Functional Electrical Stimulation Based on a Pelvis Support Robot for Gait Rehabilitation of Hemiplegic Patients after Stroke

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***Abstract*—More and more stroke survivors are suffering from physical motor impairments. Current therapeutic interventions have various limits to the efficient recovery of normal motor function of the lower limbs. Therefore, we propose a novel gait rehabilitation system for hemiplegic patients after stroke. It integrates functional electrical stimulation (FES) with a pelvis-supporting robotic system. A corresponding relationship between the gait phase and the active lateral movement of the pelvis is first constructed from experiments on simulated hemiplegic patients. By estimating the gait phase from the lateral motion of the pelvis based on this relationship, the timing of FES sent to the muscles of the lower limbs can be automatically determined during a gait cycle. After experiments on simulated hemiplegic stroke survivors with the FES control algorithm, the proposed algorithm and the gait rehabilitation system are verified to be feasible and promising.**

I. Introduction

In recent years, as aging societies have become more common, many stroke survivors are suffering considerably from a loss of physical mobility. To lead an independent life, physical health is fundamentally necessary. Thus, the recovery of lost motor function is important for those stroke survivors. Typical physiotherapy shows only limited success in motor function restoration [1], and many current approaches require the physical therapist's observation, specifically designed preparatory exercises and the direct manipulation of position movement of the lower limbs. Although research has made great strides in the field of gait recovery, more work still needs to be done to highly enhance the effectiveness of gait rehabilitation for stroke survivors. Therefore, novel therapy methods and developments are increasingly necessary. Some biomechanical solutions, however, have been shown to be highly effective. These include functional electrical stimulation (FES), which promotes motor function recovery through periodically activating paralyzed muscles. It is mostly applied to the dorsiflexor muscles for foot drop [2]. Walking ability recovery has also been shown to improve by stimulating the quadriceps muscles while patients' legs swing forward to the next step [3]. High neural plasticity and repair mechanisms for restoring motor functions can be obtained by using FES, and the duration of effectiveness of FES can be at least 24 months [4].

Additionally, a number of robotic approaches have been developed to assist movability recovery of the lower limbs based on body weight support (BWS). Conventionally, commercial BWS systems relieve body weight using an elastic spring [5], a winch [6] or a counterweight [7] with a wire harness from above. The support force of these systems commonly varies in an undesirable manner owing to the flexible nature of the spring. In addition, once it is set before training, it cannot be properly adjusted during the gait cycle. To solve this problem, researchers have developed several motor-actuated devices. A partial BWS system with a designed wire harness was developed by Glauser *et al*. [8]. It can monitor the support force and adjusts it in real time to maintain it close to a target support force during a gait cycle.

These systems, however, mainly provide movement support only in the vertical direction but restrict rotation, obliquity or horizontal translation of the pelvis [9]. Recently, it has been demonstrated that patient-specific torso movements, including pelvic movements, may play an important role in generating desired gait patterns and normal locomotion [10]. Gait motion consists of six factors, including pelvic rotation, pelvic tilt, knee flexion in the stance phase, knee mechanisms, foot mechanisms and lateral displacement of pelvis [11]. Three of these factors are related to pelvic motion, which emphasizes the importance of support to pelvic motion during gait rehabilitation. Thus, we previously developed a pelvic support robot (shown in Fig. 1) with BWS functionality. It can provide active support for vertical and lateral motions of the pelvis without a harness.

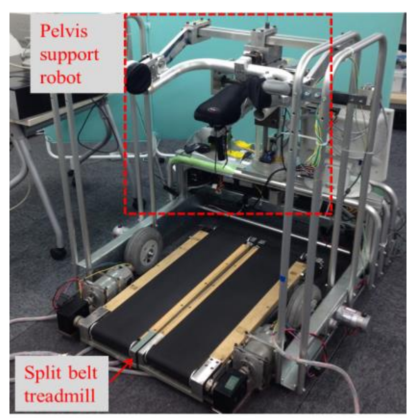

Figure 1. Pelvis support robot

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In this paper, we integrate FES with the pelvis support robot to effectively improve the quality of gait rehabilitation. We investigate FES control of the pelvic support robot in real time based on gait phase estimation.

II. Gait Phase Estimation Based on Pelvis Movement

To construct a FES control algorithm based on the pelvis support robot, it is first necessary to establish the gait cycle estimation algorithm. The proposed approach mainly involves detecting lateral movement of the pelvis with the pelvic support mechanism (shown as Fig. 2). Thus, a corresponding relationship between lateral translation movement of the pelvis and the gait cycle must be obtained in advance.

