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LOGIC CONTROL OF PARTIAL ACTIVE ORTHOSES VIA REAL-TIME COMPUTING SYSTEM

S.R. Lazarević and S.M. Jauković

Abstract

The results presented in this paper reffer to the AMOLL project, whose principal objective is to restore the function of locomotion by an active orthoses. An approach to the synthesis of the semiautomatic - logic locomotion control system within a real time computer is proposed. The described control system provides the following locomotion actions: stationary gait, sitting-down and standing-up. The adopted technique of model building has enabled the modular organization of the control system. Any motion is composed by successive activation of modules in accordance with some predéfined sequence. The main feature of the proposed control system restsmon the fact that the adopted sequence can be easily expanded, reduced or rearranged in order to include some other activities, while the modules are not affected. In addition, the gait parameters are defined via the module parameters so that the control function is invariant in respect with the given system performance.

Introduction

The design of bioengineering control system which will be capable of duplicating the function of human extremities has been long under consideration /1-3/. However, due to the complexity of the biological control system and its ability in carrying out rather complicated operations, no machine yet deviced can fully supplement the functioning of the biological systems. Moreover, There seem that such a device cannot be expected in recent future. On the other hand, growing needs for the bioengineering systems for industrial and commercial use as well as for assistance to physically handicapped persons call for continuous improvement in already existing machines and for the construction the new u ones. The increasing interest in the design of bioengineering control systems has naturally resulted in variety of different approaches to the solution of the afore mentioned problems. Among them, the semiautomatic - logic approach seems to be rather promissing since it results in the controller which is remarkably simple despite the complexity of the process which is controlled.

The interest in the semiautomatic approach has been stimulated by desire to enable the patient to achieve full interaction with the orthotic device during the locomotion in the sense of system control as well as of internal - patient energy supply. In addition, compared with analog control, logic approach due to its nonumerical nature, offers significant reduction in time required for control signals generation, which is of great importa-

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nce for the real-time control of locomotion. The logic control should also be viewed through the fact that the natural biological system generates its control in accordance with the quite similar principles, therefore, if an orthotic device should duplicate human action, than it is likely to expect that the control has to be realized on a logic, rather than numerical basis.

Since the semiautomatic control assumes the active patient's colaboration it is obvious that the main problem arises in the qualitative description of man - machine interaction. Some efforts have already been made towards the technique of locomotion model building /4/, where it is explained how any locomotion action which assumes both man and machine activities can be represented by a discrete simulation model. However, there is still a question how those particular models, describing certain locomotion actions should be assembled so as to represent the general locomotion control system.

This paper attempts to show how the locomotion control system could be designed and realized within a real-time computing system. The described control system incorporates the following locomotion actions: stationary gait, sitting-down, and standing-up. The assumed technique of model building had suggested the modular organization of the control system. That is, any motion can be achieved by only two actions - modules, and successive activation of these modules in accordance with some predefined sequence can fulfil any particular locomotion task. The main advantages of the control system lies in the fact that the adopted sequence can be easily expanded, reduced or rearranged in order to include some other actions (eg. staircase gait), while the modules are not affected. In addition, the gait parameters are represented as the module parameters,,so they can be changed in the straight forward manner.

It should be pointed out that the presented control system has been primarily dedicated to the AMOLL Project /5/. However, the described concept, due to its modular nature, can also be applied to any orthotic device which has the possibility for man - machine interaction during the gait.

Basic Control Principles

It is well known that the design of the control algorithms strongly depends on the model of the process which is going to be controlled. However, neither the state vector nor the model itself is, in general, uniquely determined. In addition, semiautomatic, logic control of locomotion calls for system description by finite set of continuous and discrete variables, rather than in the form of differential or difference equations, which implies that this system cannot be treated according with the classical principles of system theory. The complexity of the problem is further increased by the requirement that the human factor has to be incorporated in the model. Taking all this into account it is obvious that the state vector should be adopted on the ground of visual analysis and experimental measurements with human being. The on-line measurements of the human gait /6/ has indicated that in any locomotion action it sufficient to keep track on the changes in joint angles as well as the contacts with the ground. Therefore, the state vector is adopted as shown in Figure 1. It should be noticed that the concept of semiautomatic control implies the use of the crutches as the only means by which the patient can transffer his internal energy to the system. In order to

