RULE BASED CONTROL OF SEQUENTIAL HYBRID ASSISTIVE SYSTEMS (SHAS)

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ABSTRACT

The mobility improvement of spinal cord injured humans with externally controlled braces depends very much on the control applied to this complex system. A proposal to support the hybrid assistive system with sequential controller based on a finite state model and a skill based expert a system is presented here. The name sequential controller is given because of a sequential operation of uninjured structures in human, externally driven functional electrical stimulation (FES) and external mechanical brace action. This control method relies on an skill-based expert system. The skill-based expert system uses production rules methodology. Production rules are selected and designed according to specific events determined within the gait cycle called gait invariants. Gait invariants of the pathological gait associated with ambulation by paralyzed humans with hand supports were studied. A minimal set of invariants is presented with details of the possible applications to the sequential hybrid assistive system.

KEY WORDS: Sequential hybrid brace, gait invariants, gait restoration

INTRODUCTION

The proposal to support the use of FES for gait restoration by passive or active mechanical assistive has evolved into a promising rehabilitation method [1,5,7,9]. The original concept of the Hybrid Assistive System (HAS) was quite general without specifying how FES might interact with the supporting mechanical structure [11]. Since its first steps, the research in HAS progressed so much that several characteristic trends in the application of this rehabilitation method are currently clearly discernible. The term (HAS) relates to numerous different combination of bracing technique. The division of HAS into narrower categories in not only a matter of academic interest. Each trend in the application of HAS entails specific hardware and control issues. Common to all HAS designs is the man-machine interaction in order to improve the gait performance. An external mechanical brace may be activated only at specific events of the gait cycle or it can operate in parallel with FES throughout the motion activity. Present HAS systems are applied only as an extension of FES systems. This extension relates mostly to the safety features and to the prevention of some events considered as a pathological gait. However, HAS may be expected to fulfil two ambitious goals:

1) extend the inadequate FES generated step length by machine intervention,
2) transform the slow, quasi static, FES generated ambulation into a genuine dynamic process by machine intervention.

Reproduction of normal gait by HAS is intentionally not mentioned since at this time, such an ambition is not realistic. Besides, it would be very difficult, if not impossible, to define the meaning of the term "normal gait" for a man-machine dynamic system of the HAS type.

Machine intervention to support FES in the control of a individual events of a gait cycle has been predominant in the current practice. Man-machine interactions of this type are much easier for implementation both from the hardware, and, especially, from the control point of view. We shall be concerned, however, with more complex HAS designs. In this regard FES-machine interaction offers two options:

a) Parallel operation of the hybrid assistive (biological and the machine) system - HASP
b) Sequential operation of the hybrid assistive system - SHAS.

Parallel operation of biological structures and machine part of the HAS implies continues interaction of the handicapped and the machine during the complete gait cycle.

For explanation of the sequential operation of biological structures and machine in an HAS more extensive presentation is needed.

PRINCIPLES OF SHAS OPERATION

The SHAS operation consists of three phases:

1) a gait segment externally driven only by stimulated biological actuators, i.e. muscles,
2) a gait segment externally driven only by artificial actuators,
3) an intermediate gait segment within which the responsibility for motion maintenance is relayed from FES to the machine and vice versa.

Important feature of SHAS is that a segment of the locomotion cycle, depends exclusively on biological structures controlled partly by a human (volitional control) and partly by FES, while the remaining part of the trajectory is controlled by the machine and volitional neuromuscular actions. The intermediate segment serves for the transfer of motion control from biological structures to a machine. A unique controller is responsible for the execution of all phases of the gait cycle.

CONTROL OF SHAS

Taking into account that SHAS involves man as the decision maker, following hierarchical control structure is selected with following levels from the top down: 1) voluntary level, 2) coordination level, and 3) actuator level.

The basic function of the SHAS controller pertains to the coordination level. Namely, once the coordination of the joint motions is successfully solved according to the task requirements, the synthesis of the actuator control follows the standard state space procedure. The extension of the state space approach to the coordination level is, however, not straightforward. As known from the theory of hierarchical systems each higher control level requires more abstract modelling of the plant behaviour. State space equations pertain to plant dynamics, but not to joint coordination.

The modelling of the coordination level for locomotion control has been studied extensively in several papers [2,12,13]. This research has shown that the rule-based control, finite state models and skill-based expert systems are very efficient tools for the design of locomotion controllers and functional movements in general at the coordination level. This approach will be explained here.

FES CONTROLLED SEGMENT

Transfer of human skills to the computer knowledge base relies basically on identification and representation of invariant features of functional motions [11,13]. While the method works satisfactorily in modelling the gait of non-impaired persons, its application to the handicapped persons loses much of previous advantages. The reason is quite simple. Since each case of motor deficiency is, in many ways, specific, generally valid locomotion models are inadequate.

