

# Modification of Hemiplegic Compensatory Gait Pattern by Symmetry-based Motion Controller of HAL

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**Abstract**—As one of several characteristics of hemiplegic patients after stroke, compensatory gait caused by affected limb is often seen. The purpose of this research is to apply a symmetry-based controller of a wearable type lower limb robot, Hybrid Assistive Limb (HAL) to hemiplegic patients with compensatory gait, and to investigate improvement of gait symmetry. The controller is designed respectively for swing phase and support phase according to characteristics of hemiplegic gait pattern. The controller during swing phase stores the motion of the unaffected limb and then provides motion support on the affected limb during the subsequent swing using the stored pattern to realize symmetric gait based on spontaneous limb swing. Moreover, the controller during support phase provides motion to extend hip and knee joints to support wearer's body. Clinical tests were conducted in order to assess the modification of gait symmetry. Our case study involved participation of one chronic stroke patient who performs abnormally-compensatory gait for both of the affected and unaffected limbs. As a result, the patient's gait symmetry was improved by providing motion support during the swing phase on the affected side and motion constraint during the support phase on the unaffected side. The study showed promising basis for the effectiveness of the controller for the future clinical study.

## I. INTRODUCTION

Most patients with hemiplegia caused by stroke have a gait abnormality with asymmetrical movement pattern [1]. One of the main reasons is that the patients are unable to move hip, knee and ankle in isolation by remarkable increase of abnormal extension synergy in lower extremities which is frequently involved in the recovery process of stroke [2]. To avoid dragging of toe during swing phase caused by these motor impairments, in many cases, hemiplegic patients walk with asymmetrical compensatory gait including circumduction (swings out to the side in an arc), vaulting (raising the heel on unaffected side leg), and/or hip hiking (lifting the pelvis on the affected side) [3]. Moreover, the patients support their body with knee hyperextension (excessively extending the impaired knee joint) to prevent knee buckling during the stance phase [4].

Although evidences of negative effect of gait asymmetry have not been provided, Patterson et al. pointed out that long-term persisting gait asymmetry may lead to negative

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consequences which include: 1) decreased dynamic control balance, 2) musculoskeletal injury and articular joint degeneration to the non-paretic lower limb, 3) increased energy expenditure, 4) decreased bone mineral density in the femoral neck on the paretic side, 5) decreased overall activity levels [5]. Therefore, to reduce these consequences, it is important to patiently promote restoration of gait symmetry for the patients with persisting gait asymmetry.

Ankle foot orthosis (AFO) has been commonly used by persons with hemiplegia to support their paralyzed limb [6]. A standard AFO reduce weight bearing forces, stabilize the motion of the ankle joint during stance phase and prevent drop foot during swing phase using the material stiffness and elastic force generated in reaction to its deformation. Unfortunately these passive mechanisms do not allow to actively move paralyzed joints and therefore do not improve gait symmetry from the viewpoint of joint kinematics.

As advanced approaches of gait improvement, powered orthoses to provide physical assistance with actuated joints have been developed by a number of research groups. In particular, Blaya et al. developed Active Ankle Foot Orthosis (AAFO) with a series elastic actuator as a prototype model [7]. AAFO is able to adaptively control impedance of ankle joint according to gait behavior during swing phase and contact phase respectively. The device helps to avoid drop foot and foot slap of hemiplegic patients, and improve gait symmetry related to step length and step time. Sulzer et al. developed an active knee orthosis attaching series elastic actuator driven by a flexible cable transmitting mechanical force generated by a remote motor. The device demonstrated kinetically and kinematically selective support of knee flexion during swing for stroke patients with stiff-knee gait [8].

These approaches are meant as solutions to local problem of impaired joint functionality and only implicitly address improvement of kinematic symmetry in terms of joint kinematic motion, relying on the patient's potential recovery of joint coordination. In contrast, we consider an approach to improve kinematic symmetry explicitly using an exoskeletal robot which has leg joints that match those of the patient's legs. Actuators placed at the joints of the robot control joint motions to mimic normal walking patterns, and allow the patient's joints to move in synchronization with the robot's motion. The exoskeleton would make a compensatory gait pattern as close as possible to a symmetrical gait.

