

PAPER

Experimental Tests of a Prototype of IMU-Based Closed-Loop Fuzzy Control System for Mobile FES Cycling with Pedaling Wheelchair

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SUMMARY Rehabilitation training with pedaling wheelchair in combination with functional electrical stimulation (FES) can be effective for decreasing the risk of falling significantly. Automatic adjustment of cycling speed and making a turn without standstill has been desired for practical applications of the training with mobile FES cycling. This study aimed at developing closed-loop control system of cycling speed with the pedaling wheelchair. Considering clinical practical use with no requirement of extensive modifications of the wheelchair, measurement method of cycling speed with inertial motion measurement units (IMUs) was introduced, and fuzzy controller for adjusting stimulation intensity to regulate cycling speed was designed. The developed prototype of closed-loop FES control system achieved appropriately cycling speed for the different target speeds in most of control trials with neurologically intact subjects. In addition, all the control trials of low speed cycling including U-turn achieved maintaining the target speed without standstill. Cycling distance and cycling time increased with the closed-loop control of low cycling speed compensating decreasing of cycling speed caused by muscle fatigue. From these results, the developed closed-loop fuzzy FES control system was suggested to work reliably in mobile FES cycling.

key words: FES, cycling, pedaling wheelchair, closed-loop control, fuzzy

1. Introduction

Rehabilitation training of lower limb of hemiplegic subject with the pedaling wheelchair [1] was suggested to be effective through measurement of muscle activity by electromyography (EMG) during cycling [2]. Since rehabilitation training of lower limbs such as sit-to-stand and walking has the risk of falling for paraplegic subjects and patients who have severe motor paralysis, training with the pedaling wheelchair in combination with functional electrical stimulation (FES) can be useful decreasing the risk significantly.

For the purpose of motor rehabilitation, bicycle ergometer combined with FES has been studied [3]–[7]. In our previous studies, FES cycling system with the pedaling wheelchair was developed [8], [9], which can be useful both for motor rehabilitation and mobile cycling for motor disabled subjects. The system applied electrical stimulation pulses with the maximum stimulation intensity in order to

provide enough high speed cycling, because propelling the wheelchair with low cycling speed was difficult at the beginning and at making a turn, in which the wheelchair stopped right away. In addition, FES cycling under low cadence was suggested to be effective in rehabilitation training [10], [11]. Therefore, automatic adjustment of cycling speed is necessary for practical use of mobile FES cycling.

FES control for mobile cycling depends on cycling system [9], [12]–[16], in which most of the cycling system used open-loop control. Closed-loop control of cycling speed in FES cycling was tested with an enhanced experimental cycling ergometer by using fuzzy and PD controllers [12] or with a system assisted by an electric motor [14]. However, these studies were performed under the immobile cycling conditions using an orthotic brace which limits the ankle motion. In addition, the closed-loop control using only FES [12] did not adjust stimulus intensity at each time, but fixed it for each cycling period (one rotation of the crank). Therefore, a next step of FES cycling is a continuous time closed-loop control for mobile cycling without the orthotic brace, because mobile cycling is effective for rehabilitation of paralyzed lower limbs, going somewhere, recreation and so on.

The purpose of this study was to develop a prototype of closed-loop FES control system of cycling speed with the pedaling wheelchair and to test control performance of the controller in mobile FES cycling. First, a continuous time closed-loop fuzzy FES controller for cycling speed control was designed based on the fuzzy FES controller tested in knee extension angle control [17]. Since the controlled variable was different from the previous controller, control performance of the developed fuzzy controller had to be examined. Then, a closed-loop control system was developed considering clinical practical use with no requirement of extensive modifications of the wheelchair. That is, measurement method of cycling speed with inertial motion measurement unit (IMU), which could be easily attached and detached, was introduced. The developed control system was tested in cycling speed control with neurologically intact subjects. In addition, cycling including turn under the low cycling speed, and repeated cycling control until muscle fatigue occurred were tested in order to show control performance of the developed closed-loop control system and discussed issues of the developed system.

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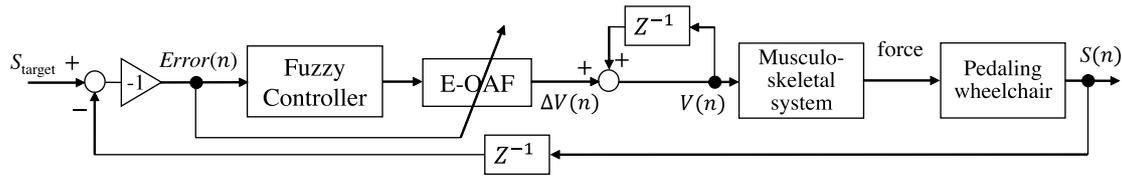


Fig. 1 Closed-loop fuzzy control system of cycling speed for mobile FES cycling. $S(n)$ and $V(n)$ show cycling speed and stimulation intensity at each time n , respectively.