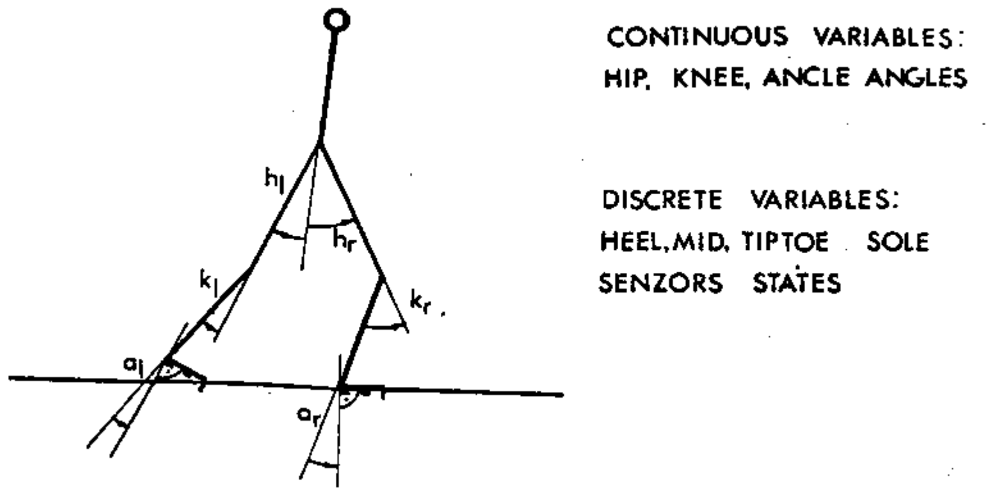


Fig. 1.

allow maximum freedom in patient's activity, the positions of the crutches were not included in the state vector.

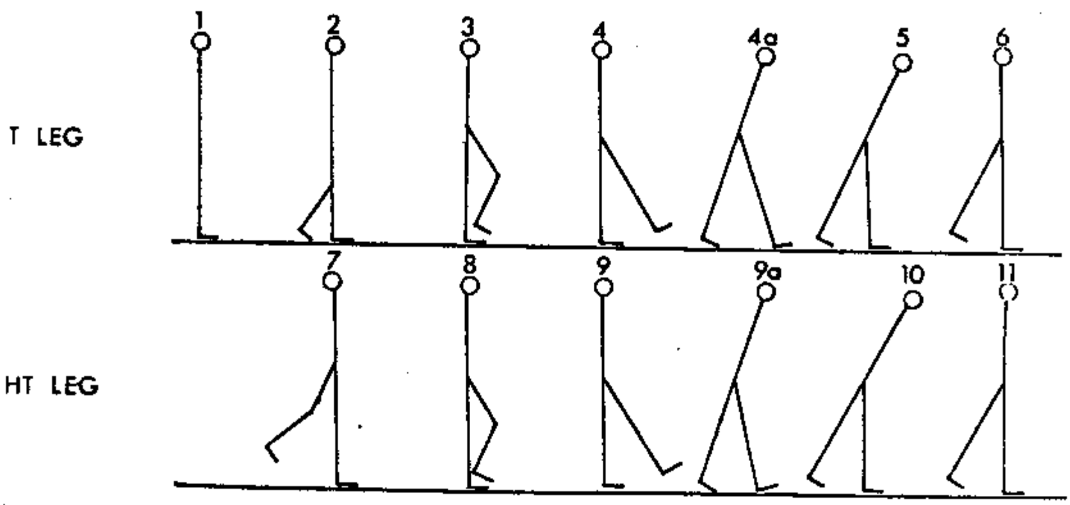
Simulation of the locomotion process requires simultaneous realization of two basic tasks: control signals generation and maintaining patient stability. Since the adopted state vector consists of two different types of variables - continuous and discrete - where the latter one apriori cannot be controlled it seems reasonable to divide state vector in functional sense as well. Consequently, joint angles should be considered as controlled varibles, while the contacts with the ground play the role in checking of patient's stability. Therefore, the synthesis of the control algorithm should be done so to provide the appropriate changes of joint angles. Furthermore, it should be adaptive to the feedback informations from the sensors in order to detect and deal with the hazardous situations.

