ABOVE - KNEE PROSTHESIS WITH ATTITUDE CONTROL

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Abstract. This paper presents a control algorithm for the above-knee prosthesis implemented in the self-contained single board microcomputer. In designing the control system the non-numerical, logical, hierarchical approach has been adoppted. The attractiveness of such an approach is enchanced by the fact that the controller might be designed using asynhronous finite state automata with sensory feedback.

INTRODUCTION

The design of a bioengineering control system capable of duplicating the function of extremities has been for a long time under consideration. However, due to the complexity of the biological control system as well as to the fact that human gait is one of the most sophisticated system of locomotion, no machine deviced can fully supplement the functioning of the biological system. Therefore, in order to enable better understanding of the possible approaches in modelling and control of the locomotion systems a relatively simple problem has been investigated: control of the above-knee prosthesis attached to the human body assuming that all other human functions are preserved (eg. maintaining stability, performing the swing phase etc.). It is required that the control system for above-knee prosthesis should assist a patient in performing a certain number of locomotion activities: gait, sitting and getting-up.

When considering the knee-locking system the basic question is from what (or how) should the controlling be done. One approach is to use a knee-locking system controlled by electromyographic (EMG) potentials from remanent muscles in the amputees stump |1|. The second one promissing control concept is to imitiate the gait pattern of the normal leg |2|. Another possible solu-

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tion, which has actually been adopted in the course of investigation, is to apply the concept of the semiautomatic, hierarchical, logical control |3-6|.

After the first promissing results in designing a microcomputer based knee-locking logical control system |7|, the technologically and conceptually advanced version of the controller has been developped |8|, whose basic aspects are presented in this paper.

II. CONTROL SYSTEM ORGANIZATION

The theoretical research of locomotion systems has pointed out that it ought to be tackled as a large system even in the simplest cases, which has imposed a hierarchical decomposition. Each level of hierarchy has to be embodied within a subsystem supporting a part of the overall control system.

The highest level belongs to the patient who voluntarily decides to initiate a particular locomotion activity. The decision is issued to the external locomotion system which further process the information at the decision level, recognizes the required activity and selects the appropriate mathematical model. This completed, the control is transfered to the algorithmic level where the informations on the values of the state vector coordinates are processed in accordance with the given model. As the result the control signals are generated and forwarded to actuator at the lowest hierarchical level.

The actuator level is expected to realize a strictly defined set of commands and consequently it might be considered as fixed from the functional point of view. This makes it very suitable for hardware implementation. Inversely, the algorithmic level comprising a variety of models for locomotion activities, together with the adaptive parameters incorporated in those models, presupposes a fair amount of flexibility. The same is true for the decision level at which the changes at the algorithmic level are reflected. It leads to the conclusion that these two levels - decision and algorithmic level - should be embodied in a software structure. The necessity of miniaturization strongly recomands a software implementation within a microprocessor system.

III. ACTUATOR LEVEL

The above-knee prosthesis is supplied with only one actuator - the hydraulic cylinder for stiffness control. From the functional point of view the actuator has two states: locked (rigid) and deblocked (loose). In the locked state the actuator prevents knee joint from flexion, while it is totally free for extension. But, when deblocked, the actuator permits free joint rotation in both directions (loose state of the joint). Therefore the actuator can be regarded as a single input (voltage signal) - single output (status of the device) system. The algorithmic level should provide the locking system input with a binary voltage signal. In addition, on the same control level the proper timing for the input signals should also be determined.

In orther to adapt the actuator for the computer control, a link mechanism driven by electrical micromotor equipped with the reduction gear is added. This electromechanical system drives then the control valve of the hydraulic cylindar into two terminal position - minimal (zero) stiffness and maximal (full) stiffness - depending upon the input signal. Consequently, the appropriate input signal is of the binary nature (0, 1), which enables the direct coupling of the controller output to the input of the electrical motor through the two-state electronic driver module, as indicated in fig.1.