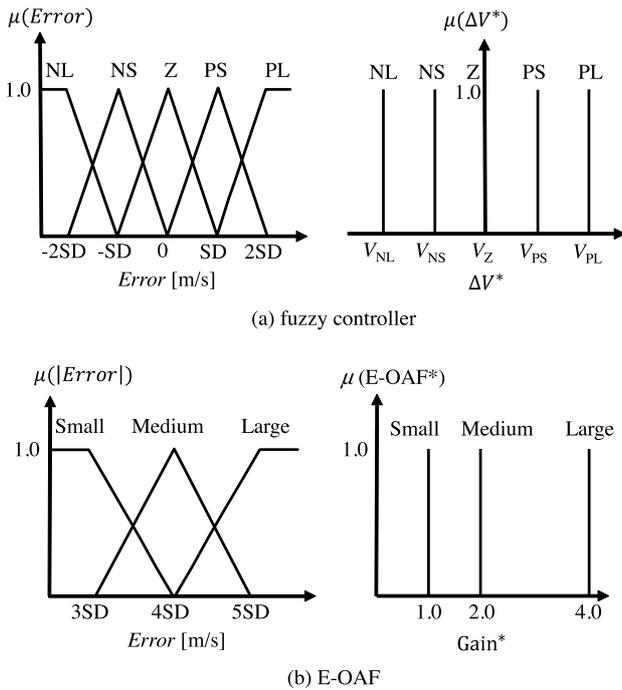


Fig. 2 Input and output membership functions of fuzzy controller and E-OAF for cycling speed control of FES cycling. The fuzzy linguistic terms of the fuzzy controller are NL (negative large), NS (negative small), Z (zero), PS (positive small) and PL (positive large), and those of the E-OAF are small, medium and large.

2. IMU-Based Closed-Loop FES Control System of Cycling Speed of Pedaling Wheelchair

2.1 Closed-Loop Fuzzy FES Controller

The block diagram of the closed-loop fuzzy FES control system for cycling speed control is shown in Fig. 1. Input of the fuzzy controller is error of cycling speed at each time and its output is change of stimulation pulse amplitude at the time. Output of the fuzzy controller is adjusted by error-based output adjustment factor (E-OAF). Gain of the E-OAF is determined by the cycling speed error, which increases the output value of the controller if the error is large [18]. As shown in Fig. 1, electrical stimulations applied to muscles produce the force that propels the pedaling wheelchair, and cycling speed is obtained as the controlled variable.

Input and output fuzzy sets of the fuzzy controller and the E-OAF are shown in Fig. 2. Input membership functions

Table 1 Fuzzy rule sets of the fuzzy controller and the E-OAF.

(a) fuzzy controller					
Error	NL	NS	Z	PS	PL
ΔV	PL	PS	Z	NS	NL

(b) E-OAF			
Error	Small	Medium	Large
Gain	Small	Medium	Large

were expressed by triangular and trapezoidal functions, and the output variable ΔV^* and Gain* were expressed by fuzzy singleton. Fuzzy rules are summarized in Table 1. The rules were configured based on that stimulation intensity is increased if error is negative, and decreased if error is positive, in which error is defined by

$$Error = (measured\ speed) - (target\ speed) \quad (1)$$

Parameter values of input membership functions were determined from standard deviation (SD) of cycling speed during voluntary cycling by 7 healthy subjects. Those of output membership functions were determined for each subject before control test.

The fuzzy inference was accomplished by using the Mamdani method and the defuzzification was processed by calculating center of gravity (COG). For example, the fuzzy inference output ΔV^* was converted into a crisp value ΔV by the following:

$$\Delta V = \frac{\sum \mu(\Delta V_k^*) \Delta V_k^*}{\sum \mu(\Delta V_k^*)} \quad (2)$$

where $k = 1, 2, \dots, K$. K is the number of the fuzzy linguistic term of ΔV^* . $\mu(\Delta V_k^*)$ is membership value of ΔV_k^* .

2.2 Measurement of Cycling Speed

2.2.1 Method of Cycling Speed Measurement

FES cycling system and attachment of IMUs are shown in Fig. 3. Although it is possible to measure cycling speed with a gyroscope attached to the center of the crankshaft (Sensor1 in Fig. 3 (c)), measurement error in instantaneous cycling speed is sometimes caused due to rotational play at the crank set. Therefore, in this study, cycling speed was measured with 2 IMUs shown in Fig. 3 using Kalman filter.

An IMU is attached on the frame that is horizontal part of the pedaling wheelchair, in which one axis of the IMU is in the traveling direction. Although integral of the acceleration signal in the traveling direction can provide cycling

