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Voluntary Motion Support Control of Robot Suit HAL Triggered by Bioelectrical Signal for Hemiplegia

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Abstract— **Our goal is to enhance the quality of life of patients with hemiplegia by means of an active motion support system that assists the impaired motion such as to make it as close as possible to the motion of an able bodied person. We have developed the Robot Suit HAL (Hybrid Assistive Limb) to actively support and enhance the human motor functions. The purpose of the research presented in this paper is to propose the required control method to support voluntarily motion using a trigger based on patient's bioelectrical signal. Clinical trials were conducted in order to investigate the effectiveness of the proposed control method. The first stage of the trials, described in this paper, involved the participation of one hemiplegic patient who is not able to bend his right knee. As a result, the motion support provided by the HAL moved the paralyzed knee joint according to his intention and improved the range of the subject's knee flexion. The first evaluation of the control method with one subject showed promising results for future trials to explore the effectiveness for a wide range of types of hemiplegia.**

I. INTRODUCTION

For patients with paralyzed extremities caused by a cerebral vascular disturbance like stroke, rehabilitation is conducted in order to recover motor function of the extremities during the acute and convalescence stages. For the chronic stage, rehabilitation is conducted in order to preserve the motor function as much as possible. However, it is not conducted therapeutically [1]. After the chronic stage, the patients who have motor paralysis will live their life using the residual function, making their daily life uncomfortable. For instance, patients who have difficulties from hemiplegia walk with a circumduction gait or a foot drop gait. These gaits are quite inconvenient to walk with in their daily life.

It is important that patients can live an independent life by compensating lost motor function using their residual function or an orthosis and cane, even if the compensated motion is different from an able-bodied person's motion. However, it would still be ideal for them to perform motions closely resembling an able-bodied person's motion. Here we suggest motion supports, by using a wearable robot. Wearable robots make it possible to support the impaired motion by creating external forces onto the paralyzed limb.

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The motion supports are designed to be as closed as possible to the motion of an able-bodied person. This motor function compensation by using wearable robots will thereby enhance the quality of life of persons with paralyzed limb. Recent examples of wearable robots focused on the motor function compensation include the development of active orthoses and active prostheses [2]-[4]. Currently, one of the most important technical problems for these wearable robots is to obtain the user's motion intention. Moreover, applications of the wearable robot to patients with paralyzed lower limb have been hardly reported.

We have developed the HAL (Hybrid Assistive Limb) to enhance and support the human motor function [5], [6]. The HAL is a wearable robot designed for a wide range of applications, from welfare to heavy work support, and built in several versions (full body version two legs version and one leg version). In order to provide motion support according to the wearer's intention to move, we have developed a voluntary controller that uses the wearer's bioelectrical signals [7]. These signals are based on the myoelectric signal detected on the skin surface of the supporting muscles.

However, this voluntary controller is not well equipped to support the impaired motion of hemiplegia patients because the behavior of the myoelectric signal obtained from the paralyzed limb is different from that obtained from a healthy limb, and it is difficult to relate the myoelectric signal from the paralyzed limb to an adequate assistive torque.

The purpose of this research is to propose a voluntary control method that compensates the impaired motion of the paralyzed limb of hemiplegia patients and confirm the effectiveness of this method through experiments.

Section II describes the HAL systems. In section III a proposed controller are presented, followed by the experiments conducted with a person with hemiplegia patient (section IV). Section V contains the experimental results, and section VI discusses the effectiveness of the HAL in the context of hemiplegia. The paper's conclusion are presented in section VII.

