Towards a Hybrid Motor Neural Prosthesis for Gait Rehabilitation: A Project Description

Heike Vallery and Martin Buss

***Abstract*— This paper describes the concept for a cooperatively controlled combination of Functional Electrical Stimulation (FES) with a motor-driven exoskeleton for gait rehabilitation in hemiplegic patients. The objective is to combine the therapy advantages of FES with a mechanical bracing system, which provides sufficient stabilization and supports a physiological gait pattern. A multi-layered, switching control architecture based on predictive optimization is designed, which allows the patient to control his movements *via* an intuitive interface. The benefit of the proposed system is tested and evaluated in real scenarios with patients using a prototype.**

***Index Terms*—Neural prosthesis, walking, rehabilitation, hybrid system.**

I. INTRODUCTION

IN GAIT therapy for stroke patients, the predominant aim is restoration of mobility and independence. In clinical practice, diverse therapy methods are applied, which can mainly be divided in two superordinate groups:

- External guidance or supported motion, e.g. by a physiotherapist or an exoskeleton.
- Artificial induction of muscle activity to provoke motion, e.g. through Functional Electrical Stimulation (FES).

Several concepts have been developed to support the lower extremities during gait with motor-driven orthotic devices, called exoskeletons. Especially in Japan efforts have been made recently on this field. The HAL system [1] can compensate the lack of muscle strength by a 2-DOF exoskeleton and a force augmenting control. This is done *via* electromyographic measurement of remaining muscle activity. It can provide superhuman forces similar to the BLEEX system [2], developed to allow carrying of heavy loads. The commercially available Lokomat [3] is clinically applied for gait rehabilitation on a treadmill. It allows patients with strong lesions to walk in a very early stage of therapy, since it incorporates a suspension system and does not require balance or the ability to stand freely. The advanced gait trainer offers a feasible alternative [4]. It uses footplates, which are operated by a doubled crank and rocker gear system.

Functional Electrical Stimulation (FES) has been investigated for decades (e.g. [5,6]) and applied for neural prostheses. However, severe problems arise with artificial electrical muscle activation: Electrically stimulated muscles are not recruited in a physiological manner, which leads to substantially increased fatigue. Surface stimulation, which does not require invasive application, cannot offer sufficient selectivity. Furthermore, in hemiplegic patients, sensory-motor mechanisms are modified and the muscles do not respond in the same way as the muscles would in able-bodied humans. Modified reflexes produce major problems since stimulation triggers unwanted responses.

Investigations on closed-loop control of FES have shown that linear PID controllers are unsuitable due to delays in the system, so control must use predictive and adaptive methods.

FES in clinical practice is predominantly applied on its own only in an advanced therapy stage, where the patient can already stand freely with a crutch or a cane, e.g. if only foot clearance during swing needs to be assisted.

With a complementary exoskeleton, the disadvantages of FES can be overcome. The restoration of motor control is supported better by a combinative therapy than by mere external guidance, which has been shown in [7]. The hybrid motor neural prosthesis for the application in paraplegic patients was introduced in early eighties by Tomović and Popović and later by Andrews [8] and colleagues. Solomonow and colleagues also investigated the application of FES with un-powered walking orthoses for paraplegics [9]. Popović *et al.* developed powered walking orthoses and introduced the concept of hybrid control [10, 11]. One important finding from these studies is that a powered orthosis has to be adaptive and lightweight and that in most cases it will not be accepted by patients.

This work investigates a cooperatively controlled combination of FES and an exoskeleton for gait rehabilitation in hemiplegic patients, especially stroke victims. The project focus is therefore substantially different from the previously mentioned investigations concerning paraplegia, since post--stroke patients on the one hand have remaining motor activity to be considered and on the other hand might accept a walking aid more readily, since it offers them not only functional walking, but also the benefit of rehabilitation. Two objectives are thus pursued: A short-term increase of mobility and independence, and a long-term training of muscles and rehabilitation of motor control.

The focus is at first to permit the essential motion

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Heike Vallery is with the Institute of Automatic Control Engineering (LSR), Technische Universität München, D-80290 München, Germany Heike.Vallery@tum.de,

Martin Buss is with the Institute of Automatic Control Engineering (LSR), Technische Universität München, D-80290 München, Germany M.Buss@ieee.org

sequences standing up, stepping on level ground and sitting down. Since individual demands and progress of each patient need to be considered, the support must be variable. In addition, the system must encourage autonomous patient activity and therefore only offers as much support as necessary.