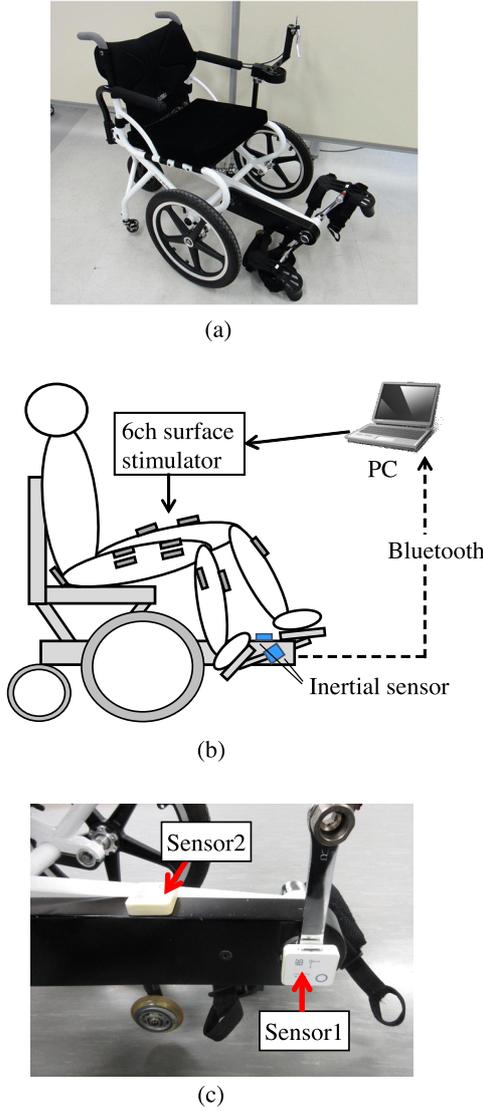


Fig. 3 Pedaling wheelchair used in this study (a), experimental setup of FES cycling control tests (b) and attachment of IMUs on the wheelchair (c).

speed, large integral error is caused. The integral error is compensated by Kalman filter using angular velocity measured with an IMU attached on the center of the crankshaft. That is, the Kalman filter estimates the error of the cycling speed measured with accelerometer $\Delta\hat{S}$ from difference Δy between the speed calculated from acceleration signal S_a and that measured with gyroscope S_g . Finally, cycling speed \hat{S} is estimated by subtracting $\Delta\hat{S}$ from S_a .

The state equation of the system is represented by the error of the cycling speed, ΔS , between the speed calculated from acceleration signal and that obtained from measured angular velocity, and bias offset of outputs of accelerometer Δb as follows:

$$\begin{bmatrix} \Delta S_k \\ \Delta b_k \end{bmatrix} = \begin{bmatrix} 1 & \Delta t \\ 0 & 1 \end{bmatrix} \begin{bmatrix} \Delta S_{k-1} \\ \Delta b_{k-1} \end{bmatrix} + \begin{bmatrix} \Delta t \\ 1 \end{bmatrix} \phi \quad (3)$$

where Δt is the sampling period, ϕ is error in acceleration measurement. Observation signal of the system is the defer-

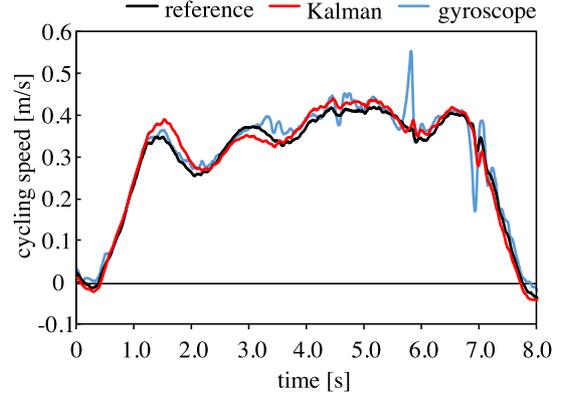


Fig. 4 Examples of estimated and measured cycling speeds under the usual pedaling without large backlash effect.

ence of two velocities, which is given by:

$$\Delta y_k = \begin{bmatrix} 1 & 0 \end{bmatrix} \begin{bmatrix} \Delta S_k \\ \Delta b_k \end{bmatrix} + v \quad (4)$$

where v is error in angular velocity measurement. On this state-space model, the Kalman filter repeats predictions (Eq. (5)) and corrections (Eq. (6)):

$$\begin{bmatrix} \Delta\hat{S}_k^- \\ \Delta\hat{b}_k^- \end{bmatrix} = \begin{bmatrix} 1 & \Delta t \\ 0 & 1 \end{bmatrix} \begin{bmatrix} \Delta\hat{S}_{k-1} \\ \Delta\hat{b}_{k-1} \end{bmatrix} \quad (5)$$

$$\begin{bmatrix} \Delta\hat{S}_k \\ \Delta\hat{b}_k \end{bmatrix} = \begin{bmatrix} \Delta\hat{S}_k^- \\ \Delta\hat{b}_k^- \end{bmatrix} + \begin{bmatrix} K_1 \\ K_2 \end{bmatrix} (\Delta y_k - \Delta\hat{S}_k^-) \quad (6)$$

where K_1 and K_2 are Kalman gains for ΔS and Δb , respectively. Notations such as $\Delta\hat{S}$ and $\Delta\hat{S}^-$ represent estimated value and predicted value for ΔS , respectively.

2.2.2 Evaluation Test

Estimation accuracy of cycling speed was evaluated using custom-made IMUs (InvenSense MPU-9150, Bluetooth Class 1 module). Cycling speed measured with a motion measurement system (OPTOTRAK, Northern Digital Inc.) was used as reference value. One neurologically intact male (23 y.o.) propelled the pedaling wheelchair under the 2 different conditions: usual pedaling without large backlash effect and unusual pedaling causing large effects of the backlash of crank set. Three measurements were performed for each condition.