II. HAL SYSTEMS

In this research we use the single leg version of the HAL, which was designed specifically for hemiplegia users. The external appearance of the HAL is showed in Fig. 1. The HAL basically consists of power units, the main controller,

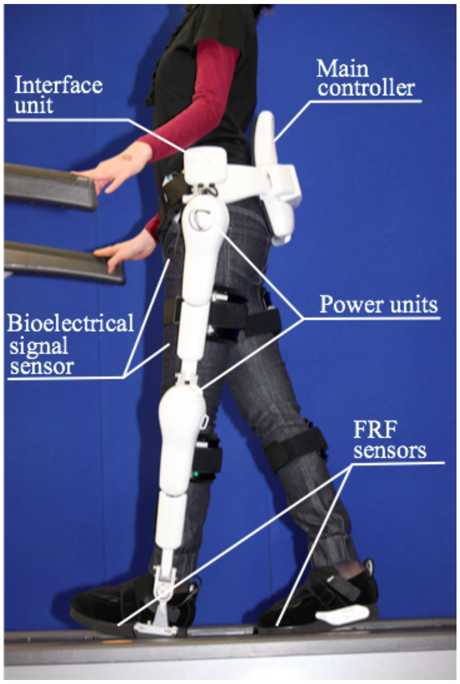

Fig. 1. Single Leg Version of the HAL

interface units (to allow the wearer to adjust the HAL behavior) and the sensing system. Furthermore, safety for the HAL and its users has been considered in the HAL's design.

The exoskeleton of the HAL system is an anthropomorphic structure designed to support the mechanical functions of the human lower body. It consists of a frame with active and free joints, and is attached to the user's hips and legs using cuffs and belts. The active joints attaching actuator (hip and knee) have one degree of freedom (DOF) in the sagittal plane. The free joints of the hip and ankle have 2 DOF in frontal and horizontal planes respectively. The required torques for motion support are generated by the power units of the active joints. Each unit integrates an actuator, a motor driver, a microprocessor and a communication interface. The motion support is achieved by transmitting the torques of the power units to the user's legs through the exoskeleton frames. One power unit is required for one joint. Actually, The HAL single leg version is thus equipped with two power units.

The control of the HAL system is ensured by the main controller. Its purpose is to control the power units, monitor the batteries, communicate with the system operator and supervise the safety of the users and the HAL. It modulates the assisting torque of each power unit in order to perform motion support for walking, standing up and sitting. Furthermore, it sends the sensors information and the system condition to the remote monitoring system, which provides a visual feedback to the operator and allows him/her to adjust the system parameters remotely.

The interface units contain an interface to adjust the parameters of the HAL, including the push buttons to tune the assistive gain. This interface allows the wearer to adjust the assist torque depending on his/her physical condition or desired level of comfort.

The HAL is equipped with a sensing system based on several types of sensors to detect the HAL's state as well as the wearer's muscle activities. Potentiometers are mounted on each joint of the HAL and used as angular sensors to measure the joint angles. The bioelectrical sensors are attached on the skin surface of the extensor and flexor muscles of the knee and the hip joint to detect their activity. Each shoe's insole contains two floor reaction force (FRF) sensors to measure the FRFs generated at the front and the rear of the foot (heel and ball areas). Furthermore, a gyro sensor and an acceleration sensor are mounted at HAL's body trunk to measure the absolute posture of the HAL based on the direction of gravity.

It is important to implement safety for the users and the HAL into the design of the HAL. As for safety mechanisms and functions, the HAL has mechanical joint limiters to remain within the human joint ranges of motion, and electrical limiters to limit the current range of each power unit. Furthermore, software safety components were integrated to detect malfunctions during motion support.

III. CONTROL METHOD

In this research, the motion support is focused on knee flexion support for disabled persons who have difficulties bending their knee due to hemiplegia but whose bioelectrical signals can still be detected from the impaired knee flexor. The control method proposed in this section mainly consists of command signal and torque generation.

A. Command signal

A command signal is used to trigger the motion support of the HAL. This signal can be controlled by the wearer's intention to move as follows. It is a binary signal obtained from the bioelectrical signal of knee flexor, BES_{kf}, and triggered when the wearer intends to bend the knee joint. The signal BES_{kf} is the signal detected by a bipolar electrode and rectified to full wave. In the following, the command signal, CS_{kf}, becomes 1 if the smoothed bioelectrical signal of knee flexor, $sBES_{kf}$, exceeds a threshold, BES_{kf_th}.

$$CS_{kf} = \begin{cases} 1, & \text{if } sBES_{kf} > BES_{kf\_th} \\ 0, & \text{otherwise} \end{cases} \tag{1}$$

The motion support for the knee flexion is triggered when CS_{kf} becomes 1. If $sBES_{kf}$ does not exceed T_{kf}, the command signal is kept at 0 and the motion support is not triggered.