In designing the actuator and its adjoint electromechanical system the vast experimental work has been undertaken. This has enabled the identification of the crucial parts at the actuator level which are responsible for the reliability of the system performance. The experimental results has led to the certain modifications of the actuator itself, concerning its high sensitivity to the lower initial position, as well as to the recognition of the relevant parameters of the electrical motor and the reduction gear |8|.

Apart from that some investigation concerning the energy consumption has also been carried out. It has been shown that the time period of about 8 msec is sufficient for driver to deblock the prosthesis. After that the control signal can be removed, while the actuator remains in the loose state. On the other hand, the preservation of the maximum stiffness state appears to be critical not only for the reliable system operation, but also from the point of view of energy consumption. Namely, there is no more need to use power to maintain the link mechanism in the terminal position once it has been reached. However, if, from any reason, the terminal state is not preserved within the necessary period, the additional driving pulse should be issued from the controller. This has been provided by introducing the microswitch between the controller's stiffness output and the driver |8|.

IV. MODELING OF THE LOCOMOTION PROCESSES

The designing of the algorithmic and decision levels has been preceeded by a synthesis of mathematical models for the selected locomotion activities. According with the concept of non-numerical, logical control |4-8|, the

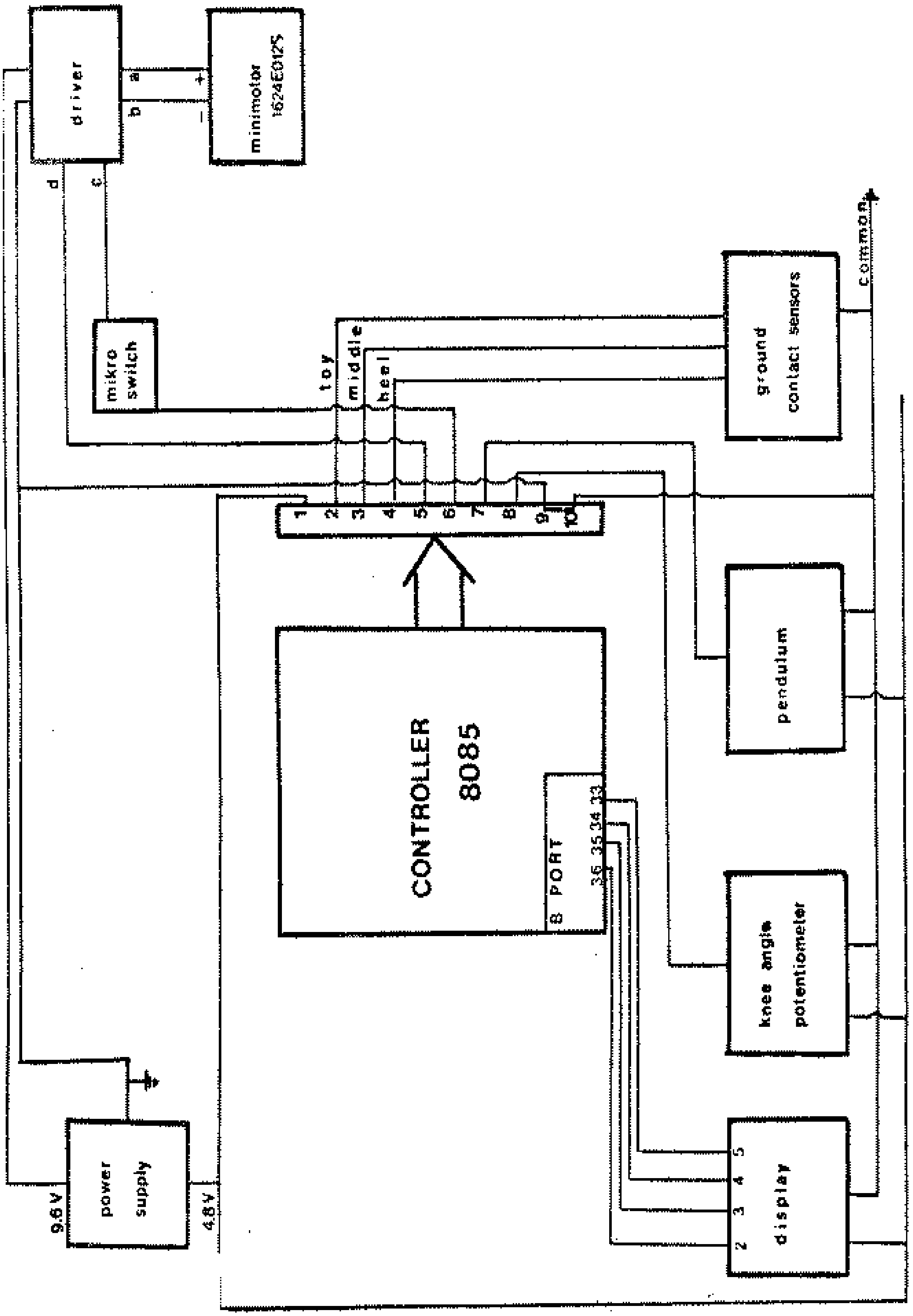
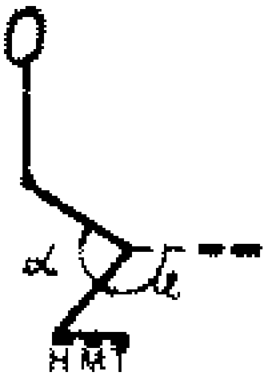