Once the state vector is divided in functional sense there is still a question which variables should be directly controlled, that is, if the logic control is considered, which variables should represent the output of cybernetic actuators /2/. Thorough analysis of experimental results has indicated that only four actuators located at the hip and knee joints might be sufficient to accomplish the particular action. In that case, the ancle joints remain passive, so that they can rotate only as a result of the internal patient's energy supplied by the means of crutches.

Synthesis of the control of any locomotion activity preassumes the existance of the process model based on the defined state vector. The analysis of the possible functional relationship of the state variables had indicated the implementation of the discrete model which represents the locomotion process with a finite set of time successive system states - events. (In analogy with digital discrete simulation where this modeling technique is vastly applied, this model has been reffered as discrete simulation model of the locomotion process /4/.) Since, in practice, every locomotion process is realized by simultaneous rotation of joints, it is obvious that discretizing procedure in the sense of events extractic certain degree of arbitrariness. Visual analysis and experimental measuring of the process will reduced the total event space to the subspace consisting of only those sequences which can successfully represent the particular activity. Once the model is adopted, control algorithm has the single task to provide the transition of the system from one event to the other. This means that prior to model building the basic control philosophy has to be precisely defined. It will act as the main criterion in the process of sequence of events extraction from the allowable subset.

It is evident that the simulation model strongly depends on the particular locomotion activity but all models have to meet certain requirements. The state varible values associated with one event are represented as event atributes and consequently can be changed during the locomotion. Besides, the adopted sequence of events permits the transition from one state to another either by cybernetic actuators, or by patient, or by coordinated mutual activities. Finally, due to the concept of man - machine interaction, the models are characterized by event - oriented timing system. Accordingly, the time is incremented from event to event, while the duration of the transition will entirely depend on the patient.

In oredr to illustrate the above mentioned model properties ome of the possible stationary gait model is presented (Fig.2). The basic philosophy reflects the fact that the patient should b· able to take the part in the gait process with the minimum effor



EVENT			TRANSITION
1	$h = k = 0$	HEEL·MID = 1	$h = 0$; $k \in [0, k_1]$
2	$h = 0; k = k_1$	HEEL+MID = 0	$h \in [0, h_1]$; $k = k_1$
3	$h = h_1; k = k_1$	HEEL+MID = 0	$h = h_1$; $k \in [k_1, 0]$
4	$h = h_1; k = 0$	HEEL+MID = 0	PATIENT'S ROTATION UNTIL THE CONTACT WITH THE GROUND
5	$h = h_1; k = 0$	HEEL · MID = 1	$h \in [h_1, 0]$; $k = 0$
6	$h = 0; k = 0$	HEEL · MID = 1	