B. Torque generation

When the HAL action has been triggered, an assistive torque is provided to the patient's knee joint. The torque generated by the HAL $\tau_{hal\_kf}(t)$ is described as follows,

$$\tau_{hal\_kf}(t) = \tau_{ka}(t) + \tau_{kd} + \tau_{kg} \tag{2}$$

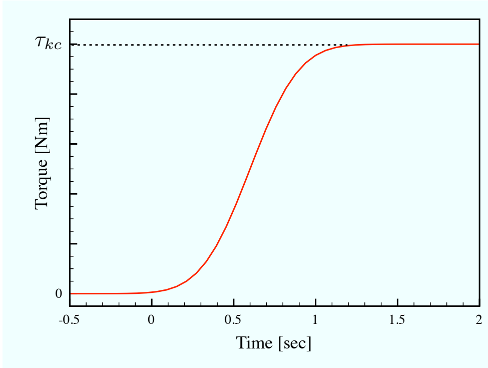

Fig. 2. Assitive torque pattern

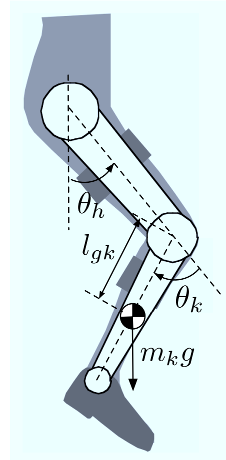

Fig. 3. Lower model of HAL and human

The torque $\tau_{hal\_kf}(t)$ basically consists of three parts: 1) an assistive torque to drive the knee joint $\tau_{ka}(t)$, 2) a viscous torque to provide damping effectiveness τ_{kd} and 3) a gravity compensating torque τ_{kg}.

1) *Assistive torque*: $\tau_{ka}(t)$ is represented as follows,

$$\tau_{ka}(t) = \frac{\tau_{kc}}{2} \operatorname{erf}(kt - T) \tag{3}$$

where $\operatorname{erf}(x)$ is defined as,

$$\operatorname{erf}(x) = \frac{2}{\sqrt{\pi}} \int_0^x e^{-s^2} ds \tag{4}$$

This function is a kind of sigmoid-shaped function (error function), τ_{kc} a constant torque, k a parameter to adjust the rise time to reach τ_{kc} and T a parameter to change the time to begin to generate the assistive torque $\tau_{ka}(t)$. Fig. 2 shows the torque pattern generated by (3). Basically, the motion support torque τ_{kc} is kept constant during knee flexion support. Error function (4) modifies the start of τ_{kc} to let the assistive torque act smoothly on the wearer's lower limb until the assistive torque reaches τ_{kc}.

2) *Viscous torque*: τ_{kd} is described as follows,

$$\tau_{kd} = -k_d \dot{\theta}_k \tag{5}$$

where k_d is a damping coefficient and $\dot{\theta}_k$ is the angular velocity of the knee joint. This torque is generated to prevent high velocity motion and thereby maintain safety.

3) *Gravity compensating torque*: τ_{kg} is calculated as

$$\tau_{kg} = m_k l_{gk} g \sin(\theta_k - \theta_h) \tag{6}$$

where, as shown in Fig. 3, m_k is the mass of the lower leg and foot of the HAL and the wearer, l_{gk} the length from knee joint to the center of mass of the lower leg and foot of the HAL and wearer, g gravity and θ_k and θ_h the knee and hip joint angle. By compensating gravity the feedforward torque can act uniformly on the wear's lower limb within the joint range of motion.

IV. EXPERIMENT

The following section presents the experiments conducted to validate the proposed control method by applying to a hemiplegic user. This section will introduce the characteristics of the trial subject and the experiment protocol.