FIG.3. INPUT-OUTPUT WIRING DIAGRAM OF THE CONTROLLER

mathematical models represent the decomposition of the process into a sequence of successive events, each being described by discrete values of the state vector. The models are characterized by an event-oriented timing system, which means that the time is incremented from event to event, while the transition duration depends entirely on the patient's activity. Consequently, once the model has been adopted, the control system has a single task to maintain the system transition from one event to another. It is left to the patient to initialize a transition, decide about its duration and a proper timing.

The theoretical and experimental analysis of the locomotion activities has suggested that the state vector should be choosen as indicated in fig.2. For tracking the knee angle and the vertical position the linear potentiometer and the pendulum, respectively, are used. The corresponding continious voltage signals are introduced into the microprocessor via the 8-bit A/D converter (Fig.1.).



α - knee angle
φ - vertical position
(shank angle)
H - heel sensor
M - mid sensor
T - tiptoe sensor

FIG.2. STATE VARIABLES

The ground contact informations are of discrete, binary nature and they are obtained by the Rancho Los Amigos in-soles wired so to form three sensitive zones. This three sensor signals are directly introduced into the microprocessor (Fig.1.)

It should be pointed out that from the data acquisition point of view the measurement of the vertical position has proved to be the critical one. The output of the pendulum is highly sensitive to all non-vertical accelerations and shocks. Therefore, at the present stage, the mathematical models include the vertical position angle only in those events where the correct behaviour of the pendulum has been asserted.

While developping the mathematical models the situations liable to disturb the stability of the patient have been recognized. Such situations are handled by introduction of the so-called emergency transitions for transfering the system directly into an event in which the patient has the possibility to regain his stability.