A. Pelvis Support Robot

The pelvis support robot comprises a robot base and a pelvic support mechanism. It provides active support for vertical and lateral motions of the pelvis by using different motors for each movement type. The pelvic support mechanism is mounted on a pedestal, which is attached to a motor-actuated device. A DC motor (Maxon, EC 45, Brushless DC, Voltage rating: 24 V, Rated power: 250 W) rotates a vertical ball screw via reduction gears and drives the pedestal to upward or downward. Another DC motor (Maxon, EC 40, Brushless DC, Voltage rating: 24 V, Rated power: 120 W), assembled with the pelvic support mechanism, is used to driving the lateral swing motion of the pelvis. Two load cells (UNIPULSE, RSCM, Rated capacity: 1,000 N) are used to detect the vertical unloading force and the lateral motion force of the pelvis. The pelvis can be fitted into the mechanism and held in the left and right parts of the anterior superior iliac spine and the ischial bone. The fit can be adjusted according to individual differences in pelvis size. The mechanism has three passive rotational degrees of freedom for natural walking motions of the pelvis.

B. Estimation of Gait Phase Based on Lateral Pelvis Movement

Since the lateral translation of the pelvis and the gait cycle are both continuous motions during gait, the gait phase can be estimated through the relationship between them. It can be obtained by comparing the variation of lateral motion of the pelvis and gait cycle in time series. The lateral position of the pelvis is calculated in time by the robot system controller by monitoring the motor driving lateral movement. The lateral movement motor does not actively drive the pelvis support mechanism to move, but passively follows the pelvic movement through feedback from a load cell mounted on it.

Gait phase estimation was established based on an experiment. A healthy young male, 68 kg, 27 years old, who was recruited following informed consent wore an ankle-foot orthosis (AFO) on the right lower limb to simulate being a hemiplegic patient. While the participant walked on a split-belt treadmill (shown in Fig. 1), heel contact was measured from separated floor reaction force plates (AMTI, OR6-7 2000, cutoff frequency for the low-pass filter: 10.5 Hz) under the treadmill. The separated floor reaction force meter measured floor reaction forces on the right and left sides. As the right side was simulated to be hemiplegic, data from the right force plate were mainly used to analyze the results. Because the main purpose of this research was not body weight support, the vertical target unloading force was set to 50 N. In consideration of the achievable walking speed of hemiplegic patients after stroke, the treadmill belt speed was set to 1.0 km/h and the gait cadence was set to 44 strides per minute by a metronome. The measurement of lateral pelvis translation and gait cycle was started simultaneously and at the same recording frequency. Heel contact during walking was determined from the floor reaction force with a threshold of 10 N. Experiments consisted of 15 trials, each for 1 minute.

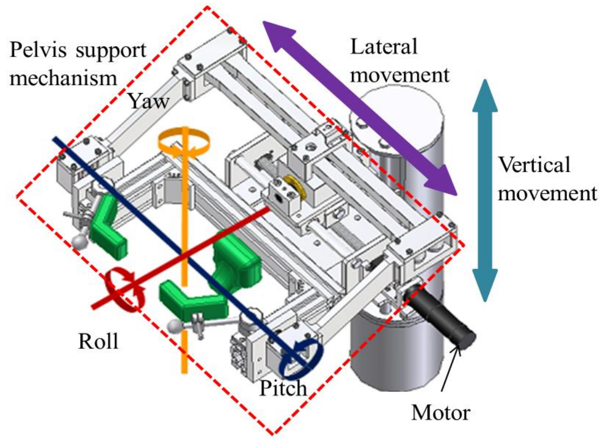


Figure 2. Motions of pelvis support robot

Fig. 3 shows a typical result from the measurement data. The results show that the lateral movement of the pelvis was similar to a periodical sine curve and had a local minimum and a local maximum in each stride. The beginning of a swing phase always slightly preceded the time of the local minimum lateral position. From statistical analysis of the measured data, the local minimum (shown as *P* in Fig. 3) was found to occur approximately $t = 0.035$ s on average after the toe-off time of the right leg. The average swing phase time was 0.65 s, and the average stride time was 1.35 s. Because hemiplegic patients are not fully able to swing the affected leg forward using the muscles of the affected lower limb, they have to laterally swing their pelvises to fully lift the affected leg and then move ahead. This means that the patient laterally swings their pelvis the most at the moment of toe-off to lift the affected leg. The value of *t* is therefore small in the experiment.