The trial subject is a 60 year old male with right hemiplegia, resulting from a stroke. He has been diagnosed with Brunnstrom recovery stage IV. For locomotion in his daily

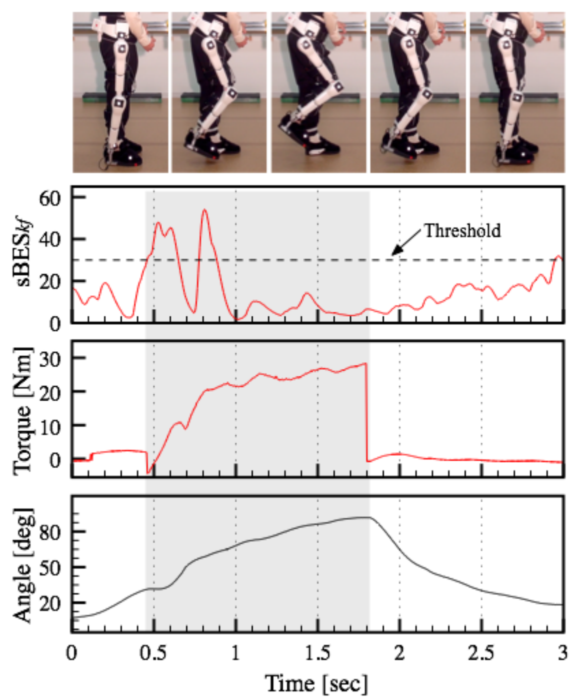

Fig. 4. Knee flexion support with the HAL during one cycle

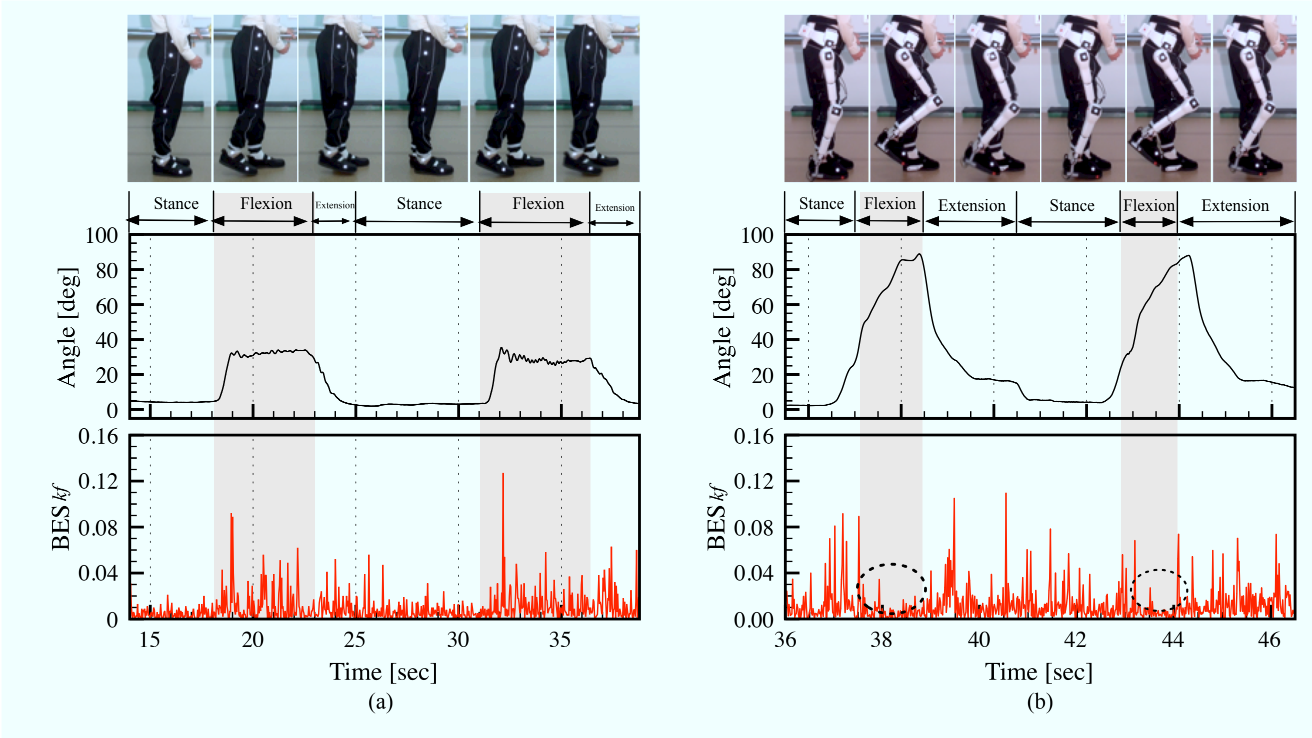

Fig. 5. Knee flexion during two cycles: (a) without the HAL, (b) with the HAL

life he wears an ankle orthosis, and walks using a cane. His walking is characterized by a circumduction gait, due to trouble to flex the right knee joint without flexing the right hip joint. We got informed consent before participating the experiment.

The trials were organized in one hour sessions, which were performed once a week, for four weeks. The experiment was performed in standing position. First, the subject stands evenly on both feet. After he shifts his weight completely to the left, the right knee flexion support is triggered by his intention, and consequently produced bioelectrical signals to bend his right knee. When the right knee joint is flexed to a preset angle the torque is released, he extends his right knee by himself and brings his weight by to his right leg. After that, he shifts his weight to the left again and motion support is performed once more.

In each trial, this step was repeated 7 times. Each time the subject has conducted a trial, some of the HAL's parameters were then adjusted to improve the knee flexion support. The parameters concerned are the threshold BES_{kf_th} which triggers the HAL action, and the amplitude of the assistive torque τ_{kc}. To ensure the subject's safety, additional support was provided to the subject. During each sessions, the subject held a handrail by his unaffected (left) hand.

In this experiment, the parameters k_d, m_k and l_{kg} in (5) and (6) are 10.0 Nm·s/rad, 4.73 kg, and 0.35 m respectively. The evaluation of the knee flexion support was made by comparing the data measured with and without using the HAL support. The comparison is based on the knee joint angle.