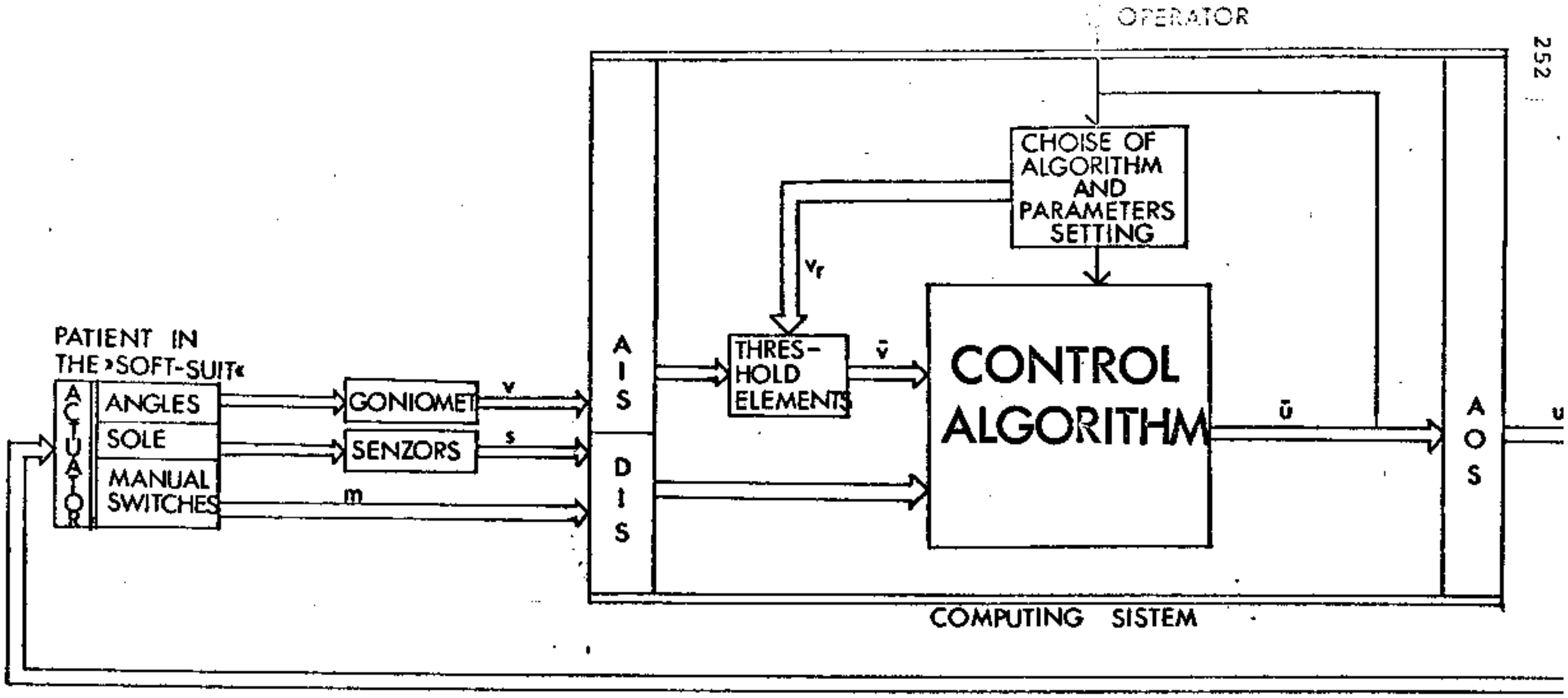
Fig. 2.

It should be emphasized that the described model provides the complete invariance of the control algorithm in respect with the change of system performance (step lenth, cadence etc.). Namely, the change of the performance will only affect the values of the event atributes, while the way in which the system transits from one state to another, remains unchanged.

The nature of the models enables the control algorithm to be easyly synthesised. The algoritm has the measured values of the state variables as input informations and processes them in accordance with the model im order to generate the cybernetic actuators signals as its output. If the control algorithm is to be realized within a real-time computing system, than the overall system organisation can be presented as in Fig. 3.

To this end, it is neccessary to consider the stability of the patient during the locomotion. Due to the fact that the simulation model splits the locomotion process into several phases, stability can be treated from phase to phase, rather than for total process. By the analysis of every event and corresponding transition it is possible to select the situations which are characteristical from the stability point of view. Hazardous situations can be divided in two groups: situations which are recognized by the patient himself either visually or hand senzor feedback; and situations which are detected on the base of the informations received from the sole. The first group has no influence on the control algorithm, since it is expected that the patient, having recognized the instability, will reach the stable position applying the crutches in the most skilfull manner. The second group, including the situations which cannot be handled by the patient, must be incorporated in the control algorithm. It should be emphasized that the experimental and theoretical investigations /6/ have proved that every disturbance affects only the status of the sole. This hint could be easily understood knowing that applied control "forces" the active joints to achieve the predetermined positions, regardless the status of the system. Because of that, logic status of the sole is the matter of crucial importance in the concept of hazardous situations handling. Being detected the hazardous situation, the control algorithm has to react in such way to provide the successful transition, from given dangerous position to the most imminent safe position. Namely, it is neccessary to extend the model with the set of events in which the system should be transfer when the instability occures. Such pairs of events - "hasard" and "responce to hasard" - are not the members of the basic sequence of events. Some particular hazard event can be expected along the corresponding transition path, while the responce to the hazard is the alternative to the next "basic" event. In addition, for every pair, the model should also include the sequence of events which will transfer the system from the responce to hazard event to one of the basic events.

