IMPROVEMENT OF HUMAN LOCOMOTION WITH POWERED LOWER LIMB PROSTHESIS

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Summary

In this paper the amputee-powered prosthesis system is examined. In the study the hierarchical organization for the control of the machine, the flow of information between the various levels, and the organization of the control of each level were investigated. Some structures for control and actuation in a monolateral amputee prosthesis are proposed.

Introduction

Complex problem concerning to design of multi-degrees-of-freedom mechanical structures the choice of energy sources of suitable power, size, and weight, and the selection of the control strategy arise in the development of lower-limb powered prosthesis.

The authors believe that at the current state of technology a solution at least of the problems of control, can be found, if the aim is limited to restoring only a certain number of the degrees of freedom of the human limb.

In this paper the underlying criteria and the general trends of the research on lower-limb powered prosthesis carried out at the C. S. S. C. C. A. of C. N. R., Instituto di Automatica-Universita di Roma are outlined. Some results obtained in the first phase of the work are also presented.

General Problems of Lower Limb Powered Prosthesis

Generally speaking, a lower-limb powered prosthesis is an unconventional locomotion machine, for it can move carrying a load by means of articulated supports. This point of view applies well to the case of bilateral hip-level prosthesis (i.e. Thalidomide children and patients with bilateral hip disarticulation); but a prosthesis is intended for rehabilitation purposes and in addition to engineering problems cosmesis and adaptability to the patients are of great relevance.

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The prosthetic system, composed of the terminal device and its control system is very complex and cannot be studied easely by the methods of classical mechanics. In fact even if one intends to restore only the active function of pelvic joints, knees, and ankles, the mechanical structure corresponding to the human must have seventeen degrees of freedom. In addition it is clear that there exists a high degree of interdependance between various movements.

Many authors have dealt with unconventional locomotion in the last decade, following different approaches. Tomović, Frank and McGhee /1, 2/ have considered walking machines as finite state automata; more recently Tomović, Vukobratović, and others /3/, Kato /4/, Ignatiev and Kulakov /5/, and our research group /6, 7/ have considered the problem of control through using different mathematical techniques, by dividing it into subproblems according to a certain hierarchy.

The control system is assumed to be split in the following levels:

- the decision level chooses gait, displacement velocity, and distribution of the machine. Extraordinary procedures for intervention in response to sudden variations of environmental conditions are provided. According to this choice suitable inputs are generated for the next level;
- the algorithmic level coordinates the movements of the limbs and generates the inputs to the servomechanisms that form the next level;
- the dynamic level is charged with the correct actuation of inputs generated by preceding algorithmic level.

In the system composed of the amputee and his prosthesis the decision level is committed to the amputee. A simplified representation of this level is given in Figure 1. The brain processes man´s will using information coming from the perceptive systems (sight, touch and equilibrium); the spinal cord and peripheral nervous systems allow transmission of this will to resudual muscles. Residual muscles are used for the transmission of man´s decisions to the prosthetic subsystem, because the use of EEG signals, as it has been done in a hand prosthesis /8/, seems not yet ready for the control of complex systems.

The problem of information exchange between man and prosthesis is of great importance. In a hierarchical structure the flow of information between various levels should be as little as possible

standing position
standing up
sitting down
level walking
going upstairs
going downstairs
right turn
left turn

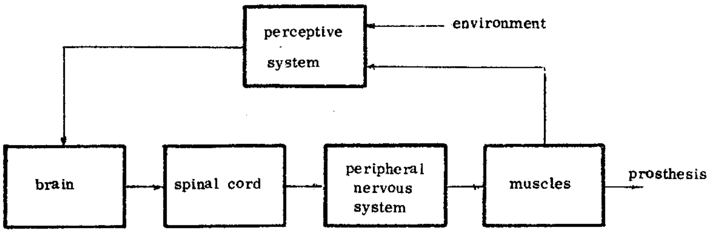


Fig. 1.