V. RESULTS

Fig. 4 shows data recorded during one cycle. The range of values of the smoothed bioelectrical signal change between trials depending on skin conditions and so on; the range of BES_{kf_th} in the figure is therefore arbitrary. The reference used for the knee joint angle is the value measured during the standing posture. Both the angles and torques are defined as positive during flexion. The parameters set during the experiments, τ_{kc}, k, and T in (3) in were 25 Nm, 15.0 and 0.02 sec respectively. It can be seen that when the bioelectrical signal of right knee flexor exceeds the threshold BES_{kf_th} which is 30, the HAL generates a positive assistive torque in the direction of the flexion and the subject's knee becomes flexed. After each parameter was fixed, the subject could repeat the knee flexion 7 times.

Fig. 5 shows the right knee joint angle and the bioelectrical signal of right knee flexor with and without assistive support, during two cycles out of a total of 7 times of knee flexion. The data shows that the right knee angle measured during flexion is larger when using the HAL. At the same time, the bioelectrical signal of right knee flexor during flexion while using the HAL (the encircled parts in Fig. 5 (a)) is smaller than that without the HAL (b).

VI. DISCUSSION

The HAL was used to assess the support provided to a person with hemiplegia. The motion support could be voluntarily provided by using a command signal based on the bioelectrical signal of the knee flexor. The range of the paralyzed knee flexion increased more than significantly. Furthermore, the use of the HAL allowed a decrease in the bioelectrical signal of the paralyzed knee flexor during knee flexion. This means that effort required for the paralyzed knee flexion decreased due to motion support from the HAL. As the result of experiments, we confirmed that the HAL could efficiently improved the knee flexion of the patient with hemiplegia.