The purpose of this research is to apply a symmetry-based controller of the wearable robot HAL (Hybrid assistive limb) to an chronic stroke patient with asymmetric compensatory gait pattern and to investigate improvement of gait symmetry.

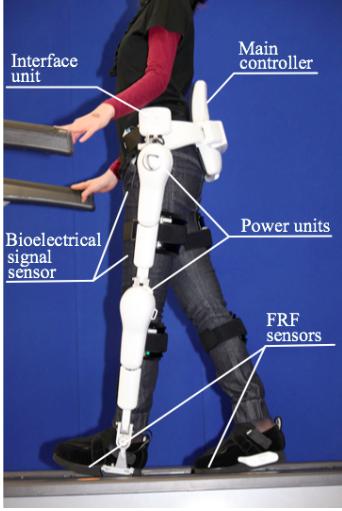


Fig. 1. Overview of the HAL system

In the symmetry-based controller, the joint angle sequence during the swing on the unaffected limb is recorded and used to generate the reference trajectory for the subsequent swing of the affected limb, so that the affected limb traces the pattern generated by the unaffected limb in the preceding step. Moreover, the controller during support phase provides motion to extend hip and knee joints to support wearer's body.

## II. METHODS

### A. HAL

The external appearance of HAL used in this research is showed in Fig. 1 [9]. The exoskeleton is designed to support the mechanical functions of the human lower body. It is an articulated structure, consisting of a frame, active joints at the hips and knees, passive joints at the ankles. It is attached to the user's hips and legs using cuffs and belts. The length of upper and lower thigh and hip size are mechanically adjustable according to wearers' body. The joints of the exoskeleton (hip, knee and ankle) have one DOF for each in the sagittal plane. The torques are generated by the power units at the hips and knees. Each power unit incorporates an actuator, a motor driver, a microprocessor and communication interface. The motion support is achieved by transmitting the torques from the power units to the user's legs through the exoskeleton frames.

The control of HAL system is integrated by its main controller. It controls the power units for assistance and safety, gathering sensor data, detecting failures and communicating the sensor data to local computers to understand the wearer's and HAL's condition.

HAL is equipped with a sensing system based on several types of sensors to detect the HAL's state. Potentiometers are mounted on each joint of the HAL and used to measure the joint angles. 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).

	phase 1	phase 2	phase 3	phase 4
Walking motion				
Left leg	swing	support	support	support
Right leg	support	support	swing	support

Fig. 2. Phase division in walking. The left leg is colored with gray.

### B. Symmetry-based Motion Controller

1) *Phase Division during Walking:* In this study walking motion can be divided into four phases (Fig. 2). The phase 1 is the swing phase of the left leg. In this phase the right leg supports the body, while the left foot leaves the ground surface and the left leg swings forward. In the phase 2 the left foot contacts the ground surface at the front of the body so that the wearer's weight is supported by both legs. The weight is then completely transferred to the left foot. The phase 3 is the swing phase of the right leg. The right foot leaves the ground surface and the right leg swings forward. The body weight is fully supported by the left leg. The phase 4 is the phase where the right foot contacts the ground surface at the front of the body and both legs support the body. The weight is then completely transferred to the left foot

2) *Motion Support:* Motion support in hemiplegic gate is performed at swing phase (Phase 3 as described above section) and support phase (Phase 1, 2 and 4) respectively. Figure 3 shows a control system to support hemiplegic leg motion on the right side (We suppose here the right side is the affected side). The active joints on the unaffected side (left) are controlled to compensate viscosities of the actuators to allow spontaneous motion of the wearer during the swing phase (phase 1) and to extend hip and knee joints to support the wearer's body during the support phase (phase 2-4).

a) *Swing phase:* In the phase 1, while the wearer swings the left leg by oneself, HAL detects the angles  $\boldsymbol{\theta}_l = [\theta_{lh}, \theta_{lk}]^T$  consisting of left hip joint angle  $\theta_{lh}$  and left knee joint angle  $\theta_{lk}$ , and calculates angular velocities  $\dot{\boldsymbol{\theta}}_l = [\dot{\theta}_{lh}, \dot{\theta}_{lk}]^T$  consisting of left hip joint angular velocity  $\dot{\theta}_{lh}$  and left knee joint angular velocity  $\dot{\theta}_{lk}$ . The  $\boldsymbol{\theta}_l$  and  $\dot{\boldsymbol{\theta}}_l$  are recorded in motion buffer connected to the switch  $SW_1$  and transformed into reference joint angles  $\boldsymbol{\theta}_{r,ref}$  and the reference angular velocities  $\dot{\boldsymbol{\theta}}_{r,ref}$ , which are used for motion support of right leg.