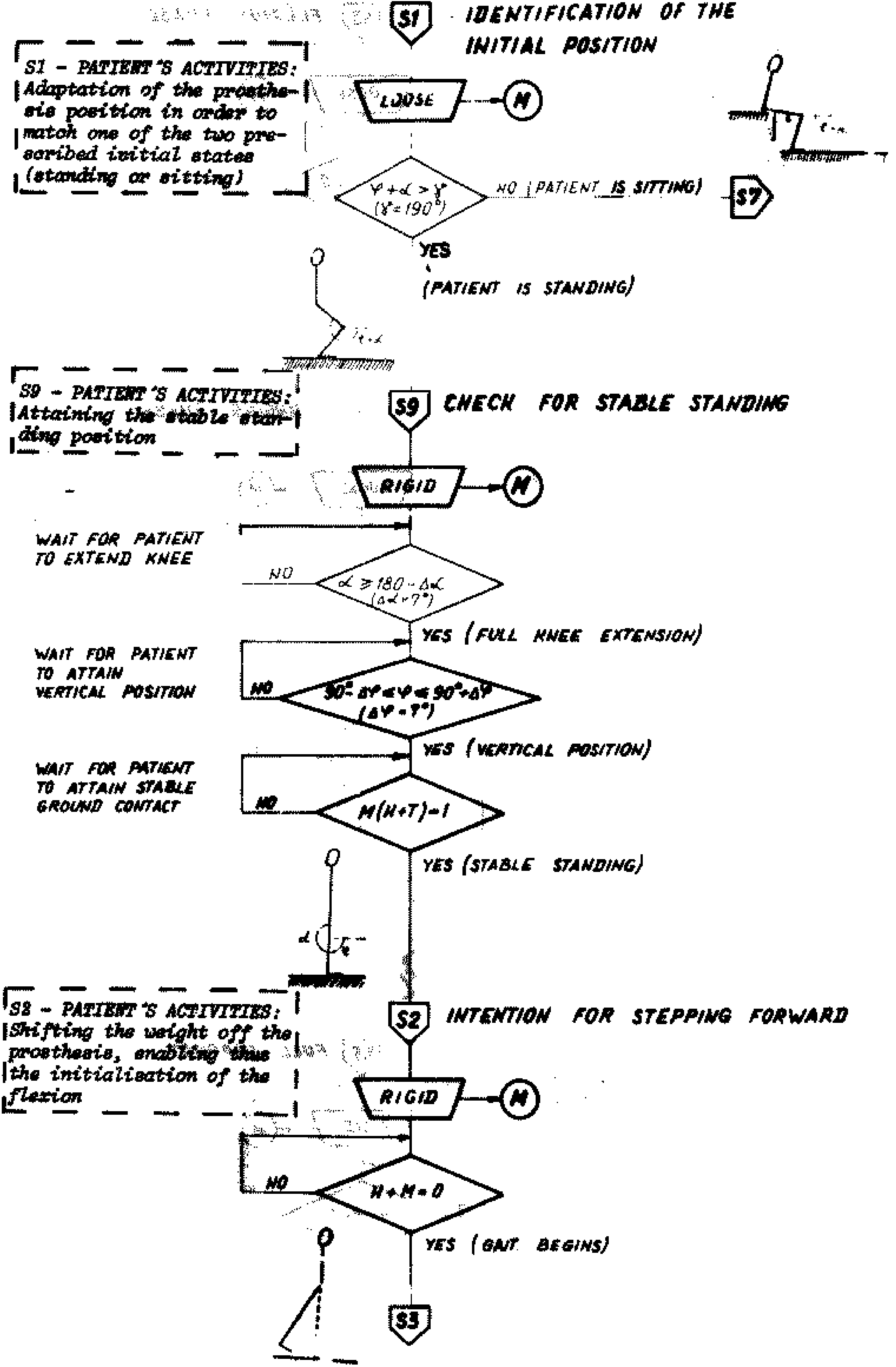
V. DECISION AND ALGORITHMIC LEVEL

The developped mathematical models for gait, sitting and getting up |8|, directly implies the overall organization of the control system as well as the distribution of control tasks between the decision and algorithmic level. The basic flow diagram which indicates the models and the corresponding control sequence is represented in Fig.3. As it can be seen the patient has the option to choose his initial position (sitting or standing) and to initialize the gait, sitting down or getting up activities. The desired activity is recognized by the control system and the control is transfered to the appropriate algorithmic subunit (module). From the functional point of view the control sequence can be divided into several subunites: stable standing, flexion and extension (swinging) make normal gait loop, while the sitting and getting up are considered as the separate parts. Since the functioning of the control system strongly depends on the patient, the patient's activities expected within the transition from the one state to another are also included in the control algorithm description (Fig.3.).