Orthoses have been commonly used by persons with hemiplegia. Basically, standard orthoses assess paralyzed joints by through their stiffness property; it is impossible to move the joints voluntarily. However the HAL allows voluntarily support of the wearer's joints and actively compensates the motor function limitations caused by paralysis. Therefore the HAL appears to be more appropriate than orthoses as an assistive device for hemiplegia.

In the near future, the proposed method will be applied to the motion support of walking, standing up and sitting. Additionally, it would be a possible for a patient with hemiplegia to use this method as a rehabilitation tool. It has already been shown that voluntary activation is more effective than passive drive in the rehabilitation of paralyzed arms and wrists [8], [9]. We plan to conduct the rehabilitation of patients with paralyzed lowers limb by using this voluntary control method.

VII. CONCLUSIONS

In this research, a motion control method for the HAL robot suit was proposed in order to support the paralyzed leg of a hemiplegic wearer voluntarily. The bioelectrical signal detected on the skin surface around the muscle was used as the command signal, which is regarded as the wearer's intention to move, and a sigmoid-shaped torque pattern was adopted as an assistive torque to add the force to the wearer smoothly. To investigate the efficiency of the proposed method, the knee flexion support was assessed with one hemiplegic subject who has difficulties bending his right knee. As a result, the HAL allowed the subject to bend his knee voluntarily, which improved the knee flexion significantly, compared with the behavior without the HAL. This research suggests powered assistive devices may be more effective than orthoses for enabling natural motion of impaired limbs. The next step will be to explore the effectiveness of this proposed control method for a wide range of subjects showing different types of hemiplegia.

VIII. ACKNOWLEDGMENTS

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REFERENCES

[1] T. Barskova and G. Wilz, Interdependence of stroke survivors' recovery and their relatives' attitudes and health: A contribution to investigating the causal effects, in *Disability and Rehabilitation*,Vol. 29, No. 19, 2007, pp. 1481-1491.

[2] Blaya J.A. and Herr H., Adaptive control of a variable-impedance ankle-foot orthosis to assist drop-foot gait, in *IEEE Transtions on Neural Systems and Rehabilitation Engineering*, Vol. 12, No. 1, MARCH 2004, pp. 24-31.

[3] Stephen M Cain, Keith E Gordon, and Daniel P Ferris, Locomotor adaptation to a powered ankle-foot orthosis depends on control method, in *Journal of NeuroEngineering and Rehabilitation*, 2007, 4: 48.

[4] TSUKISHIRO Keiichi, A New Prosthetic Knee joint System C-Leg, in *Journal of the Japan Society of Mechanical Engineers*, 107(1033), 2004, pp.934-935.

[5] Junpei Okamura, Hiroshi Tanaka and Yoshiyuki Sankai, EMG-based Prototype Powered Assistive System for Walking Aid, in *Proc. of Asian Symposium on Industrial Automation and Robotics (ASIAR'99)*, Bangkok, 1999, pp. 229-234.

[6] K. Suzuki, G. Mito, H, Kawamoto, Y.Hasegawa, Y. Sankai, Intention-Based Walking Support for Paraplegia Patients with Robot Suit HAL, *Advanced Robotics*, Vol. 21, No. 12, 2007, pp. 1441-1469.

[7] Hiroaki Kawamoto and Yoshiyuki Sankai, Power Assist System HAL-3 for Gait Disorder Person, *Proc. of the 2002 International Conference on Computers Helping People with Special Needs (ICCHP 2002)*, 2002, pp.196-203

[8] L. Dipietro, M. Ferraro, JJ Palazzolo, HI Krebs,BT Volpe, and N. Hogan, Customized interactive robotic treatment for stroke: EMG-triggeredtherapy, *EEE Trans Neural Systems and Rehabilitation Engineering*, Vol. 13, No. 3, SEPTEMBER 2005, pp. 325-334.

[9] Lotze M, Braun, C, Birbaumer N, Anders S, Cohen LG, Motor learning elicited by voluntary drive, *Brain*, Vol. 126, No. 4, 2003, pp. 866-872.