The control of the system is to a large extent transferred to the patient *via* a quickly learnable interface. Suitable input and output parameters will be investigated in order to let the patient communicate his intention and to provide him with biofeedback about his motion.

This paper contains an outline of the hierarchical control concept, followed by a more detailed description of the constituent layers.

II. METHODS

A. Hierarchical Control Concept

The neural prosthesis is divided into four parts (Fig. 1):

1) **Patient interface,** consisting of Input Device and Output Device.
2) **Motion planning.** The High-Level Control supervises the gait cycle and generates reference trajectories for angles and joint torques.
3) **Low-Level control** for the coordination of the redundant actuators (muscles and motors) with integration of voluntary motor activity of the patient.
4) **Hardware:** actuators (muscle stimulator and exoskeleton) and sensors.

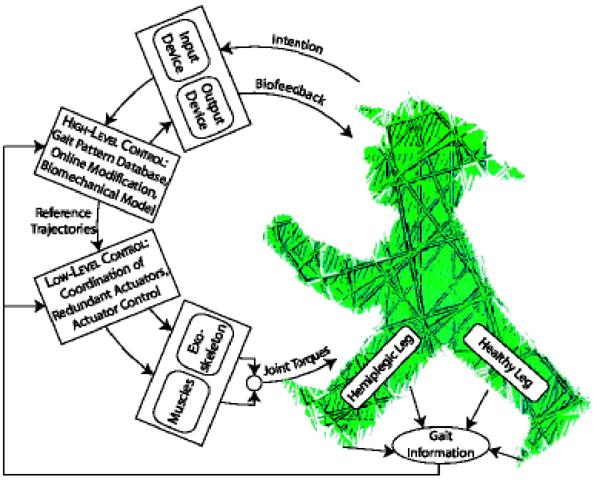

Fig. 1: Components of the neural prosthesis

The first and the last two items are mutually strongly correlated. First the motion planning under patient participation will be discussed, followed by the concrete realization of the communication between system and patient *via* the patient interface. After a description of hardware components, the low-level control will be presented, which realizes the demands of the motion planning by means of these hardware components.

B. Motion Planning (High-Level Control)

The healthy human gait is a continuous feedback control process, where an adaptation to an almost arbitrary environmental situation is performed on the basis of a broad experience database. With the help of the so-called extrapyramidal motor system, motion patterns can be performed almost subconsciously, which only need to be adjusted, e.g. to overcome obstacles or to change speed. In a stroke patient, the brain area containing motion control often is disturbed. Following the process in healthy gait, the neural prosthesis offers pre-defined trajectories, which the patient can modify during gait.

For biped robots, stable gait patterns are provided in a similar way by offline trajectory calculation and online modification [12]. Results from these investigations can therefore be incorporated in the design.

A step can be divided in several phases. For one leg, there are three states: single support (stance on this leg), double support (stance on both legs) and swing. “Stance phase” does not denominate a static state, it merely refers to contact between foot and ground. For the different phases, different control strategies are applied, handled by a superordinate switching control.

The proposed motion planning disposes of three components: a database with pre-specified trajectories, a reference generator, which determines reference signals for the joint angles based on comparable trajectories, and a simplified biomechanical model, which calculates the corresponding joint torques.

The *database* is filled before initiation with trajectories of comparable healthy subjects. During gait, it adapts these trajectories gradually to the patient's individual gait pattern by processing data from the healthy leg. This is only possible after some time and iteratively, since the paralyzed leg affects the motion of the healthy leg as well.

The *reference generator* processes on the one hand signals from the sensors of the healthy and the hemiplegic leg to detect the momentary gait phase, on the other hand it processes the patient's requests from the interface. Based on the recent angle and velocity trajectories and comparable trajectories from the database, reference trajectories are deduced. E.g. the deceleration of the healthy leg during swing is interpreted as the intention to walk slower, thus the motion of the hemiplegic leg is decelerated as well. This intention detection can already generate a certain pilot control to relieve the patient. He can always directly interfere by transmitting his desires *via* the interface, the reference trajectory is then modified correspondingly. That way the patient can actively modify the trajectories without being overloaded by control tasks.

With the help of a *biomechanical model* of the leg, joint torques can be calculated corresponding to the joint angle trajectory. Both angle and torque trajectories are forwarded to the low-level control. In the other direction, the motion planning communicates with the output device by providing processed information about the current state to the patient.