/9/. This is particularly true in this case, because the amputee is requested to control the prosthesis by muscles that can be different from those naturally used in locomotion. Table 1 shows the main types of motion that the amputee can choose voluntarily; it is inferred from this that he has to send to the control system at least an information vector with three binary components. As a physical support of this vector myoelectric signals or the forces exerted by swelling muscles against suitable sensors can be used. If the necessary number of control sources is not available, or if their activation requires unnatural movements to the amputee a manual selection should be used.As far as concerns the return of information from the prosthesis, the amputee should receive information on the state of the prosthesis, so that a form of proprioception can be restored. But it seems possible to the authors only to realize feedback channels measuring relative angles between the segments of the limbs and foot-ground contact, and sending this information to the control system of the prosthesis (algorithmic or dynamic level) without directly involving the patient´s body. This approach is consistent with considering the prosthesis as a walking machine that carries the amputee´s trunk, and satisfies conditions of minimum man-machine information exchange.

The algorithmic level, in case of not too severe amputations could be committed to the amputee if the effort of attention required of him is not too heavy. In a bilateral hip-level prosthesis the algorithmic level is very complex; in fact, not only is required the coordination of movements by also the equilibration function. A satiafactory approach to the study of algorithmic level is deemed to be that of considering the kinematic variables as the outputs of the control system: velocity or position rather than torques and forces. In this approach the controller´s outputs at the algorithmic level are waveforms of velocity or position at each joint of the multi-link mechanical structure (terminal device). These waveforms should conform the human normal gait. Many studies on biomechanics of human gait have been carried out, especially for walking on level ground. Some of these studies /10/ have pointed out that normal human gait satisfies conditions of minimum energy demand. In fact, in a complex system the algorithmic level will clearly perform some form of optimization of the motion. It is deemed that minimization of energy engaged in locomotion is a fundamental criterion which should be respected. If

$$I = \int_T \sum_{i=1}^{N} C_i(\tau)\dot{\theta}_i(\tau)d\tau$$

N=number of degrees of freedom of mechanical structure

T=pace period

$C_i(\tau)$=torque reffered to the i-th degree of freedom

$\dot{\theta}_i(\tau)$=angular velocity reffered to the i-th degree of freedom,

the optimal condition of movement is given by:

$$\min I = \{\dot{\theta}_i, t_{ki}\} \qquad i=1,2,\dots,N; \qquad k=i,\dots,n$$

n=number of pace phases in the considered gait

$$\dot{\theta}_i = \dot{\theta}_i(\underline{v}, l_1, l_2, \dots, l_h, \underline{m}, t_1, t_2, \dots, t_u, v)$$

$\underline{v}$=vector of information transmitted by the amputee

$\underline{m}$=vector of information concerning the state of the prosthesis

$l_1, l_2, \dots, l_h$ the lenghts of the segments of the multi-link structure

v=pace velocity

It could be interesting also to consider another index I', related to the total energy absorbed by the power source. If w_i^+ and w_i^- are the absolute values of the power supply and absorbed by the i-th actuator from the system, the toatl energy supplied by the actuator in T is:

$$E_{si} = \frac{1}{\eta_1} E^+ - \eta_2 E^-$$

where:

$$E^+ = \int_0^T w^+(t)dt \qquad \eta_1\text{: efficiency corresponding to supplied energy}$$

$$E^- = \int_0^T w^-(t)dt \qquad \eta_2\text{: efficiency corresponding to absorbed energy}$$

the index I' is so defined:

$$I' = \sum_{i=1}^{N} E_{si}$$

The authors think it is not convenient to insert this processing in line; it seems better to perform preliminary calculations in order to achieve a reasonable compromise between the aesthetics of gait and its efficiency.

The problem of equilibrium could be treated profitably by the sensitivity matrix. This approach has beed adopted by the group of Belgrade /11/. In our research group the problem of equilibrium is being faced from the point of view of a system of inverted pendulums by minimum time and energy criteria.