The flow of the control is straightforward and only few explanations might be necessary. Entering in any state is followed by issuing the appropriate control signal - rigid (blocked), or loose (deblocked) - to the actuator. Next, the conditions for the transition towards the next possible states are checked. The control system is looping waiting for the patient to perform required activities. As soon as one of the prescribed conditions is met the system transits into the corresponding state. At the present level only thre emergency transition are introduced. First, in the flexion phase (S3) the patient is expected to support flexion expressing at the same time his desire to walk or to sit, by pulling up his leg, or leaning on it. But if the ground contact still exists while the angle α has decreased beyond the stability limit, the patient is assumed to be in the emergency position. Therefore the prosthesis should be locked and the patient has to regain the upward stable standing position (S9). Next, during the extension phase (S4) the knee angle α may start to decraese (eg. from hitting an obstacle). If the decrease is greater then the given tolerance the emergency situation is assumed, which leads towards the blocking of the actuator (S5). The patient can retry extension or simply terminate the swinging by putting the prosthesis down on the ground, which will result in a somwhat shortened step. The similar situation may occure while getting up (S8).

All control algorithm parameters are subject to adjustments depending

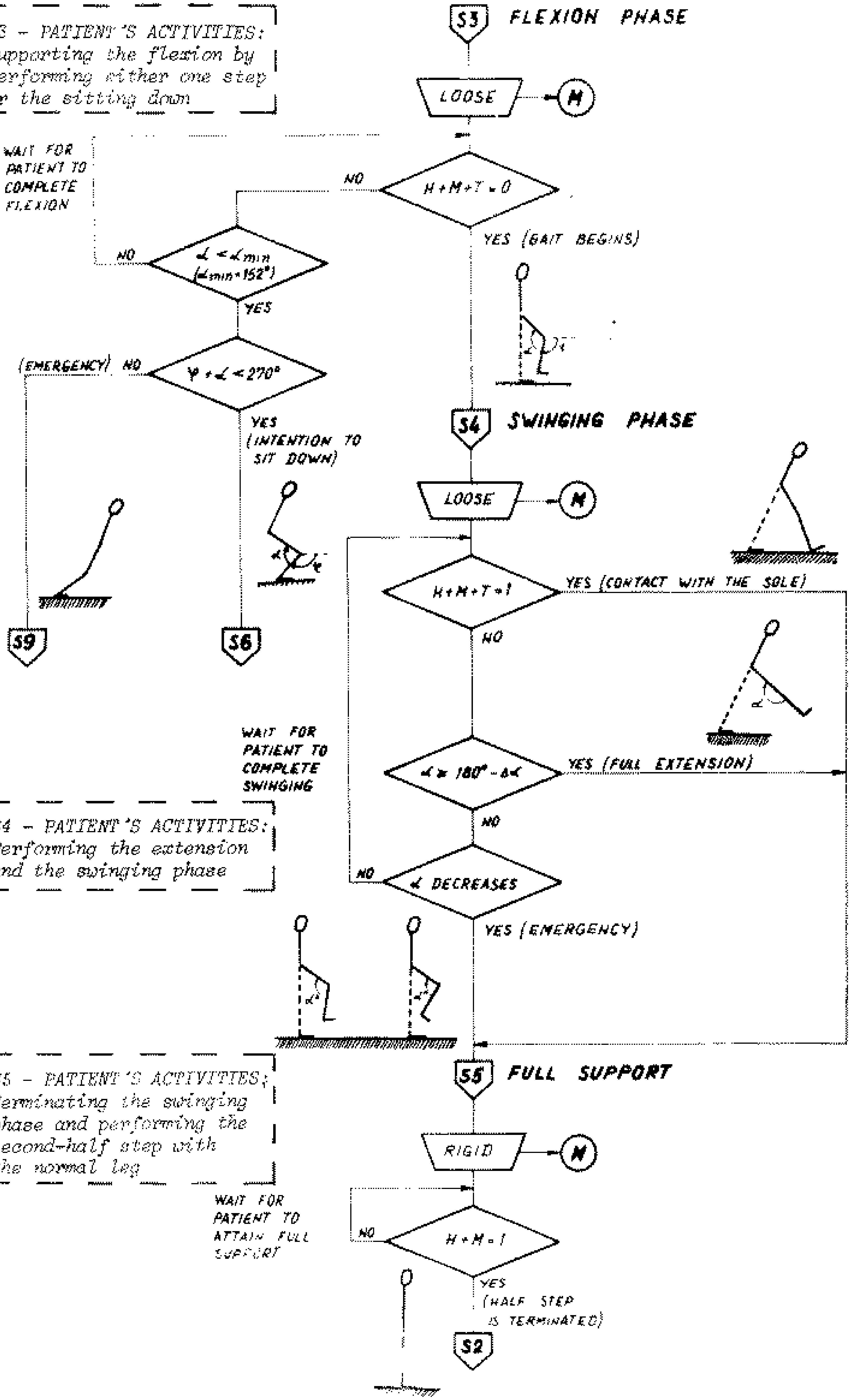
FIG.3. THE BASIC CONTROL FLOW DIAGRAM



S1 - PATIENT'S ACTIVITIES: Adaptation of the prosthesis position in order to match one of the two prescribed initial states (standing or sitting)

S9 - PATIENT'S ACTIVITIES: Attaining the stable standing position

S2 - PATIENT'S ACTIVITIES: Shifting the weight off the prosthesis, enabling thus the initialisation of the flexion



S3 - PATIENT'S ACTIVITIES:
Supporting the flexion by performing either one step or the sitting down