An example of estimated cycling speed by the Kalman filter is shown in Fig. 4, in which measured speeds with gyroscope (Sensor1) and with motion measurement system are also plotted. Although the result was obtained under the usual cycling condition, the effects of backlash (vibration) is observed on the plot of the speed calculated from angular velocity.

Root mean square error (RMSE) and correlation coefficient (CC) between the estimated speed and the measured speed with motion measurement system are shown in Fig. 5. In Fig. 5, low-pass filtering ($f_c = 2\text{Hz}$) of the calculated

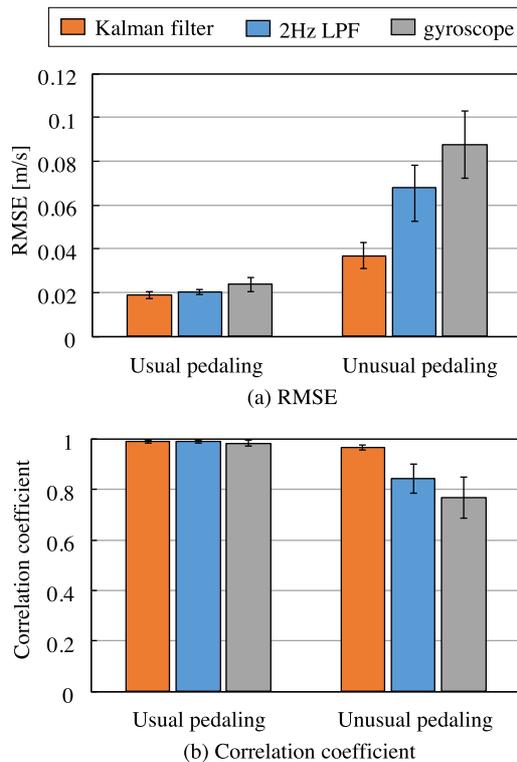


Fig. 5 Evaluation results of cycling speed estimation. RMSE and correlation coefficient were calculated for estimated speed by Kalman filtering method, low pass filtered calculated speed and calculated speed from angular velocity.

speed from angular velocity is also shown as another method of removing the backlash effects. Under the unusual pedaling condition, Kalman filtering method improved significantly RMSE and CC values. The Kalman filtering method was more effective than the LPF for compensating backlash effects in measurement of cycling speed.

3. Tests of Closed-Loop Control of Cycling Speed

3.1 Methods

Closed-loop control of 4 different cycling speeds were performed with 3 neurologically intact subjects (22–23 y.o.). Three target cycling speeds (1.1, 0.7 and 0.4 m/s) were determined from voluntary cycling of 7 healthy subjects under the cycling conditions of fast speed, moderate speed and slow speed. In addition, 0.2 m/s of target speed was also tested for FES cycling under low cadence in rehabilitation training [10], [11].

Control tests were performed in 10m straight cycling on level floor. A set of control test was performed in the order of target speed of 1.1, 0.7, 0.4 and 0.2 m/s, in which 3 minutes rest was taken between trials. Three sets of the test were performed with each subject. The resting time between sets was 5 minutes. Subjects were instructed to relax their lower limbs and not to propel the wheelchair during FES cycling control tests.

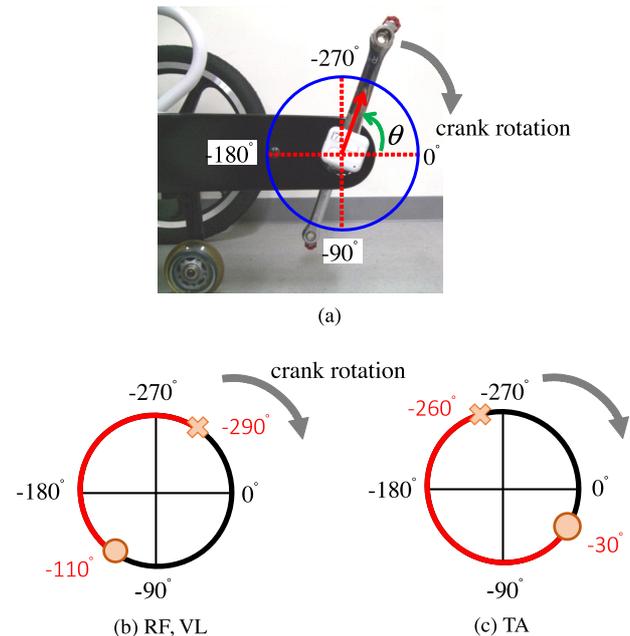


Fig. 6 Definition of angle of the right crank (a), and stimulation timing pattern of the right rectus femoris (RF) and the vastus lateralis (VL) (b) and that of the right tibialis anterior (TA) (c). Electrical stimulation to the right side muscles are applied from the crank angle shown by the circle to that by the cross, which are shown by red line. The timings of the left side muscles are shifted by 180 degrees.