In order to overcome this difficulty, but still be able to apply the rule based control for SHAS, we tried to identify the minimum set M of invariant features which must be present in any FES controlled gait cycle. The term "minimum set" implies that no ambulation of any kind, regular or irregular, will take place if the invariant features contained in M are lacking. The set M represents, thus, just the necessary conditions for locomotion to occur. The analysis of pathological gait was used for the determination of the minimum set M. The term "pathological gait" relates to a gait of normal individuals braced with different external modules of an brace. Using different modules and their combination with the Self-fitting modular orthosis [6] the degrees of freedom were reduced in our experimental subjects. The unilateral and/or bilateral ankle, knee and hip braced locomotion with hand supports over under-elbow crutches was studied. The gait pattern was compared with the gait of paralyzed humans with FES and simple HAS systems controlled by an operators. The TV system (Sonny CCD900) was applied for the kinematic study and ground reaction forces were recorded with strain gauge transducers built in the shoe soles (Institute Jožef Štefan, Ljubljana, Yugoslavia). The EMG activity was simultaneously recorded using multichannel portable recorder [10].

Physiological studies speak strongly in favour of locomotion invariants [2]. The minimum set must guarantee that locomotion be maintained in all circumstances at any price but leaves also space for adaptivity and optimization. The basic role of biped locomotion is simply goal directed motion without falling down. In our case, invariant structure of locomotion will serve to derive formal rules which can be used for control purposes. The very nature of legged locomotion generates, thus the elements of the set M. For example. one element is

$$\text{sign}\,\Delta p(r) = 1 \quad \text{when} \quad \text{sign}\,\dot{\alpha}\,(H,l) = -1 \tag{1}$$

where Δp is the integral pressure difference between the two feet, the right one being taken as reference; $\dot{\alpha} = d\alpha/dt$ is the derivative of joint angle, the coordinate system is chosen so that with flexion the angular velocity decreases, i.e. becomes negative value; H - hip, r - right, l - left. Clearly a symmetric rule is obtained by interchanging the references pertaining to legs.

The rule (1) holds for the hip flexion phase. Maintaining the same references, the next rule is valid in the extension phase:

$$\text{sign}\,\Delta p(r) = -1 \quad \text{when} \quad \text{sign}\,\dot{\alpha}\,(H,l) = 1 \tag{2}$$

Rules 1 and 2 imply that the pressure difference between the two feet must change the sign within a walking cycle. According to rule 1, the respective leg in the "swing" phase must not be completely off the ground. It can be dragged maintaining the contact with the ground, or touch the ground intermittently, provided that the hip continues to rotate. Such a relaxed rule for the "swing" phase is essential for the acceptance of all kinds of unpredicted gait events which may occur in paralyzed persons (clonus, spasms or some other uncontrolled movement).

Further basic locomotion invariants are described by inequalities,

$$\alpha_t^f(H,l) < 0^o \;\; V \;\; \alpha_t^f(H,r) < 0^o \tag{3}$$

f stands for flexion and t for the joint angle terminal value. Rule 3 states that the hip flexion terminal angle must cross the vertical line if the forward body displacement has to take place. a is the angular displacement of the body segment from the vertical line. Positive angles correspond to leaning forward. The value 0^o stands here as the symbol for a reference line(parallel to the gravity vector), rather than an angular number. The case $\alpha_t^f = 0^o$ is excluded since it means zero forward displacement. For forward propulsion, the case, $\alpha_t^f > 0^o$ is clearly forbidden. On the other hand, any value of $\alpha_t^f(H)$ satisfying (3) is regular. Even the smallest forward steps are accepted although such gait dynamics will not satisfy the patient. In fact, one of the main roles of the mechanically driven gait segment is to extend the step length generated by FES and improve the gait dynamics.

The next basic rule reads as follows

$$\text{if} \quad \Delta p = 0 \quad \& \quad \beta_t^e - \Delta\beta < \beta < \beta_t^e \;, \quad \text{than } B(K), \text{ left or right} \tag{4}$$

B meaning locked joint state; β the knee angle; β_t^e the terminal extension angle which must not correspond to fully extended knee; $\Delta\beta$ is the knee bounce which takes care of shock absorption response and K is the knee articulation. This is condition sine qua non to maintain the upright position. Evidently, it is desirable to have the knee fully extended when reaching the terminal leg position, but locomotion is possible even with a bent knee provided that its rigidity is controllable. Again, an unsatisfactory knee extension terminal angle can be corrected by the active brace.