As the walking phase transits to the phase 3, the controller is connected to the terminal  $s_1$  of the motion buffer in the switch  $SW_2$ , and generates assistive torque on the right side  $\boldsymbol{\tau}_r = [\tau_{rh}, \tau_{rk}]$  consisting of right hip joint torque  $\tau_{rh}$  and left knee joint torque  $\tau_{rk}$ , which are calculated using PD controller based on the differences between motion reference ( $\boldsymbol{\theta}_{r,ref}, \dot{\boldsymbol{\theta}}_{r,ref}$ ) and the current right angles  $\boldsymbol{\theta}_r = [\theta_{rh}, \theta_{rk}]^T$  consisting of right hip joint angle  $\theta_{rh}$  and right knee joint angle  $\theta_{rk}$ , and right joint angular velocities  $\dot{\boldsymbol{\theta}}_r = [\dot{\theta}_{rh}, \dot{\theta}_{rk}]^T$  consisting of right hip joint angular velocity  $\dot{\theta}_{rh}$  and right

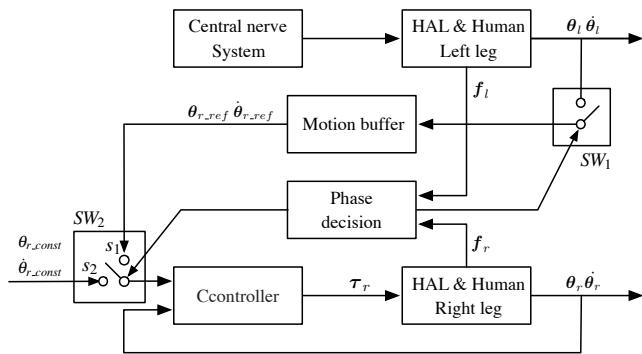


Fig. 3. Block diagram for the symmetry based-controller on the affected side of hemiplegia patients.

knee joint angular velocity  $\dot{\theta}_{rk}$ .

b) *Support phase*: As during Phase 1, 2 and 4, the controller is connected to the terminal  $s_2$  in the switch  $SW_2$ , and generate assistive torque on the basis of PD controller with the difference from the constant reference ( $\Theta_{r,const}$ ,  $\dot{\Theta}_{r,const}$ ) and the current right angles and angular velocities ( $\Theta_r$ ,  $\dot{\Theta}_r$ ). The motion support of stance phase is basically performed in the direction of the hip and knee extension.

c) *Phase transition*: Detection of transition between the phases is based on thresholding on the values of left FRF  $\mathbf{f}_l = [f_{lf}, f_{lr}]^T$  consisting of left front FRF  $f_{lf}$  and rear FRF  $f_{lr}$ , and right FRF  $\mathbf{f}_r = [f_{rf}, f_{rr}]^T$  consisting of right front FRF  $f_{rf}$  and rear FRF  $f_{rr}$ .

### *C. Clinical Evaluation*

**1) Participant:** We applied the developed control method to a 51 year old female patient with hemiplegia on the right side resulting from stroke. In her daily life, she walks using a cane. Her walking is characterized by a compensatory gait on both sides, which has difficulties to flex the knee joint of the affected side (right side) during swing phase and unnaturally flex the knee joint of the unaffected side (left side) during support phase. She gave an informed consent before participating.

2) *Intervention:* The clinical trial was conducted once a week for four weeks, and each trial was 60 minutes. For the former two weeks, we mainly explored the parameters related to the detection of phase transition and gains related to the amount of the assistive torque during the swing and support phases based on reports from the participant and observation by the experimenters. For the latter two weeks, She learned to walk being aware of the integrated motion between HAL and her limbs with the fixed parameters. Figure 4 shows the patient receiving the HAL's motion support.