Electrical stimulation was applied to the rectus femoris (RF), the vastus lateralis (VL) and the tibialis anterior (TA) of both lower limbs. These stimulated muscles were determined based on our previous studies [8], [9]. Briefly, the hamstrings and the gluteus maximus were not used considering practical application, because attachment of stimulation electrodes to these muscles sometimes needs troublesome operation with a few staffs in the case of subjects with lower limb paralysis. Stimulation intensities of the RF and the VL were adjusted by closed-loop controller, while that of the TA was fixed to the maximum value because the main function of the TA is the ankle dorsiflexion that does not contribute to cycling speed directly. Applied electrical stimulation was biphasic pulses with 30 Hz of frequency and 300 μ s of pulse width. The maximum and the minimum stimulation pulse amplitudes were determined for each muscle of each subject. The maximum amplitude was determined in order to develop enough knee extension or ankle dorsiflexion without uncomfortable feeling and the minimum amplitude of the RF and the VL were determined as the value that the knee extension begins to be developed.

Figure 6 shows definition of crank angle of the right side and stimulation timings of the right side muscles. Although the angle is defined in the clockwise direction, crank rotates anti-clockwise direction. Therefore, angles in figures are shown by negative values. The stimulation timings shown in Fig. 6 were determined from EMG signals during voluntary cycling by healthy subjects [19]. The RF, the VL and the TA were activated from 0 to -290 deg, from -110

to -290 deg and from -30 to -260 deg of crank angle, respectively, during fast speed cycling. However, stimulation timings based on these muscle activations caused backward propulsion with one subject. This was considered to be because the RF has function of hip flexion and knee extension, while the VL has function of knee extension. Therefore, the start timing of the RF has been modified from the timing of muscle activation in order to avoid inappropriate knee extension that causes backward propulsion. At the beginning of each control trial, the maximum amplitude was applied in order to achieve sufficient cycling speed. Closed-loop control was executed after the cycling speed reached the target speed.

3.2 Results

An example of closed-loop control result of cycling speed (0.4m/s of target speed) is shown in Fig. 7. Closed-loop control was started after the controlled speed reached the target value, which was at about 0.8s in this case. Cycling speed was appropriately controlled after about 5s. Overshoot of cycling speed at around 2.5s was caused because of the maximum intensity stimulation at the beginning of cycling.

Average cycling speed of each cycle is shown in Fig. 8, in which cycles including open-loop control were removed. Cycling speed of a single cycle was calculated from -180 deg of crank angle. In the 1st trial for the target speed of 1.1m/s of subject B, stimulation intensity did not almost change from the maximum value, although closed-loop control was started in the 4th cycle. The 3rd control trial for 1.1m/s of subject B did not reach the target speed (average speed was 0.86 ± 0.08 m/s). In the 2nd and the 3rd control trials of the target speed of 0.2m/s of subject A, backward propulsion was caused after decreasing of the cycling speed

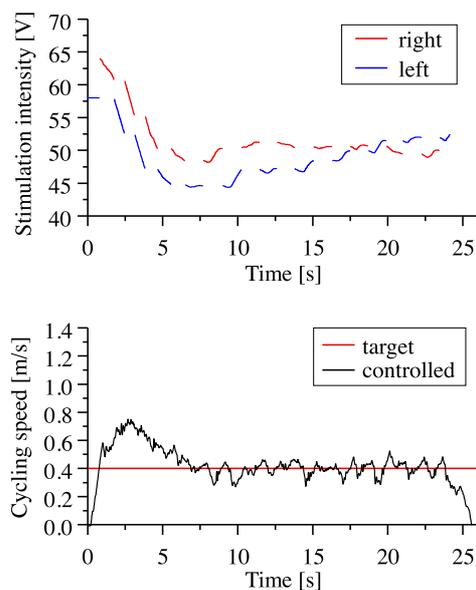


Fig. 7 An example of closed-loop control result of cycling speed (target speed: 0.4m/s, subject A). Stimulation intensity shows values for the RF.

and the wheelchair stopped.

Overshoot of cycling speed at the beginning of cycling is seen in Fig. 8 in trials with low target speeds (0.4 and 0.2m/s) and some trials of target of 0.7m/s. In those cases, cycling speed was regulated appropriately from about the 4th, the 3rd and the 6th cycle for subjects A, B and C, respectively. For 0.7m/s of target speed, cycling speed decreased temporarily at around the 8th cycle with subjects A and C.

Average values of mean error of cycling speed of each cycle were shown in Table 2, which were calculated removing cycles that were affected by overshoot. The closed-loop control system could regulate cycling speed in the steady state cycling with enough high accuracy, especially for higher cycling speed than 0.2m/s. Although average values of the mean error in a cycle were small, variation of the cycling speed during one cycle was larger a little bit than those of voluntary cycling. Average values of standard deviation of cycling speed of each cycle are shown in Fig. 9 comparing to those average values of voluntary cycling with 7 healthy subjects. The variation of the cycling speed in a cycle was larger under the lower speed control.