The analysis of the stationary gait model (Fig.2) indicates, for example, that "touching of the ground with the top of the shoe" should be treated as one of the possible hazardous situations. Moreover, it is evident that this disturbance can occure in transition from the second to third event (left leg), or from seventh to eighth event (right leg). Therefore, the hazard event is described by the signal (logical one) from the appropriate tiptoe senzor. The responce to hazard event is locking the active hip joint (left or right). Next events consist of realising the appropriate leg and actuator rotations until the basic event "one" is achieved.



v - JOINT ANGLES

v_r - REFERENT JOINT ANGLES

$\tilde{v}$ - JOINT ANGLES

s - STATUS OF THE SOLE

m - STATUS OF THE MANUAL SWITCHES (choice of locomotion action and parameters)

$\tilde{u}$ - CONTROL ACTION

$\bar{u}$ - $\in \{\tilde{c}, \bar{t}, \bar{i}\}$

$\tilde{c}$ - CONTINUATION OF ROTATION

$\bar{t}$ - CHANGE ROTATION TYPE

$\bar{i}$ - INTERRUPT SISTEM RUN

Fig 3.

Software support

The software support is organized as the programming system which will, when once initialized and guided into the "system ready state", monitor the computer run, controlling thus the active orthose functioning as well. Since the programming system should provide two basic tasks: initialization in the sense of parameter values settings and initial patient positioning, and the monitoring and controlling of the system throughout the gait in accordance with the given algorithms, it can be divided from the operational point of view in two operating modes - system initialization mode and system gait mode.

Prior to the transfer of the control to the programming system, several operations which will prepare the patient and the system for the gait process should be performed. Namely, it is neccessary to help the patient to achieve vertical position, when he is left to its own maintaining his stability by the means of crutches. Meanwhile, the programming system asks for settings of relevant data and parameters in accordance with the defined source language. When these preparatory operations are completed, the "system ready" message is produced, and computer control is transfered to the system gait mode, which is responsible for the all future actions. Precisely, the programming systems enters the "patient stability trap" where it is left until the stable position is recognized through the logic status of the sole. When the patient is found to be stable, the message "patient ready" indicates the exit from the patient stability trap. The programming system enters the "wait trap", which controls the system in the option phase, waiting for the patient to decide whether he wants to sit-down, stand-up, change the gait type or start to walk.

The analysis of the adopted models and emergencies has led to the conclusion that gait can be realized by four basic modules which execute actuator actions and monitor patient activities during the normal gait and disturbances. The execution and monitoring of only two activities is sufficient for normal gait. One of them concerns the motors activation, the other is reffered to the patient external rotation of the body, by the means of crutches. Both system activities, regardless to the actual transition which has to be performed requires essentially the same informations: possibility for the hazard occurrance; transition type (performed by the actuators or the patient); affected joints (which have to be activated); rotation type (backward, forward, locked); referent value or expression whose fulfilling will terminate rotation; next transition, or other system status, which has to be initiated when the transition is terminated. This feature permits every gait transition to be represented with one record which contains all neccessary informations. Furthermore, the defined discrte models enable the arrangement of these records in the"gait string"in which every record is labeled and can be reffered with the gait pointer value (Table 1.).

The two basic gait activities are performed by two program modules: "actuator module" which guides the actuators;"patient - module"which monitors the system during the patient's external manipulations. Both modules have only one input value - the pointer which indicates one record from the gait string. The output is also single-valued, represented by the new pointer value which marks the next transition. In order to provide the hazardous situations handling, these modules also include testing of hazard occurrance together with the transfer of the control to the emergency part of software.

It should be pointed out, that the requirement that the termination of any gait action should provide the patient with the possibility to choose some new action, has implied the extension of the gait string to include the wait trap as possible module output.

The similar analysis can be applied when the hazardous situations are considered. The detection of hazard as well as the proper system reaction requires the following information: detection; type of reaction (actuators or the patient); affected joints which has to be actuated, or affected leg which has to be unloaded; type of joints rotation; referent value or expression to terminate rotation; next transition, or other system status. These informations can again be grouped in one record, and the records can farther be arranged so as to form the "emergency string" (Table 2.).

When the possible types of hazard reactions are analysed, it is apparent that they can be realized by only two software modules, one of which will keep track of automatic control of the joints - "emergency actuator module", and the other which will provide the system monitoring during the patient's external activities - "emergency patient module". Input in the module is always through the emergency pointer which can achieve any value from the emergency string. But, the output pointer can reffer to only two possible system actions: it can transfer the system back to the record of the gait string which has called for the emergency reaction, or it can transfer the system into the patient stability trap, where it will remain until the patient succeed in achieving the safe standing position.