3) *Data analysis:* The patient was fitted with retroreflective markers, which are placed on the points referring to a seven segment body model consisting of a trunk, right and left thighs, shanks, and feet. The markers data were collected using a 16-camera motion capture system (120 Hz; Vicon, Oxford, UK). From the marker data, right and left hip and knee joint angle ranges during swing phase (joint



Fig. 4. A hemiplegic patient walking with HAL during the experiment.

kinematic parameters), swing and stance phase duration (temporal parameters), and step lengths and heel height from floor during swing phase (spatial parameters) were calculated respectively. Hip joint angle range during walking with HAL was exceptionally obtained from HAL sensing system because the markers of hip segments covered by HAL structure were not adequately captured by the cameras.

To analyze changes of joint kinematic, temporal, and spatial symmetry between without and with wearing the HAL, a symmetry index (SI) was used as follows [10].

$$SI = \frac{V_{affected} - V_{unaffected}}{0.5(V_{affected} + V_{unaffected})} \times 100 \quad (1)$$

where  $V_{affected}$  is the value of a gait parameter (kinematic, temporal, and spatial parameter) recorded for the affected side, and  $V_{unaffected}$  is the corresponding value for the unaffected side. The symmetry index ( $SI$ ) normalizes the difference of values between both sides according to the average of the both sides as a reference value in the denominator. The magnitude of the index reflects the degree of asymmetry, and the sign indicates the direction of asymmetry. A value of 0% represents perfect symmetry for this index. A positive (negative) sign means a larger value of the gait parameter for the affected (unaffected) side.  $V_{affected}$  and  $V_{unaffected}$  of each gait parameter are obtained by averaging data from ten strides. The change of gait is evaluated by comparing  $SI$  values without and with HAL.

### III. RESULTS

Hip and knee joint angles during walking without wearing the HAL are shown in Fig. 5. The reference used for the joints angle is the value measured for the standing posture. The angles are positive in the flexion direction and negative in the extension direction. It is clear that the knee joint of the right (affected) side was not completely flexed during the swing phase (Fig. 5 (a)) and the knee joint of the left (unaffected) side was unnaturally flexed during the support phase (Fig. 5 (b)). The maximum extension positions of both hip joints at later support phase were around 0 rad (Fig.5 (c)).

Figure 6 illustrates the joint angles and command torques of hip and knee joints in both sides during walking wearing

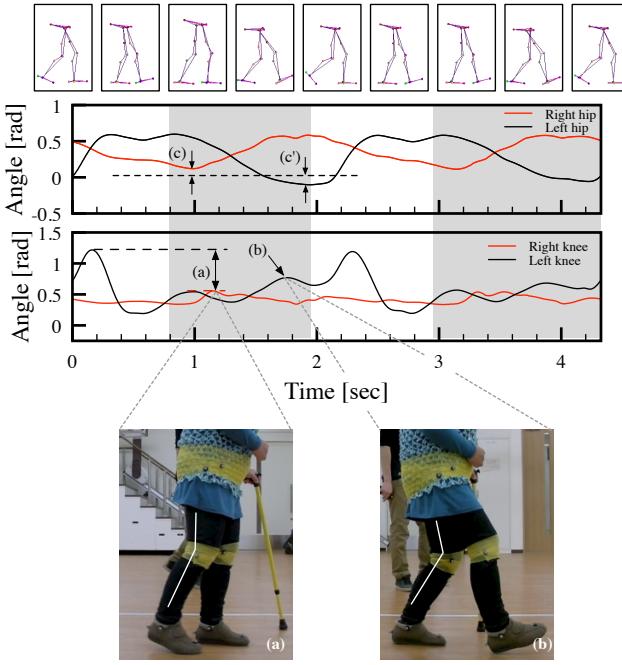


Fig. 5. Hip and knee joint angles of both legs for two gait cycles without HAL. The swing phase of right (affected) side is colored with gray. In (a), the knee joint on the affected (right) side is not flexed during the swing phase. In (b), the knee on the unaffected (left) side is abnormally flexed during the stance phase. In (c) and (c'), the hip joint extension on both sides is insufficient during the stance phase.