The dynamic level is formed by servosystems. They receive as input waveforms with determined temporal relations. This level has to compensate for disturbances caused by the amputee´s weight and the variable surface of the ground. It would be convenient to establish a certain degree of interaction between the servomechanisms so that a disturbance acting upon one of them would be transmitted to the others, suitably weighted as to guarantee a correct global performance.

What has been said can be regarded as a general approach to the problem of the prosthesis as a walking machine.

The bilateral hip-level prosthesis is a very complex system; for this reason as a first stage for the development of the project the realization of a one-degree-of-freedom prosthesis for the monolateral AK amputee has been chosen. For this prosthesis the algorithmic level is much simplified, and the dynamic level consists of only one servomechanism. The minimization of energy used is achieved by the maximization of the efficiency of the actuator. An amputee in good condition could be charged by the algorithmic level; for this purpose, it is necessary to determine if the information supplied by the amputee to the prosthesis in a natural way is sufficient to control it continuously.

Experimental Work

A series of myoelectric potential measurements on monolateral lower-limb amputees at various levels has been carried out in order to obtain information on the following points:

a) existance of myoelectric potentials that can be used after suitable processing as control signals;

b) whether the patient can calibrate muscular contraction so that a continuously variable control signal can be obtained;

c) whether or not there is interaction in the contraction of different muscles.

The measurement installation consisted of two independent channels for the extraction, amplification, and processing of the myoelectric potentials. Each channel was connected to a servo recorder or to an oscilloscope in order to supply visual feedback to the amputee. The block diagram of one measurement channel is shown in Figure 2.

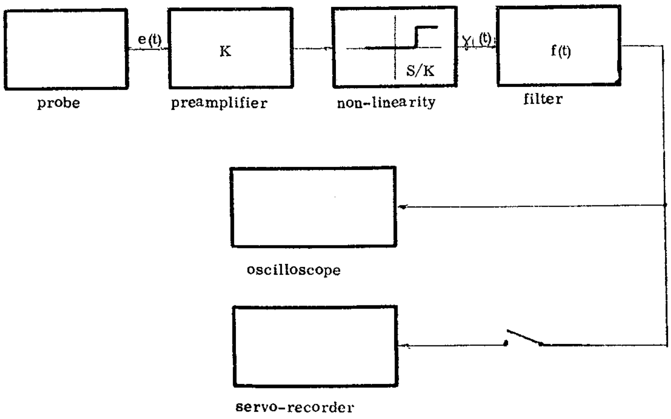


Fig. 2.

The myoelectric potentials were extracted by surface electrodes; the use of conducting pastes or skin treatment had been discraded. The potentials were processed by obtaining from them a variable according to the equation:

$$p(t) = k \int_0^T f(t-\tau)g(\tau)d\tau$$

where:

$$f(t) = \frac{1}{2} + \frac{1}{2}\text{sign}\left[e(t)-s\right]$$

with: $e(t)$ = instantaneous value of the myoelectric voltage
s = threshold amplitude
$f(t)$ = weighting function
T = averaging interval.

Concerning the choice of the points of extraction, the musscles that are used for normal walking and for walking with conventional prostheses were selected. In addition signals were extracted from those muscles that had not lost their anchorage to the skeleton through amputation. Accordingly, the following muscle groups were chosen:

- extensors and abductors of the hip (gluteus)
- flexors and, partly, adductors (ileo-psoas).

A great variety in the response among the various subjects was noticed. This is because at least two factors affected the response: subject´s muscular power before amputation and the quantity of connective subcutaneous tissue, because hypokinesis due to amputation, lying between muscle and probe. The signals can be improved by training the amputee and by improving the system of extraction.