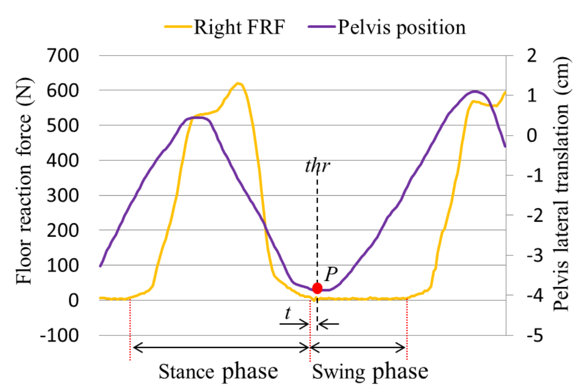


Figure 3. Relationship between gait phase and pelvis movement

III. FES CONTROL ALGORITHM BASED ON GAIT PHASE ESTIMATION

Since different muscles of the lower limb are activated at different times during walking, the timing of FES to these muscles should be considered. The FES device in this study includes two pairs of independent and non-invasive electrodes attached to the skin surfaces of the Tibialis Anterior (TA) and the quadriceps. The strategies used to trigger the electrical stimuli are based on the gait state.

A. FES Sent to Quadriceps and TA

The TA is the main muscle for foot dorsiflexion and the quadriceps is crucial for walking or running, and for assisting normal walking, the quadriceps and TA should be activated at the beginning of the swing phase to support the leg swing [12]. That is, once the gait phase of the affected lower limb is determined as being in the swing phase, command signals to trigger the electrical stimuli will be sent to the FES device from the controller of the pelvis support robot. The electrical stimulus should be sent to the quadriceps and TA of the affected lower limb for a period of time. The ideal stimulation duration lasts the whole length of the swing phase.

Compared with the total swing phase time, $t = 0.035$ s can be considered negligible in terms of the control of FES. Thus, the local minimum of pelvis lateral position can be approximated as occurring at the toe-off of the affected leg. As the quadriceps and TA should be activated during the swing phase, the FES should be triggered at the moment when the local minimum lateral position occurs. Its threshold can be set as *thr* (shown in Fig. 3).

The timing of the stimulation duration is shown in Table I.

TABLE I. ACTIVATION PROPORTION OF THE TWO LOWER LIMB MUSCLES DURING A GAIT CYCLE

Gait phase		Stance phase	Swing phase
Percentage in a gait stride		62%	38%
Muscle stimulation duration	**TA**	OFF	ON
	Quadriceps	OFF	ON

B. Experiment on FES Control Algorithm

Two able-bodied participants were recruited and simulated hemiplegic gait by wearing the AFO. Informed consent was obtained from them before the experiments began. Their personal information is shown in Table II.

TABLE II. PARTICIPANT INFORMATION

Subject	**Personal Information**			
	Gender	***Weight (Kg)***	***Simulated hemiplegia side***	***Age***
No.1	Male	60	right	29
No.2	Male	71	right	28

As shown in Fig. 4, participants wore the AFO in the pelvis support robot with two pairs of separate electrodes attached to the TA and quadriceps. The lateral movement motor was set to passively follow the pelvis movement to detect the gait phase. Participants' right legs were simulated as the affected hemiplegic side. The timing of the electrical stimulation was controlled by the FES control algorithm mentioned above. The FES device (STG4002, Multi-Channel Systems, MCS, GmbH, with a stimulation range from –8 to +8 V) was connected to the controller of the pelvis support robot. The other side was connected to the lower limb muscles with two pairs of separate non-invasive bipolar electrodes. Since muscles function differently during a gait cycle, the amplitude of the electrical stimuli to different muscles should be carefully controlled. In this experiment, the stimulus from the FES device was set at +7 V with a pulse width of 400 ms for TA stimulation and 200 ms for quadriceps stimulation. The waveform was a rectangular pulse. Electrode positions were decided by trial and error until the best possible response to the stimulation was found. Every participant was tested ten times, for 1 minute each time, at three different belt speeds (0.5, 1.0, 1.5 km/h). The participants followed a walking cadence of 44 strides per minute in all experiments.

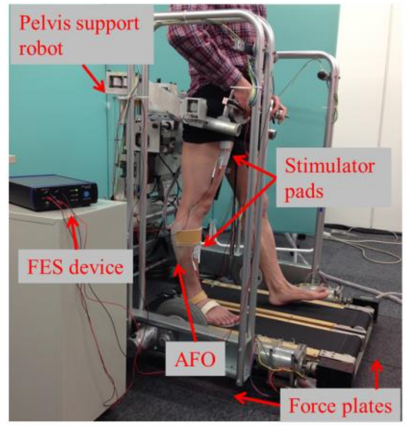
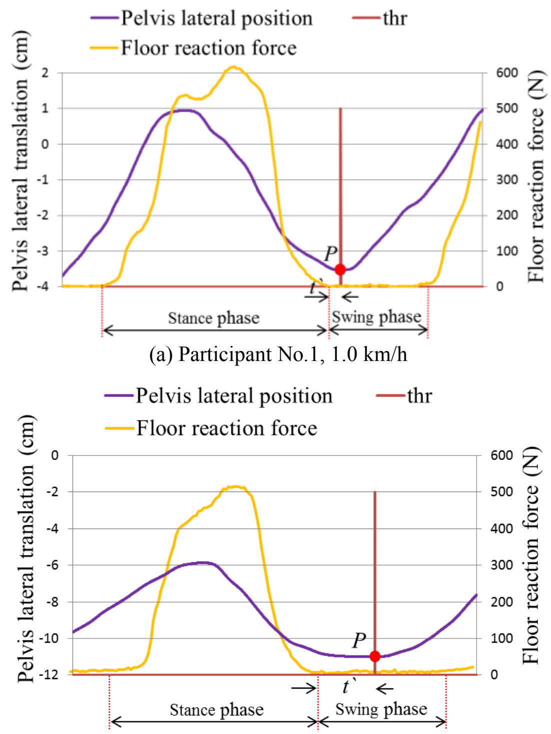


