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DEVELOPMENT OF A CONTROL PROGRAM FOR THE ACTIVE ABOVE-KNEE PROSTHESIS

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Summary

This paper discusses the experience acquired during gait restoration in the process of developing a control program for an active above-knee prosthesis. The control system is composed of bionic sensors, cybernetic actuators, and microcontrollers with control based on the principles of artificial intelligence and artificial reflex arcs. Three programs, with different requirement complexity have been tested. Problems of development and practical implementation of the program are discussed, as well as demands brought about by individual features of the patient.

Introduction

The advances in electronics and cybernetics as well as the latest findings in neurophysiology paved new ways for rehabilitation of patients with dificient locomotion. Functional electrostimulation (FES), active orthoses and prostheses, use of artificial intelligence and artificial reflex arcs in the control of assistive devices demonstrate wide possibilities being offered.

Our longterm objective is to develop, by combining selfadjustable modular orthoses (SFMO), FES, and cybernetic actuators, such systems that will provide a new quality in the rehabilitation of the disabled by maximal use of patient´s remaining ability.

Each of these systems has a control system composed of following elements:

- A set of bionic sensors that provide data on the foot-ground contact, angular values of joints and data on the vertical.
- Cybernetic actuators with four discrete states (free-locked, flexion-extension).
- A microcontroller based on microprocessor technology with the control program storred in the memory.

- Mechanical orthotic or prosthetic elements.

The aim of this paper is to point out certain aspects of control program development for the above-knee active prosthesis in the scope of gait restoration.

Methods and the Material

Possibilities of employing cybernetic actuators for deficient locomotion functions and especially in the development program for hybrid systems were studied on a standard skeletal prosthesis with a cybernetic actuator at the knee joint (1,2,3,4). A patient was selected: male aged 60 years, of medium height and build whose right leg was amputated above the knee in 1977 and who has been permanently using classic skeletal prostheses since 1979.

The control diagram for the active knee prosthesis is given in Fig. 1.

Program I

This is a basic program developed experimentally on a laboratory model, which provides locomotion with full active flexion and extension (limited by terminal angles). The foremost aim of this program was to examine the applicability of the laboratory model in practical work with the patient.

Program II

After mastering the previous program and eliminating the discovered errors we began to develop and to work out this program, which differes from the previous one mainly in that the knee extension is active only in the initial stage in order to make better use of the acceleration and deceleration forces while walking.

Program III

Intending to achieve a gait that will be as close as possible to the normal, we increased our demands. Features of the second program have been supplemented in the stance phase (after initial ground contact) by a short passive flexion with an additional active extension, in order to achieve amortization of the heel-ground contact.
Periodic analyzing has been conducted via TV recording and egzoskeletal goniometry.

D i s c u s s i o n

Program I is adequately suited to ts purpose. Aside from verifying the applicability of the laboratory model in practice, certain faults have been detected in both the hardware and the software, which was of great help in our future work.

Basic shortcomings were the following

a) A very limited possibility of system control by the patient i.e., the patient was unable to use actively the stump during the swing phase in order to control gait speed and the length of steps. The possibility of using acceleration and deceleration force was minimal.

b) A delay of flexion often occurred caused by inadequate foot sensors. It turned out that the contact should be broken off earlier i.e., when greater forces are at work, so that the flexion should start before too long step forward is made by the sound leg. One of the ways to perform this was by building - in protrusions inbetween the sensors and the ground in order to increase sensitivity and to influence sensors´ transmission treshold. An additional problem related to the flexion occurred when attempts were made to change gait rhythm because the flexion was limited only by the angle and not by time, which induced constant duration of individual gait phases.

c) The intensity of the voltage supply (power) needed for an adequate flexion speed was too high for the extension. Thus, too fast and too strong an extension was produced that lifted up the extended prosthesis, thus causing a delay in the heel-ground contact.

This pointed out insufficient adaptive control that,among other consequences, resulted in a dynamically inadequate model as the basic shortcoming of the system.

d) The other group of problems was related to large energy consumption and the unsatisfactory quality of the built-in mechanical elements. Prosthesis oscillations occurred, caused by collision of elements, as well as overheating and breaking-down of the motor during prolonged standing periods (caused by continuous break application).

Once again a difference in requirements was detacted in the case when the prosthesis is used as a model in program demonstration and in the case when it is tried on a patient. From the viewpoint of the patient this program was not satisfactory because of the imposed rhythm and length of the steps (inadequate dynamics).

The shortcomigs observed during the testing of the basic program have mainly been eliminated or alleviated so that the Program II produced a far better result and a gait much closer to normal, which provided for the use of stump´s muscle strength in determining the length and the speed of steps. A favorable effect of acceleretion and deceleration forces on the gait regularity was observed. This program was modified mainly in adjusting the terminal of angles flexion and initial extension.

Program III.