In conclusion, the following points can be made:

a) in general, it is possible to extract myoelectric potentials which can be processed to obtain a signal correlated with the amputee´s will;

b) patients in good condition are able to calibrate their muscular contraction so as to keep the processed signal at a given level, i.e. one of four different values;

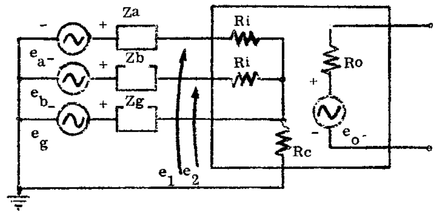
c) concerning interference in the contraction of separate muscles, it was seen experimentally that abductors always take part in the contraction of adductors whereas the reverse is not true. This effect seems to be due to the greater power and hence ability to produce myoelectric signals, of the abductors. This situation could be improved greately by training.

During the course of the measurements, it emerged that, due to movement of the probe on the skin, uncontrolled variations of the signal related to amputee´s will could be produced. These disturbances cannot be eliminated acting only on the circuit of extraction and processing, as can be seen by the model of the first stage of the circuit shown in Figure 3. In fact, the voltage that is actually amplified is given by:

$$e_0 = k\,\frac{R_i\left((Z_a+Z_b+R_i)(R_c+Z_g)+2R_cZ_g\right)(e_a-e_b)}{(Z_a+Z_b)(R_cZ_g+R_iR_c+R_iZ_g)+(Z_aZ_b+R_i^2)(R_c+Z_g)+2R_iR_cZ_g}$$

$$-\,k\,\frac{R_i(R_c+Z_g)(e_aZ_a-e_bZ_b)+R_cR_i(Z_b-Z_a)e_g}{(Z_a+Z_b)(R_cZ_g+R_iR_c+R_iZ_g)+(Z_aZ_b+R_i^2)(R_c+Z_g)+2R_iR_cZ_g}$$

This interfering effect could be avoided by using probes whose two electrodes can be applied separately to the skin, so that movements of the latter do not cause variations in contact conditions.



$$e_o = k(e_1 - e_2)$$

Fig. 3.

Characteristics of the Prosthesis Realized

The prosthesis is represented in the block diagram of Figure 4.

The terminal device is fitted with an actuator that provides only one degree of freedom for the movement of the knee. The prosthetic foot is jointed elastically to the leg so that dorsal and plantar flexion, pro- and supination, and intra-extra rotation of the foot can take place allowing a satisfactory gait. For this reason a cable type joint is used at the ankle.

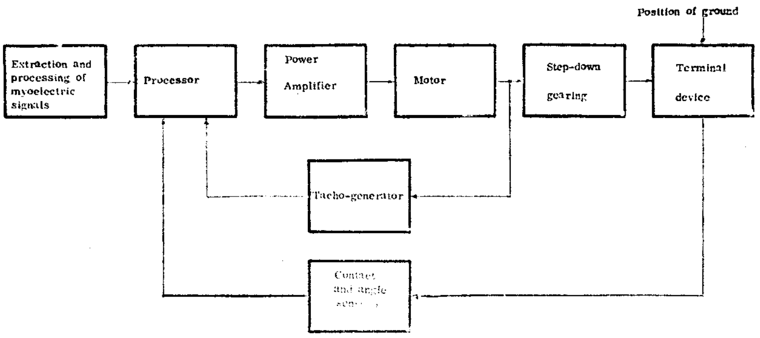
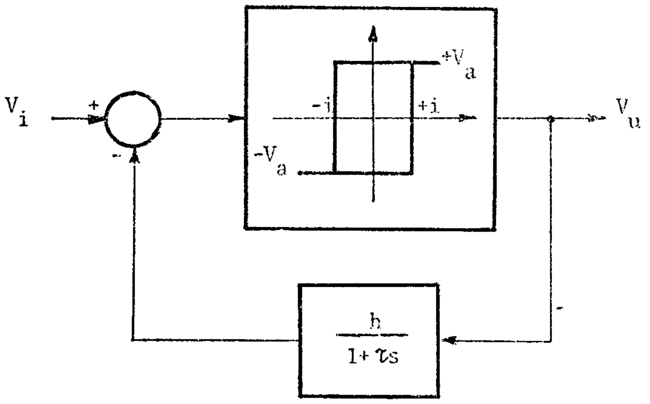


Fig. 4.