The above rules define the set M= {1,2,3,4} which describes formally any pathological walking process. For a normal subject, the set M represents minimum task conditions which must be extended in order to arrive at solutions optimal in terms of biological and biomechanical requirements.

The knowledge base for the FES controlled segment consists, thus, of a core of four rules. By monitoring the gait process with the aid of these rules, the controller is able to decide if the conditions for propulsion are fulfilled, i.e. if FES output did provoke the expected muscular response. Thus, the monitoring level has to decide if the man-machine system is functional at all or not. Then, appropriate instructions are sent to the lower control levels.

FES CONTROL

The FES system developed for SHAS system consists of a multichannel biphasic stimulator. Constant current pulses can be supplied with surface or implanted electrodes. The pulse duration and pulse amplitude modulation can be controlled at the local control level. Local control level includes artificial and natural sensors. The term natural sensor is used to express the EMG monitoring from the stimulated muscle. The specific circuitry, based on sample/hold amplification is capable to reject the stimulation artifact and M wave and to provide an accurate information of the efficacy of the stimulation. The general level of stimulation is graded upon the type of locomotion and it is determined at the strategic control level. The EMG feed-back provides additional information on the fatigue of the stimulated muscle, and enables the switching to an external mechanical brace. The artificial sensory feed-back is used as an information of the length of the muscle. The length of the muscle and EMG activity correspond to the level of the produced force. Each of the channels of the stimulator is isolated and independent simultaneous stimulation is possible. Cross-talk between stimulation channels is reduced in comparison with monopolar stimulation using a common anode and interleaved pulses.

The number of channels needed for locomotion is still unanswered. The research on the FES for walking [3] points out that it may be necessary to control most of the legs and trunk muscles, even some of the muscles which are under voluntary control in order to synthesize walking. Even this speed of the gait is impressive in comparison with other techniques [1,3,5,7]. The distance of the gait is still limited and the pattern of the movement very much resembles robot walking. Present experience points out that each of the medical indications considered for gait restoration requires a specific stimulation configuration. Our SHAS consideration is limited to stimulation of functional movements such as hip flexion, withdrawal reflex (Peroneal nerve stimulation), extension reflex (Tibial nerve stimulation), knee extension (Quadriceps stimulation), hip extension and adduction with some external rotation of the hip (Gluteus medius and maximus m. and Hamstring stimulation).

EXTERNAL BRACE CONTROL

The role of the active brace in SHAS, as stated in Introduction, is to increase the locomotion speed and improve gait dynamics. The active brace, in our case, is controlled by artificial reflexes organized in a skill based expert system. Design principles of skill based expert systems for rehabilitation purposes have been described in the literature [7,8,13,14]. However, the controller for SHAS must satisfy additional requirements so that it cannot be designed in a straight forward way. In fact, complex and challenging control problems must be solved in order to make the SHAS function as intended. Namely, the transition from biological to mechanical actuators, and vice versa, is not trivial at all. It is necessary to identify, in the swing phase, the overlapping interval when the motion driven by biological actuators should be replaced or supported by mechanical actuators. The same applies to the stance phase with the difference that the mechanical actuators are now handing over the propulsion task to the muscles. As will be seen, this transfer has its own specific control aspects.

The external brace in our case, as was said, is the Active Self-Fitting Modular Orthosis.

CONTROL TRANSFER IN THE SWING PHASE

In the swing phase, the SHAS controller must take care of the following tasks:

a) to identify individual joints which need external power;
b) to identify the state of the FES output at which the transfer should occur,
c) to activate the supporting mechanism.

The structural diagram of the SHAS controller is seen in Figure 1. Since rule based FES and brace control relies on previous designs, only issues related to the intermediate locomotion phase will be discussed here. The term intermediate relates to state when FES controller and external brace controller are working in parallel.

The transfer logic consists of following rules:

At any instant, M rules must be valid to make the forward body motion happen. Hence, all M rules act as monitoring input to the transfer logic and are matched continuously with the corresponding patterns stored in the knowledge base.

Rule 5. If the matching result of M rules in the FES driven phase is 1, than the brace controller is off. "1" mean true in a boolean expression and "0" mean false.

Rule 6. If the matching result of M in the FES driven phase is 0 than the brace controller is on. This rule assures the regular evolution of the gait cycle.

The interval when a rotating joint under FES control needs external power in order to improve the gait quality can be detected by monitoring its speed.

Rule 7. If the angular speed $\dot{\alpha}_n$ of the hip, knee or ankle joint becomes less than threshold parameter b_n

$$\dot{\alpha}_n < b_n , n = 1,2,3$$

than transfer to the brace controller.