4. Validation Tests of Closed-Loop Control System

4.1 Methods

4.1.1 Cycling Including Turn

Since FES cycling with constant intensity stimulation sometimes came to a standstill in making a turn in a preliminary test, closed-loop control of FES cycling including turn was tested on the level floor with 3 healthy subjects (22–23 y.o.). The subjects were instructed to relax their lower limbs and not to propel the wheelchair during FES cycling control tests. FES cycling in this test consisted of straight forward cycling of 10m, left U-turn and straight forward cycling of 10m (Fig. 10). Three sets of trials of an open-loop control with the maximum stimulation intensity and a closed-loop control were performed. Resting time between trials was 3 minutes and that of between sets was 5 minutes. Parameters of electrical stimulation and determination method of those values were the same as those of the cycling speed control test described above. Target speed for the closed-loop control was 0.4m/s, since low cycling speed caused easily standstill with constant stimulation, while open-loop control used maximum stimulation intensity.

4.1.2 Repeated Cycling Control

Straight forward FES cycling (25m) on the level floor was performed repeatedly without rest until the wheelchair stands still with 3 healthy subjects (22–23 y.o.), in which the same instruction as that in the previous sections were given to the subjects. First, open-loop control was performed and then the feedback control test was performed after 5days'

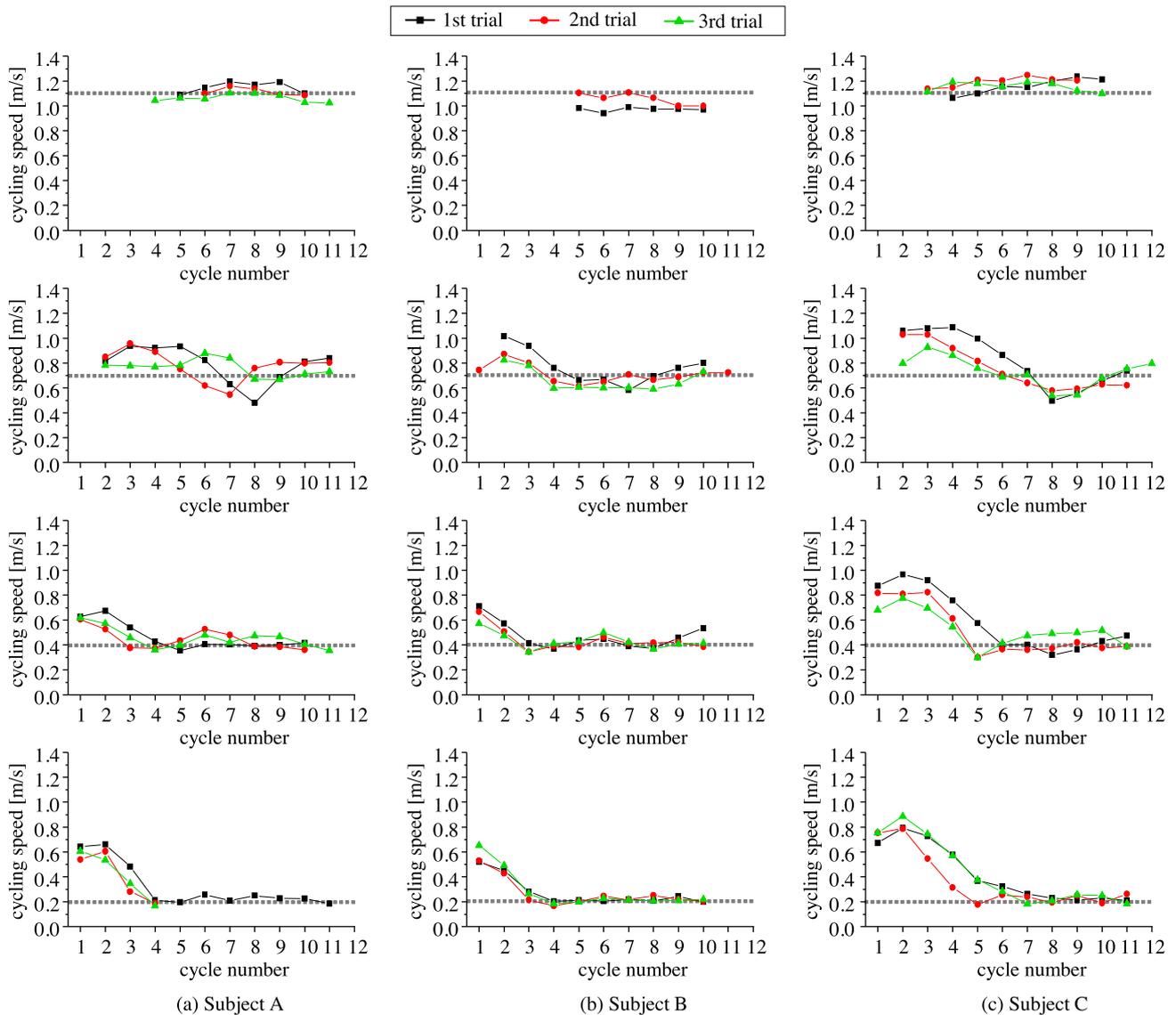


Fig. 8 Mean cycling speed of each cycle during closed-loop control. Gray dotted line shows target cycling speed, which are 1.1, 0.7, 0.4 and 0.2 m/s from the top. From the left, results of subject A, B and C are shown.