C. Patient Interface

1) Input Device: Without an active intervention, the control is already capable to a certain extent to deduce the patient's intention on the basis of the motion of the healthy leg. In addition, the option for an active patient intervention has to be given, especially for sudden stops, for standing up and sitting down, as well as in the case of intention misinterpretation.

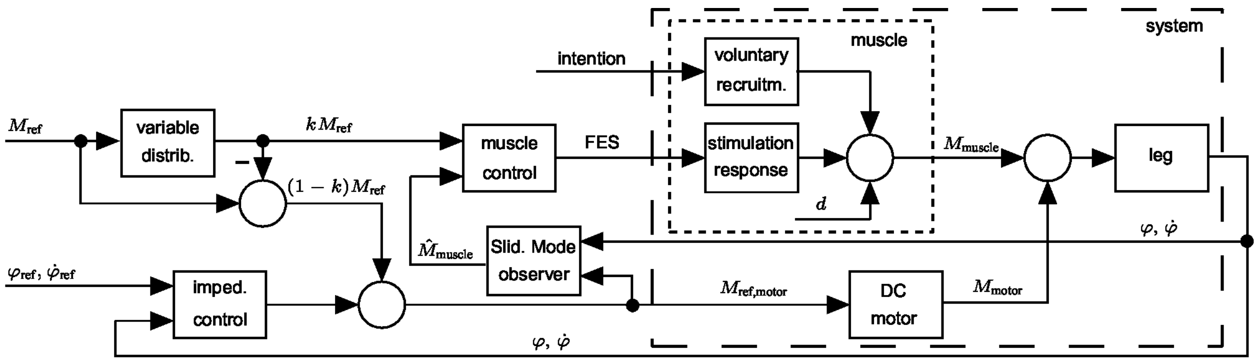

Fig. 2: Concept for the Low-Level Control. Muscle stimulation is embedded in a torque control loop and provide a variable fraction of the necessary torque, the exoskeleton DC motors ensure trajectory tracking.

The input device has to be intuitive and may not strain the patient excessively. A hemiplegic patient can generally only control half of his body and needs the unaffected arm for support on a walking aid. Many patients actually latch onto their crutch. Thus, arm and fingers can only be of very limited use for the control of the neural prosthesis.

In the proposed system, the contralateral, unaffected leg is predominantly used for control. The patient can use degrees of freedom which do not affect his functional gait, such as a slight external rotation of the leg during swing or a change in foot rolling during stance. This way he can indicate his intention for the step with the affected leg, e.g. stopping or increase of foot clearance. The configuration of this patient control will be object of research and investigated with experiments.

2)Output Device and Biofeedback: With regard to a suitable biofeedback two questions have to be answered:

- Which information does the patient need in order to control his motion?
- Which way is this information provided?

Essential information in healthy gait are the contact forces between foot and ground during stance. After a stroke, this information lacks on one side, which is one reason why often a disturbed balance can be observed. Feedback about the position of the center of mass could improve the patient's perception of his body and his ability to control his gait. Information about balance is contained within the ground contact forces. Therefore, these contact forces are processed and made available to the patient in a simplified way.

Several approaches concerning the form of feedback will be evaluated, including a haptic display, sensory substitution with vibratory [13] or electrical feedback, and acoustic feedback. A haptic display bears the disadvantage of high complexity, but demands the least degree of abstraction from the patient. For an everyday application, a minimization of complexity needs to be pursued, so acoustic feedback would be the choice. Furthermore, the sense of hearing is the only sense which is always active. Thus, it could be sensible to apply various methods parallel during a phase of adaptation, and to offer merely acoustic feedback in the end. To improve the intuitive learning, not only the reaction forces from the hemiplegic leg are fed back, but also those of the unaffected one.

3) Test and Evaluation of the Patient Interface: During the development of the interface, preliminary studies can be performed with healthy subjects. The experimental setup consists of the prosthesis with a special shaft, which can be worn by a non-amputee with her leg bent. In this way the actuation of the knee joint can be performed by a motor to test the input device. Furthermore, the foot is insensible to foot-ground contact, so artificial biofeedback can be tested.

D. Hardware: Actuators and Sensors

The exoskeleton must control at least two degrees of freedom, i.e. exert torques on hip and knee. This way the most important movements for gait can be produced. In such a configuration, weight shifting from one leg to the other has to be performed by the patient, since this requires hip ab- and adduction. Additional passive elastic mechanisms may be included to support and constrain hip motion and to stiffen the ankle joint. Later an extension is possible which also includes actuation of the lateral hip motion.