In accordance with the assumed gait and emergency string, the control of the program flow during the computer run is done by the means of the gait pointer which acts as instruction adress register. That is, when the programming system is initialized it enters the wait trap and wait for the gait pointer setting. The given pointer value will refer to the one particular entrance of the gait string, whose record will provide appropriate modules conection which will further enable the performance of the required transition. At the end of the transition, the pointer value is redefined to indicate the new string entrance. This procedure will go on until either the entered gait string record does not transfer the system into the wait trap, or the hazard causes the emergency string to be entered and to transfer the system into the patient stability trap.

The presented gait and emergency string are developped in accordance with the three adopted models for the stationary gait, sitting-down and standing-up /4,7/. Since the standing-up and sitting-down models are represented by one basic pair of events which is successively repeated until the final state is achieved, the illustration of the corresponding parts of the gait string (table 1.) is simplified by description of only these basic pairs and the final states.

Conclusion

The proposed concept of the control system organization has been verified through the experiments with the Vukobratović's active exoskeleton. These experiments /7/ have proved the invariance of the control algorithm in respect with the modification of system performance via the change of referent expression in the strings. However, due to the nature of the medium it was not

Table 1.

GAIT POINTER	EMERGENCY POINTER	MODULE	ACTUATED JOINTS	ROTATION TYPE	REFERENT EXPRESSION	NEW POINTER	REMARK
1	0	actuator (A)	left knee (LK)	backward (B)	k_1	2	GAIT
2	1	A	left hip (LH)	forward (F)	h_1	3	
3	0	A	LK	F	0	4	
4	2	patient (P)	none	with the crutches	sole L heel·Lmid = 1	5 or 11	
5	0	A	LH	B	0	6	
6	0	A	right knee (RK)	B	k_1	7	
7	3	A	right hip (RH)	F	h_1	8	
8	0	A	RK	F	0	9	
9	4	P	none	with the crutches	sole R heel·Rmid=1	10 or 12	
10	0	A	RH	B	0	1	
11	0	A	LH	B	0	wait trap	GAIT TERIMINATE
12	0	A	RH	B	0	wait trap	
13	5	A	LRK	B	k_2	14	STANDING-UP
14	0	A	LRH	B	k_2	15	
15	5	A	LRK	B	0	16	
16	0	A	LRH	B	0	wait trap	
17	6	A	LRH	F	h_2	18	SITTING-DOWN
18	0	A	LRK	F	h_2	19	
19	6	A	LRH	F	h_{final}	20	
20	0	A	LRK	F	h_{final}	wait trap	

Table 2.

EMERGENCY POINT	DE[illegible]ON	MODULE	ACTUATED JOINTS	ROTATION TYPE	REFERENT EXPRESSION	NEW POINTER	REMARK
1	Ltip	EA	LH	locked (L)	–	11	Touching the ground
11	–	EP	unload left leg (ULL)	–	Ltip+Lmid+ +Rheel=1	12	with the top of
12	–	EA	LH LK	B F	0	Patient Ready Trap	the left shoe
2	Lheel=0	EA	RH	F	Rheel=1	21	Falling down
21	–	EP	unload right leg (URL)	–	Rheel+Rmid+ +Rtip=1	22	in forward
22	–	EA	RH	B	0	23	direction
23	–	EP	ULL	–	Lheel+Lmid+ +Ltip=1	24	
24	–	EA	LH	B	0	PRT	(during the gait)
3	Rtip=1	EA	RH	L	–	31	Touching the ground
31	–	EP	URL	–	Rtip+Rmid+ +Rheel=0	32	with the top of
32	–	EA	RH RK	B F	0	PRT	the right shoe
4	Rheel=0	EA	LH	F	Lheel=1	41	Falling down
41	–	EP	ULL	–	Lheel+Lmid+ +Ltip=0	42	in forward
42	–	EA	LH	B	0	43	direction
43	–	EP	URL	–	Rheel+Rmid+ +Rtip=0	44	
44	–	EA	RH	B	0	PRT	(during the gait)
5	Ltip·Rtip=1	EA	LRK	L	–	51	Falling down in forw.dir.
51	–	EP	LRA	Increase	Ltip·Rtip=0	back to the gait module	(during the standing up)
6	Ltip·Rtip=1	EA	LRH	L	–	61	Falling down in forw.dir.
61	–	EP	LRA	Increase	Ltip·Rtip=0	back to the gait module	(during the sitting down)