The actuator is permanent magnet motor type 2315-P26 of Indiana Genearl Corporation followed by a step-down gearing with step-down ratio of 225/1.

These devices are inserted in a velocity servomechanism: the feedback element is a tacho-generator whose transduction constant is 7 V/1000 rpm.

The power amplifier is a time-division pulse modulation system with variable pulse repetition frequency. The average value of the output voltage applied by the power stage across the motor windings is proportional to the input voltage. The block diagram of the amplifier is shown in Figure 5. The pulse repetition frequency range depends on the value of the parameters indicated in the figure and is between 400 and 2000 Hz. This range has been chosen as a compromise to decrease overall power dissipation due to both amplifier and motor.



$V_a = 60$ V
$i = 0.35$ V
$h = 0.1$
$\tau = 0.002$ sec

Fig. 5.

A settling time (5%) equal to half of the shortest pace phase and an overshoot less than 10% were assumed as specifications for the servomechanism design. According to this the correction network has been found by the usual control system design methods.

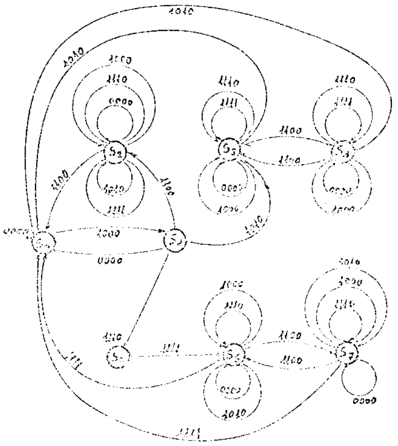
In relation to amputee's conditions /12/ two ways for the generation of input signals to the servomechanism (control signals) can be adopted. In the former case the flow of efferent information is continuous, and thus the control signal is continuously correlated with the amputee's will. In the second case the flow of efferent information is discrete and the control signal is therefore generated by selection and coordination organs activated by discontinuous voluntary signals.

In the case discontinuous flow of information, the selection organ enables the coordination organ which follows it to generate control signals appropriate to each of the following motions: walking on level at a given speed, and going up and downstairs and ramps, standing up and sitting down, and maintenance of standing position.

Under the assumption that the amputee can activate two muscles, one of which is at least independent of the other, and can calibrate the duration of contraction, the selection organ is an asynchronous machine whose outputs coincide with its stable states. Its state diagram and the truth table of inputs are shown in Figure 6. If the assumption is not verified a manual selection can be used.

I muscle activity	II muscle activity	Period of I muscle activity	Period of II muscle activity
0	0	0	0
1	0	0	0
1	0	1	0
1	1	0	0
1	1	1	0
1	1	1	1

T_o : fixed period



S_o	standing position
S_2	level walking
S_3	going upstairs
S_4	going downstairs
S_6	standing up
S_7	sitting down

Fig. 6.

The coordination organ generates the control signals for the various pace phases of the listed motions. Its structure was chosen under the assumption that the amputee could activate only one muscle for a very short time. There is a coordination organ for each type of motion. One of them can be described by the block diagram of Figure 7. Different level generators are used for the various motions; the same applies to the functions realized by the logic network and to the number of maximum and minimum angle sensors and the contact sensors. The contact sensors are microswitches that are installed on the sole of the foot and in the socket. Examples of velocity waveforms for various motions are shown in Figure 8. A detailed description of this system is given in /13/.

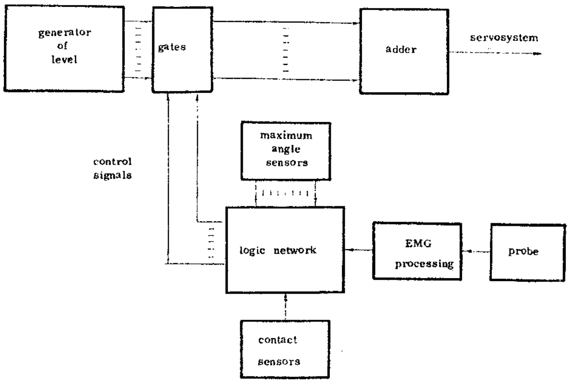


Fig. 7.