Table 2 Average values of mean error of cycling speed of one cycle during closed-loop control of cycling speed.

Target speed [m/s]	1.1	0.7	0.4	0.2
error [m/s]	0.008±0.073	0.001±0.051	0.014±0.024	0.025±0.010

interval. Parameters of electrical stimulation and determination method of those values were the same as those of the cycling speed control. Target speed for the feedback control was 0.4m/s, and open-loop control used maximum stimulation intensity.

4.2 Results

4.2.1 Cycling Including Turn

The closed-loop FES controller achieved a target cycling

speed and propelled the wheelchair during U-turn as shown in Fig. 11. Although the open-loop controller could propel the wheelchair, cycling speed decreased during U-turn. Mean cycling speed of one cycle before the turn and after the turn are shown in Fig. 12. The cycling speed did not decrease during the turn with all the trials of closed-loop control. The open-loop control could make a U-turn because cycling speed before turn was higher than 0.4m/s.

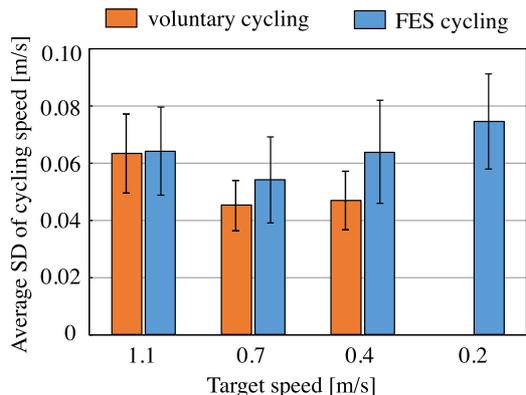


Fig. 9 Average values of standard deviation of cycling speed in one cycle. Those values of voluntary cycling are also shown by 1.1, 0.7 and 0.4m/s for fast, moderate and slow cycling, respectively.

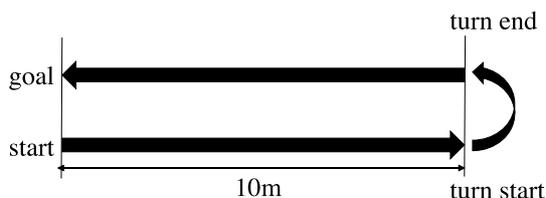


Fig. 10 Outline of cycling course for the test of cycling including U-turn.

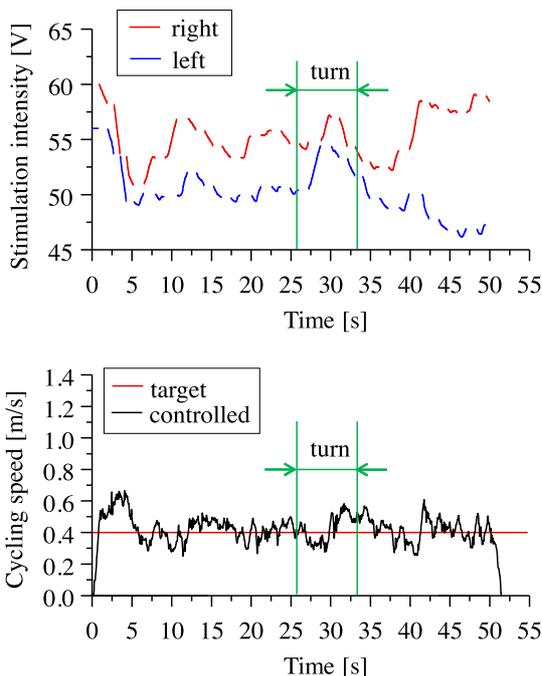


Fig. 11 An example of closed-loop control result of FES cycling including U-turn.

4.2.2 Repeated Cycling Control

Total cycling distance was calculated from the number of cycles as shown in Table 3. Each cycle was detected at

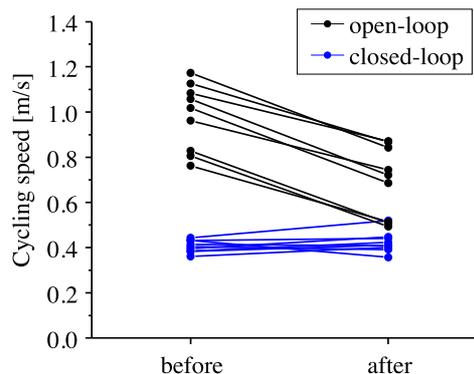


Fig. 12 Average cycling speed of one cycle before and after U-turn.

Table 3 Cycling distance in the repeated cycling test [m].

	Subj.A	Subj.B	Subj.C
open-loop	166	86	211
closed-loop	297	397	314

-180deg of crank angle. Therefore, the first and the last cycles were not included into the distance. Cycling time was from 12 to 16 minutes for closed-loop control, and from 3 to 7 minutes for open-loop control. Muscle fatigue is considered to be caused earlier with the open-loop control because of the maximum intensity stimulation. Therefore, comparison of the distance between control methods is for reference. However, closed-loop control increased cycling distance and cycling time compensating the decrease of cycling speed caused by muscle fatigue with all the subjects.

