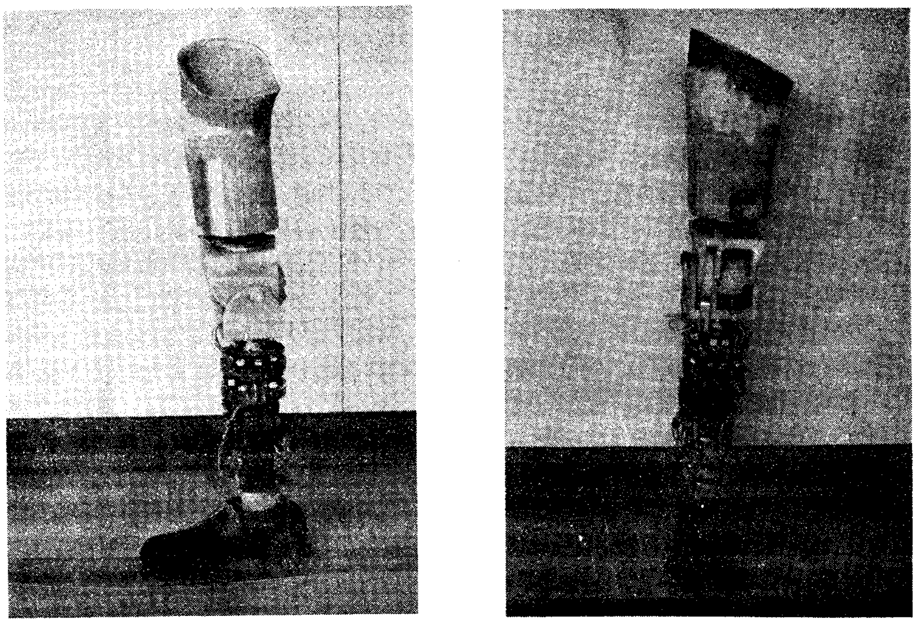

Fig. 10. Above-knee prosthesis.

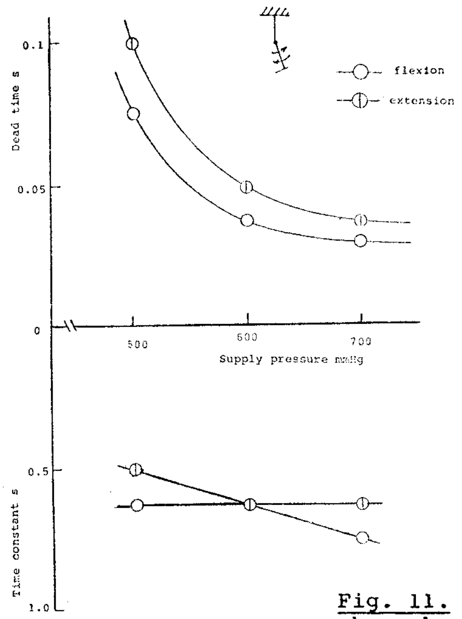

Fig. 11. Transient charateristics of the above-knee prosthesis.

5. Conclusion

It is essential that the joints, and in particular the knee joint, of an above-knee prosthesis to be used as part of the human body be provided with active function. Because the trial device at this time was intended chiefly to be used to point out the feasibility of the design, there is considerable room for improvements in terms of mechanism, but it was possible to show the usefulness of providing a prosthesis with active functions.

In the meantime, powered control of the ankle joint remains as the next problem to be undertaken.

Acknowledgements

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