the HAL. The hip and knee joint angle trajectories recorded during the swing phase in the left side (Fig. 6 (a), (a') dashed line), are used as target trajectories in the joint angle of the right side (Fig. 6 (b), (b') dashed line). The command torque is positive in the flexion direction and negative in the extension direction. It was observed that the HAL generates the command torque during the swing phase of the right side based on the target trajectories, and the joint angles of the right side were provided in close accordance with the target trajectories (Fig. 6 (c), (c')). The maximum extension positions of both hip joints at later support phase were achieved more than 0 rad (Fig. 6 (d), (d')). Moreover, the HAL generates the command torque in the direction of the knee extension during the support phase of the left side, and the knee joint angle obtained the smooth trajectory (Fig. 6 (e)).

Figure 7 shows the *SI*s in the joint kinematic, the temporal, and spatial parameters respectively. In the joint kinematic parameters (Fig. 7 (a)), the *SI*s of the range of hip and knee joint during the swing phase without wearing HAL demonstrated negative values which means that the range of hip and knee joints on the affected side were smaller than those on the unaffected side. In contrast, in the case with HAL, the *SI* value on the hip joint was close to 0%, and the values on the knee joint reached a positive value where the range of knee joint on the affected side was larger than that on the unaffected side. In the temporal parameters (Fig. 7 (b)), the *SI*s of the swing and the stance duration without HAL had positive values that these durations on the

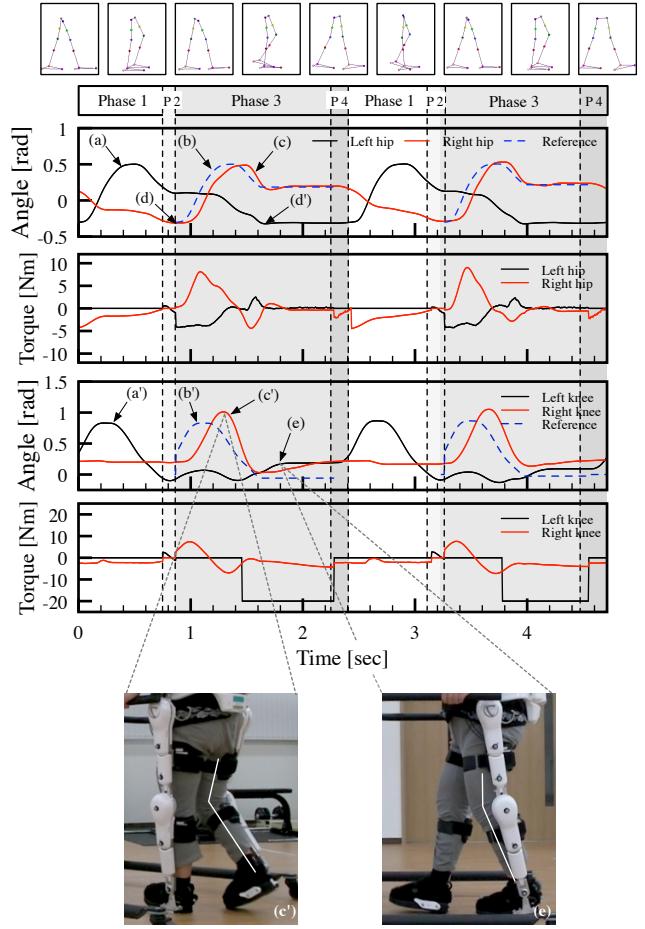


Fig. 6. Hip and knee joint angles and command torques of both legs for two gait cycles with HAL. In (a) and (a'), the trajectories on the joint angles during swing phase of the left (unaffected) side. In (b) and (b') the target trajectories on the joint angles during swing phase of the right (affected) side. In (c) and (c'), the actual trajectories on the joint angles follow the target trajectories. In (d) and (d'), the hip joint extension on both sides is extended during the stance phase. In (e), the knee joint angle on the left (unaffected) side generates smooth trajectory.

affected side were longer than that on the unaffected side. In contrast, these *SI*s with HAL approached 0% respectively. In the spatial parameters (Fig. 7 (c)), the *SI* of the step length had a positive value and that of the maximum heel height had a negative value. In contrast, these *SI*s with HAL was closed to 0%.