possible to evaluate the models neither from the energy consumptions point of view, nor from the gait elegancy aspect. Namely, it may be argued that the presented models are not optimal in any sense, but the models cannot be verified without the experiments with an orthotic device which permits the semiautomatic control, i.e. the full man - machine colaboration. If the patient should have the expected degree of freedom on the decision level, than his colaboration with the machine should be only qualitatively, rather than quantitatively prescribed within a model. Since, the actual patient's contribution cannot be predicted, it is obvious that the optimality may be establish only through the experimentation with the patient.

The principal objective of this paper is to show how the semiautomatic, logic control system should be designed and realized within a real-time computing system, while the described models should be viewed as an illustration of the string construction. Besides, the models are oriented to enable the patient to successfully overcome the begining difficulties concerning the interaction with the device, so they present an acceptable starting point in the experimentation phase. Finally, it should be emphasised that it is the modular organization of the programming system which provides the full adaptivity of the proposed control system which is of special interest since the models and the parameters can be choosen so as to meet patient's requirements in the best possible way.

References

/1/ R.Tomović, R.Bellman: A Systems Approach to Muscle Control, Math. Biosc., Vol. 8, (1970), pp.265-277.

/2/ R.Tomović, R.B.McGhee: A Finite State Approach to the Synthesis of Bioengeeniring Control Systems, IEEE Trans., Vol.HFE-7, N.2, (June 1966), pp.65-69.

/3/ R.Tomović, P.Rabischong: Trends in Assistive Devices for Upper and Lower Extremities, Proc. V Symp. on Ext. Cont. of Human Extremities, Dubrovnik, (August 1975.)

/4/ S.R.Lazarević: General Design procedure of the Control Algorithm for the Semiautomatic Logic Control of Locomotion, Proc. XIX Conf. ETAN, Ohrid, (June 1975), (In Serbocroatian)

/5/ P.Rabischong, M.Pelegrin, R.Tomović, A.Clot et alt. : The AMOLL Project, Proc. V Symp. on Ext. Cont. of Human Extremities, Dubrovnik, (August 1975)

/6/ V.S.Cvetković: A Qualitative Amalysis of Locomotion Functions in respect to the Stationarity, Simetry and mean characteristic recognition of normal gait, M.S. Thesis, Faculty of Electrical Engineering, University of Belgrade, Belgrade, (Nov. 1974.), (In Serbocroation)

/7/ S.Japković, N.Jauković: An approach to the Logic Control of Locomotion, Proc. XIX Conf. ETAN, Ohrid, (June 1975.), (In Serbocroatian).

THE INFLUENCE OF THE WALKWAY SURFACE ON THE LOAD DISTRIBUTION IN MODULAR PROSTHESES

U. Boenick and R. Zeuke

Abstract

In order to reduce the dimensions of weight bearing parts, modular prostheses must be designed using the principles of lightweight construction. This requires that sufficient data concerning the acting forces and moments are available for the design engineer. However the force studies now in existance were carried out primarily under laboratory conditions, i. e. on smooth floors, and this to some extent does not meet the real conditions.

In this investigation, a study was made to determine the variability of prosthesis loading, when an amputee walks on outdoor grounds. For this purpose the shank of an AK-prostheses was instrumented with a strain gauge pylon which allowed for the measurement of the axial forces, the shank axial rotation moments, and the bending moments in the sagittal and frontal plane at two different levels.

The tests were carried out on asphalt flooring, cobble-stone, rubble and fine grain sand. Each run was conducted at velocities of 2 km/h, 3 km/h and 4.5 km/h. For signal transmission a multi-channel telemetry system was used. After demodulation in the receiver the results were plotted on a recorder and analysed by means of statistical methods.

The results indicate that the greatest axial forces were obtained on rubble. At a velocity of 4.5 km/h their magnitude was about 150 % of bodyweight, which is an increase of nearly 40 % when compared to the value which occurs on asphalt. The asphalt flooring used here is similar to the laboratory floor of previous investigations. The maximum torque was measured on sand. Its was about 40 % higher than that measured on asphalt. The greatest bending moment appeared in the sagittal plane when

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