An important element is still missing from the set of transfer rules. When the need for the external support of a joint is detected, several options remain still open to the controller (flexion or extension). In addition to joint states, the brace controller must be also informed about the output of the FES controller at which the transfer has to take place. In other words, the initial state of the brace controller must correspond to the current state of the FES controller.

CONTROL TRANSFER IN THE STANCE PHASE

Once the brace controller has taken over the task to improve the inadequate leg motion, the FES subsystem is in the stand-by state. Since the optimization goal is to maximize the participation of the patient in the locomotion process, attempts must be made in the backward walking as well to transfer a segment of the gait trajectory to FES. However, inverse control transfer is not a replica of the forward one. Rule 7 is now not valid since the brace controller is supposed to function correctly. Indication to switch over to the FES control is, thus, lacking. Facing, besides the unreliable response of the biological system, the trial-and-error method has been chosen to transfer the control from the machine to FES in this case. In the vicinity of the hip angle where the brace controller has taken over the control task in the forward leg motion, the external power supply is switched-off and FES control activated. As long as the controller obeys the inequality $\dot{\alpha}_n > b_n$, the brace controller is in the stand-by position. When the respective joint rotation speed is below the threshold b, the mechanical subsystem is in charge of the motion. The outcome of such a control transfer may be a few oscillation between the FES and brace controller before the desired terminal leg position is reached.

IMPLEMENTATION

Basic parts of a SHAS system are seen in Figure 1: the FES stimulates the active brace and the rule based controller. The choice of the FES subsystem is dependent upon the nature of motor deficiency. Different micro-computer based or specially designed electronic stimulators may be used as the component of SHAS. Interfacing with other parts of the control system requires only that the current discrete state of the FES stimulator be applied to the M-monitor of the brace controller. Output of a skill based expert system is most easily fed to cybernetic actuators since they have four controllable states: flexion, extension, blocked, loose [6]. Concerning the brace, we have used the active modular orthosis fitted with cybernetic actuators at the hip and the knee [8]. Such an orthosis may be without any alterations as a unilateral or bilateral assistive system. This equipment is also commercially available in limited quantities. The same applies to sensors. Only those sensors which are used anyway in an active modular orthosis are needed in SHAS: angle, contact, pressure and vertical line transducer.

An essential part of the SHAS is the software. In addition to the knowledge base pertaining to orthosis control the software must take care also of the monitoring function and the transfer control. Although the available expert system packages may be utilized it turns out that it is more efficient to generate a goal oriented software. Such a software must meet two requirements: to assure real time monitoring of joint trajectories and to allow man-machine interaction. Namely, optimal SHAS performance cannot be fixed in advance. The main adjustment comes only when the handicapped is actively involved himself in locomotion. This entails easy, supervised rule and parameter corrections in real time.

CONCLUSION

The main goal of this paper was to introduce a new area of hybrid system application. In our opinion, integration of FES and active mechanical braces fitted with cybernetic actuators offers a powerful tool to assist severely handicapped patients. Control aspects of sequential hybrid assistive systems were the exclusive concern of this paper. Namely, the design of the SHAS control system, which is the challenging problem in itself, is the prerequisite for any further work. Patient selection and clinical experience will be elaborated in a separate publication.

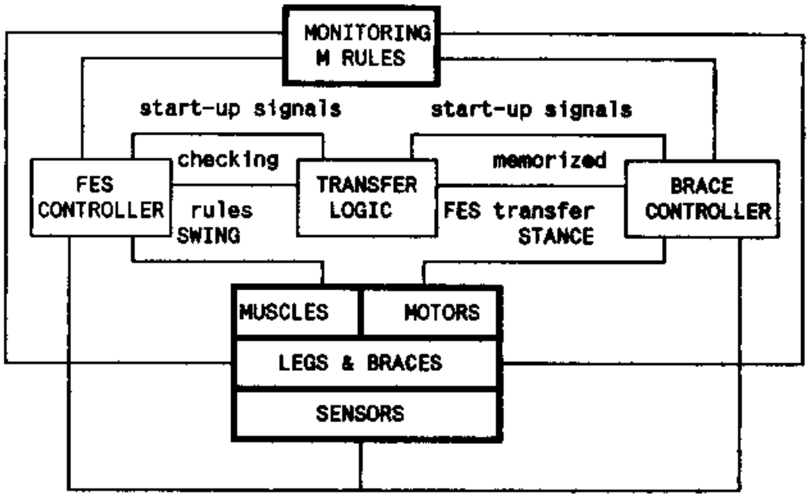

Figure 1. The model of the sequential hybrid assistive system

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