Figure 4. Experimental setup

C. Results and Discussion

Fig. 5 (a, b) shows typical stimulation results of the two participants walking at 1.0 km/h belt speed. As the participants walked within the gait training system, the lateral translation of the pelvis shows some periodicity. The swing phase of the affected side could be estimated by our proposed estimation algorithm. At the instant that the estimated gait phase was determined to be the stance phase, a command signal to trigger the stimuli was sent from the robot controller to the FES device. In other words, at the red point *P*, when the lateral translation position of the pelvis reached a local minimum, the electrical stimulus was sent to the quadriceps and TA of the affected side.

To compare the error of stimulation timing using the applied FES algorithm with the previous error *t*, *t'* is set up here in the experiments(as shown in Fig. 5). Fig. 6 shows the mean and standard deviation of *t'*, for the two participants at three speeds. The two subjects' lateral translation ranges of pelvis with the same belt speed are almost the same. The mean values of *t'* at 0.5 and 1.0 km/h belt speed are almost the same, and almost equal to the previous *t* value preliminarily set above. The *t'* at 1.5 km/h is, however, increased. The results also show that, while walking at a higher belt speed, the lateral translation range was smaller than that while walking at a



(a) Participant No.1, 1.0 km/h

(b) Participant No.2, 1.0 km/h

Figure 5. Experiment results. *t'* is the error of stimulation timing in current experiments

lower one. This is because the lateral pelvis movement became unstable during higher belt speed walking.

The experiment was conducted with the constant cadence of steps. Therefore, if the belt speed is higher, the stride width was larger so that participants could follow the cadence. This phenomenon may lead to the pelvis swinging in a more narrow range to avoid falling down, which might produce difficulties in gait training hemiplegic patients. As belt speed increases, the time of the local minimum of the lateral pelvis position occurs later. That is, the beginning of the swing phase estimated by the proposed algorithm is considerably later than that measured by the reaction force plates, a result of the larger inertia of the lateral swing of the pelvis during walking.

Generally, we conclude that the proposed algorithm is applicable for the control of FES without needing to tune parameters at a 0.5 or 1.0 km/h walking speed. However, the results reveal that additional attention should also be paid during the walking procedure. Steps should avoid large gait fluctuations as much as possible. Besides, there is another limitation of sufficient subjects participated in the experiments.

IV. Conclusion and Future Work

This study describes a novel system integrating FES with a pelvis support robot to train the paralyzed lower limb muscles of hemiplegic patients. We used experiments on simulated patients to preliminarily test and verify the feasibility of the proposed FES algorithm based on gait phase estimation. Nevertheless, we intend to improve the system by delivering more accurate stimulation during walking and training.

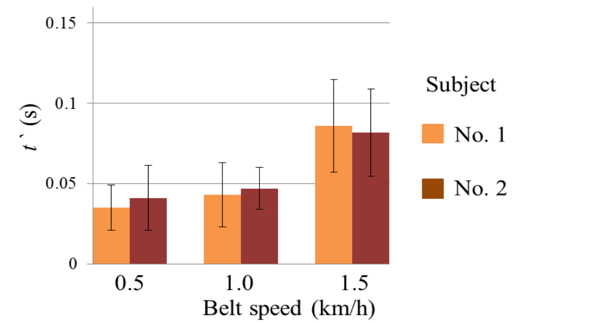


Figure 6. Comparison of mean value of *t'* and standard deviation, for two participants at three speeds

To work towards restoring as close a normal gait pattern as possible by the end of the rehabilitation program, we will continue to investigate accurate FES timing and the amplitude control of electrical stimulation in the future. To fully verify the proposed system, more subjects needs to be recruited and discussed in future. Finally, experiments on hemiplegic patients will be conducted to verify the effectiveness of the proposed method for gait rehabilitation.

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