A neural prosthesis experimental setup [13], originally developed for paraplegic patients, can be used. To measure the joint angles, combined angle and angular velocity sensors (goniometer-gyroscopes) are utilized. The contact forces are measured with insoles.

E. Low-Level Control

The low-level control fulfills three tasks: actuator control, actuator coordination, and integration of voluntary motor activity from the patient. The patient can interact with the system in two different ways: On the one hand through commands *via* the interface, on the other hand through voluntary muscle activity in his affected leg. The rehabilitative control strategy does not consider this as a disturbance, but encourages autonomous activity whenever it is coordinated with the intention. Thus, the control is designed such that the reference motion, consciously transmitted *via* the interface, forms a frame in which autonomous activity can take place.

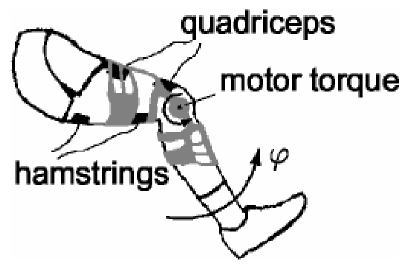

Fig. 3: Experimental Setup for the Low-Level Control: An actuated orthosis and artificially stimulated muscles provide torques to control the knee joint angle.

The reference torque is distributed *via* an adaptive convex combination, such that not the entire necessary torque is to be realized by the muscles, but only a variable fraction k. The rest, $(1\text{-}k)M_{ref}$ is used for a feed-forward pilot control of the DC motor. The factor adapts to the muscle's capabilities and different levels of fatigue, such that the muscles are challenged to a suitable extent.

Through impedance control, the exoskeleton does not exert any forces as long as the patient moves his leg close to the reference trajectory calculated by the motion planning. In case of deviation, the patient feels a correction of his motion, similar to the support of a physiotherapist. The impedance needs to be adjustable to different levels of patient autonomy.

Coordinated with the exoskeleton motion, the patient's muscles are stimulated, whereat the joint torque produced by the muscles is controlled. This torque depends nonlinearly on various factors such as stimulation parameters, joint angle, joint velocity, and fatigue. The predictive control approach is based on identification and simplified modeling of the muscle's activation dynamics and recruitment curve and an observation of the exerted torque $\widehat{M}_{muscle}$.

The Sliding Mode observer contains a model of the DC motor and the leg and deduces muscle activity *via* a residual analysis. For reliable estimation, the actuator dynamics as well as biomechanical properties of the leg (segment inertia and passive joint torques) are modeled as accurately as possible.

The contribution of voluntary patient activity is deduced *via* internal observation in the muscle controller and is integrated into the predictive optimization algorithm to determine adequate complementary stimulation. If the patient shows active participation, the closed-loop strategy decreases stimulation intensity. In this procedure, the difference in muscle fiber recruitment between voluntary control and artificial stimulation must be taken into account.

Since the observation of patient activity and FES-provoked torque depends on the identification of muscle properties and on the predecessing muscle torque observer, the calculated contributions produced by FES and patient activity are not too reliable. A possible way out of this problem would be the use of EMG, which can already be performed well even in the presence of electrical stimulation [14], but this increases complexity furthermore. The resulting modeling error is combined with other disturbances, e.g. originating from spasticity, and the sum is represented by a single disturbance d. These errors in the calculation of the optimal torque, disturbances in muscle stimulation and uncoordinated patient activity can be compensated to a large extent, since the reliable DC motors of the exoskeleton are responsible for trajectory tracking.

The low-level control strategy is first evaluated in a one-dimensional experimental setup for the knee joint [15], which incorporates an actuated orthosis and FES stimulation of hamstrings and quadriceps (Fig. 3).

III. CONCLUSION

A concept for a hybrid neural prosthesis for gait rehabilitation in stroke patients has been presented. Focal points in the following investigation will be the patient interface and the cooperative control of the redundant system.

The motion sequences standing up, sitting down and stepping on level ground do not by far exhaust the possibilities of such a system: In contrast to a wheelchair, legged walking can conquer obstacles and stairs. A later project phase can therefore deal with an extension of the gait patterns.

Another possible extension of the application concerns the user group, the results are e.g. transferable to paraplegics and amputees.

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