Introduction of the heel-ground contact amortization into the program was important concerning the intention to produce a gait close to normal not only visually but also to provide a qualitatively new sensation for the patient as well as protection of the stump and the spine.
The initial knee flexion at the moment of the heel-ground contact and an additional extension have been introduced into the program, relying on values obtained from gait analysis (7). Firstly, we tried to build-in a variable knee rigidity that was performed by the interchange of the locked and the free phase and their relationship in time. This mode of operation of the break was found to be unsatisfactory, both technically and functionally, because realization of even the slightest rigidity caused the knee to lock. This is why we tried to make use of the mechanical damping caused by friction between the components, which proved to be a successful solution. Yet another problem was the range of the initial flexion. Physiological ranges could not have been used because the patient is not fully in control of the initial flexion, which induces an enlarged insecurity factor (fear of falling). Therefore, the initial flexion was limited to 7^{o}, although not very precisely ($\pm 2^{o}$) due to the technical features of the system. There is another aspect to this problem i.e., that patient´s feeling of shock absorbens was lost during a long period of using classic prosthesis that are not endowed with this feature. Since we could not have used planned values for the initial flexion we `est program parameters back to minimal values to be gradually increased during the training. Also, owing to above probelms, a time limit was introduced for the initial flexion.

Concerning the additional extension, there were also some problems. Inadequate sensor sensitivity was in question, as before. In this case the ground was detected too late i.e., when the whole body weight was shifted onto the prosthesis and the center of gravity was already in front of the ankle joint; when the flexion should have already begun. Using existent sensors, the problem could be solved to some extent by a territorial and sensory separation of the middle of the foot sensors, because in different locomotion phases, the program seeks information on the existence (or lack) of ground contact for different force (transmission threshold) values. The information supplied by on-off sensors was found insufficient. The only appropriate way to solve this problem was by building-in analog contact sensors that would provide information on the actual weight (force) distribution throughout the foot area. Introduction of the initial flexion with an additional extension into the program increased considerablly the requirements for the mechanical performance of the system and for the quality of the components. The reliability of switch operation is one of the most delicate sections of the system. Eventual hazardous situations (meaning also irregualr switch operation) demand a more adequate treatment. A control program based on ideas and methods of artificial intelligence could, in our opinion, meet these requirements.

Patient using A-K prosthesis

Two main aspects of the work with the patient can be pointed out. Active involvement of the patient in program testing and modifying is irreplacable because it provides a versatility of feedback information nuances that should accompany technical measurements and gait analysis. On the other hand, we realized that we should abide by basic principles of gait restoration when mastering the program. Also, here holds the rule of progressing "from simple to more complex" since this enables the patient to master gradually, one by one, each new program element after the previous program has been fully mastered (and the patient has acquired the necessary confidence). We have to consider also the gait pattern which the patient has adopted using classic prostheses and to gradually create a new one, suited to the need of the new system and to the ability of the patient.

Conclusion

In designing a control program for the active above-knee prosthesis one should be guided by biomechanical and kinesiological needs and by system's technological possibilities.
In developing a program, we must progress from simpler to more complex issues (with the active cooperation of the patient) because of the following demands:

- The need to test practically individual elements of the experimentally developed program.
- Faults or erroneous conjectures can be corrected more easily in simpler programs.
- Easier work with the patient when restoring gait.

When practicing new elements, a deviation often has to be made from theoretic values because of individual features, habits and safety feeling of the patient.
The final program must result from the theoretical model as well as from the individual features and demands of the patient concerning prosthesis functionality.

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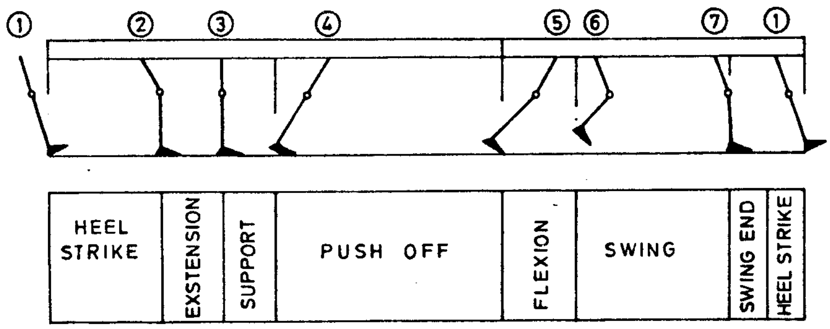

	Phase	Sensory input	Reflex	Sensory pattern	Joint state
1*	Initial flexion	heel strike	shock absorbens	H = 1	free
2*	Extension following Initial flex.	terminal angle of initial flexion	extension	α_t^p $\alpha \leq 180-\alpha_t^p$	extension
3	full support	knee extended ground contact	locked	$M(H + T) = 1$ $\alpha \geq 180-\Delta\alpha$	locked
4	Push off	heel off	flexion	$\bar{T} + H + M = 0$	flexion
5	Swing	toe off	flexion	$H + M + T = 0$	flexion
6		flexion terminal angle	extension	α_t^f	extension
7		extension terminal angle	locked	α_t^e	locked
1*	Initial flexion	heel strike	shock absorbens	H = 1	free

* only in program version III

α_t^f	terminal flexion angle	$\sim$ 120^o
α_t^e	terminal extension angle	$\sim$ 165^o
α_t^p	initial flexion terminal angle	$\sim$ 10^o
$\Delta\alpha$	tolerance angle for the extended knee	$\sim$ 7^o

Fig. 1