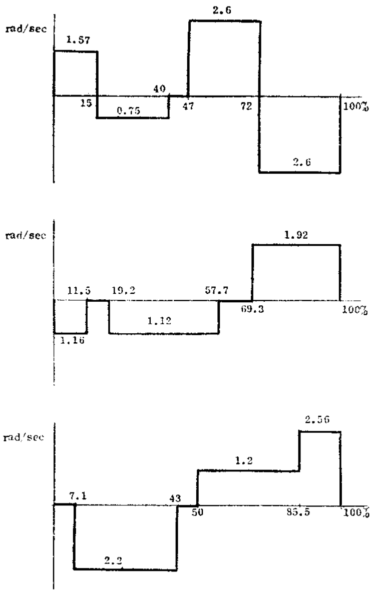


Fig. 8.

The system that extracts and processes myoelectric potentials is shown in Figure 9. It is made up by a pair of surface electrodes mounted on probes realized in our research group /14/ and by an amplifier with differential input, 66 dB gain over the band 150 - 2000 Hz, a common made rejection ratio better than

90 dB, and internal noise less than 10 μV r.m.s. for source resistance equal to 100 kohm. The non-linearity is a half-wave rectifier with saturation; the filter is a third-order approximate averaging type with critical frequency of about 6 Hz.

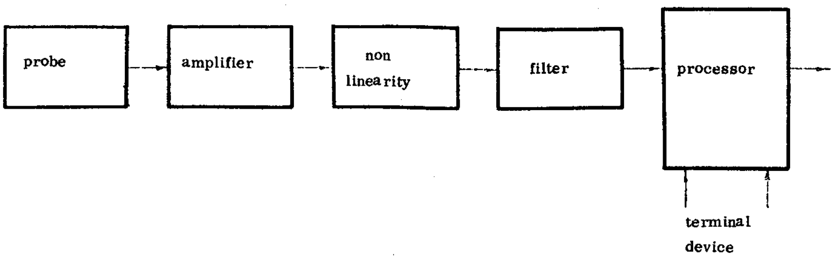


Fig. 9.

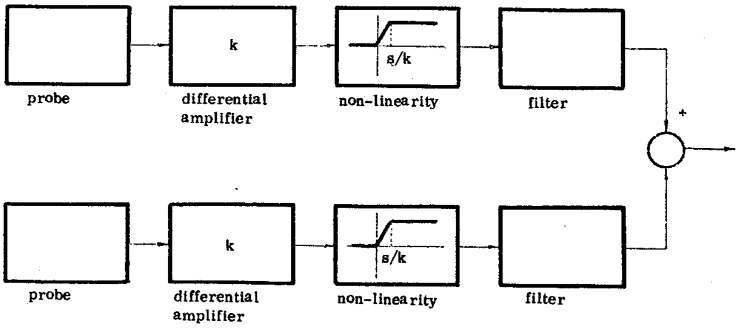


Fig. 10.

In the case of continuous flow of efferent information the system of generation of the control signals is shown in Figure 10. The control signals are obtained by the difference of signals obtained from extraction and processing chains, and are equal to those described before. Myoelectric potentials can be extracted from antagonist muscles.

Conclusions

The research program will proceed with the development of a two-degree-of-freedom prosthesis for monolateral hip-level amputees as a step for the investigation of prostheses for bilateral amputees.

A model of kinematic and dynamic aspects of human locomotion is thought necessary. Practicularly the problem of optimization with respect to a quality index related to energy expenditure will be studied.

The investigation of systems that can at least partly take the place of proprioception and the means of conveying to the prosthesis information connected with the amputee´s will should be deepened.

Fluidic actuators will be tried: this solution seems particularly suitable to hip-level monolateral and bilateral prostheses.

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