#### IV. DISCUSSION

The clinical study was conducted with a hemiplegic patient who had compensatory gait pattern in order to investigate improvement of gait symmetry by the HAL's assist controller. The remarkable compensatory knee joint pattern reducing flexion on affected side during the swing phase (Fig. 5 (a)) and unnecessarily increasing flexion on unaffected side during the support phase (Fig. 5 (b)) are respectively modified to improve the flexion (Fig. 6 (c')) and prevent the abnormal flexion (Fig. 6 (e)). Furthermore, we showed that the *SI* values for the joint kinematic, temporal and spatial parameters were changed to be close to the symmetrical gait

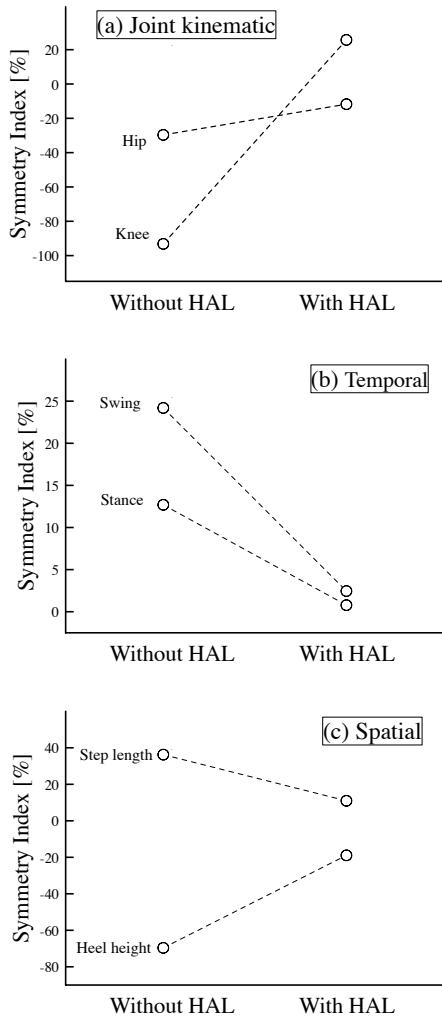


Fig. 7. Differences in the symmetry index of the kinematic parameters (a), the temporal parameters (b), and the spatial parameters (c) with and without wearing the HAL

by HAL's motion (Fig. 7). Therefore we confirmed that the HAL's assist controller could improve walking symmetry for the patient with compensatory gait. As a next step we should confirm the improvement of walking symmetry by the HAL in the clinical study with patients who have several type of compensatory gait pattern.

The symmetry-based controller is expected to contribute to functional recovery of compensatory gait for hemiplegic patients. It is based on the observation that the control system repeatedly provided the symmetrical motion as locomotor treatment. In case of the patient in this clinical study, the controller performed the motion support for the knee joint which she has difficulty to conduct to flex during the swing phase on the affected side. Moreover, the controller worked for preventing the knee flexion which occurred as the compensatory phenomenon during the support phase on the unaffected side. The patient remarked that she understood to force to flex the knee joint by receiving the HAL's torque to make the HAL to extend the knee joint. It would mean

that HAL exhibits motion (force sensation) to the patient to prevent abnormal motion. Therefore, patients would be able to learn symmetrical gait using the control system as motion support as well as force sensation. In the near future, we will extend the clinical studies to examine gait functional recovery.

## V. CONCLUSIONS

We applied a symmetry-based controller of HAL to modify compensatory gait pattern of hemiplegia patients. The controller was designed to realize symmetric gait by providing joint motion on the lower limbs based on the spontaneous joint motion of the unaffected limb and also individual characteristics of compensation. To investigate the modification capability of the controller, clinical trials with a hemiplegic subject was conducted. As a result, the controller provided the subject with the support for the hip and knee flexion motion during the swing phase on the affected side and motion constraint for excessive knee flexion during the support phase on the unaffected side, which lead to significant improvement of gait symmetry in terms of joint kinematic, temporal, and spatial symmetry. In the near future, we will apply this control system to subjects showing different types of compensatory gait pattern.

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