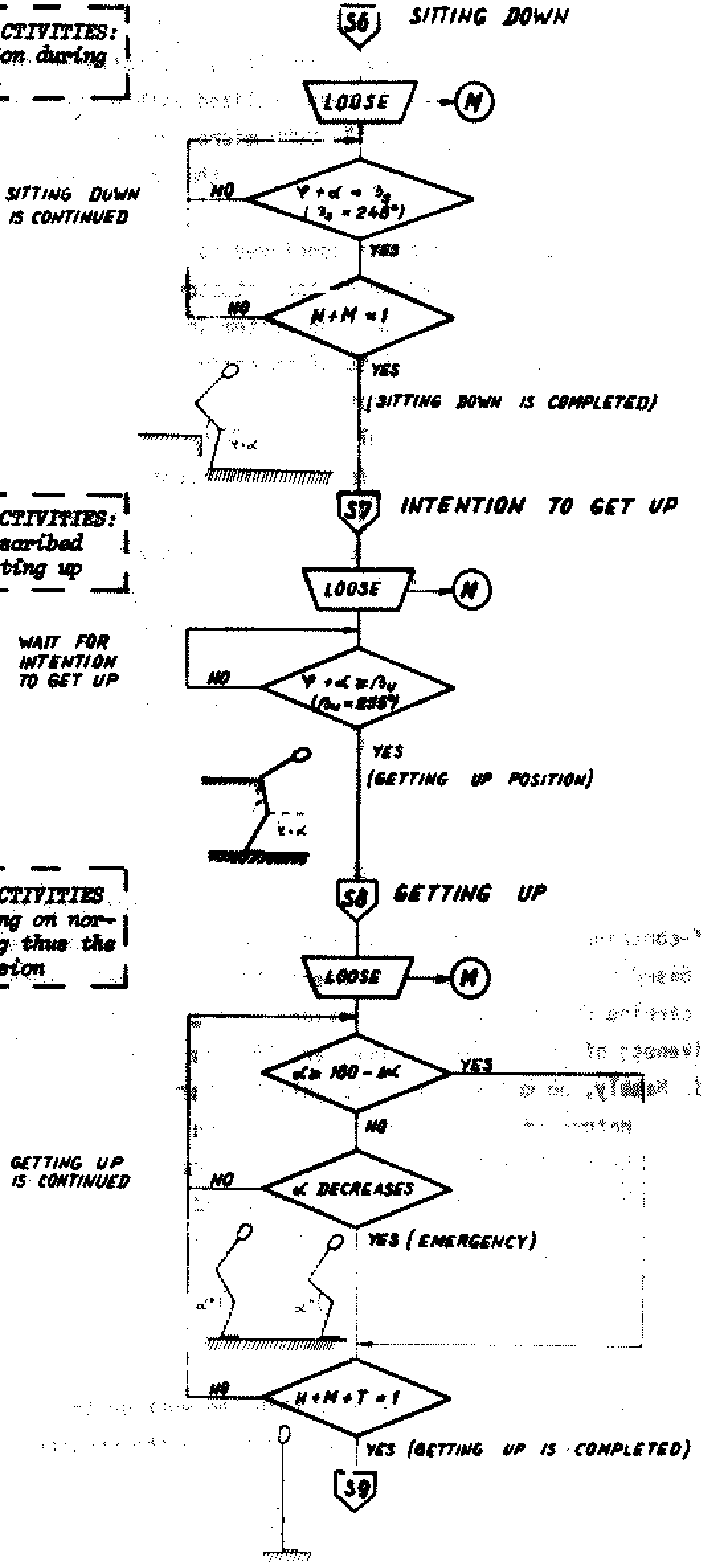
S4 - PATIENT'S ACTIVITIES:
Performing the extension and the swinging phase

S5 - PATIENT'S ACTIVITIES:
Terminating the swinging phase and performing the second-half step with the normal leg

S6 - PATIENT'S ACTIVITIES:
Supporting flexion during the sitting down

S7 - PATIENT'S ACTIVITIES:
Adopting the prescribed position for getting up

S8 - PATIENT'S ACTIVITIES
Getting up leaning on normal leg, enabling thus the prosthesis extension



upon patient's needs. This is taken care off by the software design.

The control algorithm has been realized within the programming system - LOCOM and implemented using the 8080 microprocessor. The detailed description of the LOCOM can be found in |8|, so that only some basic aspects will be underlined.

The programming system was concieved so to allow the maximum flexibility concerning the modification of the actuator. Namely, it should enable the control of the passive knee-locking system in which the transitions depend entirely upon the patient as well as of an active knee prosthesis, where the transitions are supported by the motor unit. In order to meet these requirements it was assumed that the model itself will remain basically unchanged no matter how the transitions are performed. Consequently, the control signals are generated within the separate modules which are assigned to the different actuators. The proper choice of modules is done automatically based on the type of the actuator which is actually mounted in the system.

In addition, since the computing time is not critical, the data acquisition system is also modular. Several subroutines were developped which provide the tracking of the apropriate state variables. This offers the possibility of direct testing of the measuring equipment. Besides, any change in the system model can be easily done in the straight forward manner.

VI. CONCLUSION

This paper presents a knee locking control algorithm implemented in the self-contained single board microcomputer. The design of the control system was based on the semiautomatic, hierarchical, logical approach. The experiments carried on in simulated conditions were successfull and indicates the attractiveness of such approach. However, the vast experimental work is still expected. Namely, no matter how thoroughly the theoretical work has been done the specific nature of the prosthesis area leaves many questions to be answered through the experiments. For instance, the crucial requirement for man-machine collaboration is almost unpredictable and should be viewed on the case-to-case basis. Besides, the convinient values for some gait parameters could be found only through experimentation.

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