5. Discussions

Most of different target speeds were controlled appropriately by the closed-loop control system, although some of very slow speed (0.2m/s) control trials were not achieved with one subject. All control trials of low speed cycling (0.4m/s) including turn achieved maintaining the target speed without standstill. In addition, increasing of cycling distance and cycling time by the closed-loop control was suggested in the repeated cycling trials. Control ability of cycling speed shown in this paper, which showed errors less than 5% for target speeds of 0.4m/s or higher, is considered to be enough for mobile FES cycling, because closed-loop FES control of average cycling speed of each cycling period with cycling ergometer showed errors less than 5% before muscle fatiguing [12]. In addition, currently, rehabilitation with FES cycling has been tested under the condition of large differences of cadences (low or high), in which the differences were more than or equal to 30rpm (about 0.48m/s for the pedaling wheelchair) [10], [11]. Such difference of cycling speed would be achieved with the developed controller. These suggests that the developed fuzzy closed-loop control system works reliably in mobile FES cycling.

Overshoot of cycling speed at the beginning of cycling was large for the low target speed. The overshoot was caused by the maximum stimulation intensity at the begin-

ning of cycling for getting enough cycling speed. As seen in Figs. 7 or 11, this was because of large difference between the maximum intensity and stimulus intensity required for the target speed. Such overshoot is considered to be improved by increasing gain of the controller, and/or increasing fuzzy sets of the E-OAF so as to adjust gain for large error more than 5SD. On the other hand, closed-loop control method at the beginning of cycling is also desired to be modified.

In the closed-loop control of 0.7m/s of target speed, temporary decrease of cycling speed was large with 2 subjects. Delay and time constant of muscle response to electrical stimulation of the rectus femoris were more than about 180ms and 270ms in the sitting position, respectively. Since duration of applying electrical stimulation for one side was 500–600ms under the 0.7m/s of cycling speed, it is considered that cycling speed did not become stable in the duration, because applying electrical stimulation was switched between the left and the right sides before electrical stimulation developed enough large movement. Stimulation timing used in this study was determined based on EMG signals during voluntary cycling [19]. Although muscle response time [9] was tested in using with EMG-based stimulation timing, there was no significant improvement [19]. Further studies for compensating the delay in muscle response is necessary.

Variation of cycling speed in a cycle was larger during closed-loop FES control compared to voluntary cycling as shown in Fig. 9, especially for low target speeds (0.4 and 0.2m/s). One of possible cause of the variation is considered to be difference between the left and the right sides. Such variation of cycling speed is considered to lead early muscle fatigue by inefficient use of both side muscles for cycling. In addition, difference in using muscles between the left and the right sides causes difference in training effects in case of rehabilitation with the FES cycling. Therefore, improvement of the controller is desired for equal contribution of both side muscles to FES cycling or controlling the contribution to the cycling for rehabilitation of weakened side would also be necessary. Since characteristics of muscle response differ between muscles, between left and right sides and between subjects, fine tuning of the controller gain during cycling or learning type controller would be effective.

Stimulation timing used in this study was determined based on EMG signals during voluntary cycling under the fast speed condition [19]. Therefore, it was considered that large variation of mean cycling speed of a cycle and backward propulsion were caused in the case of very low target speed condition (0.2m/s). Activation timing of the rectus femoris was different both in the start and the end timings from those of the vastus lateralis in moderate and slow speed cycling conditions [19]. Considering muscle function during cycling, stimulation timing pattern of the RF for moderate and slow speed cycling have to be determined because the RF has the function of knee extension in addition to hip flexion. Variable stimulation timing control would also be required for changing cycling speed.

The closed-loop control achieved cycling time longer than 12 minutes with low cycling speed. The cycling time is considered to be useful for a single session of rehabilitation. Although the results were obtained with neurologically intact subjects, it would be possible to achieve such cycling time with paraplegic subjects after strengthening of paralyzed muscles. For rehabilitation of hemiplegic subjects, controlling contribution of muscles of the paralyzed side to the cycling would be required as discussed above.

In this paper, a closed-loop fuzzy FES controller was developed based on our previous study and a prototype of closed-loop FES control system for mobile FES cycling with the pedaling wheelchair was examined. Since FES control of mobile cycling depended on cycling system and there was no continuous time closed-loop FES controller for mobile cycling, a fuzzy controller was tested in controlling knee extension angle comparing to PID controller [17] and the developed prototype system was not compared with other control systems. Then, although ankle joint was controlled by FES during cycling without a orthotic brace, a closed-loop control for the ankle joint was not implemented. Since some subjects showed improvements of FES mobile cycling by using the hamstrings or the gluteus maximus [8], [9], it may be possible to stabilize mobile FES cycling by including these muscles. The closed-loop FES control system for mobile FES cycling is desired to be improved based on the control performance shown in this paper.

6. Conclusion

A prototype of closed-loop fuzzy control system for mobile FES cycling with pedaling wheelchair was developed using IMUs for cycling speed estimation. Most of different target speeds were controlled appropriately by the closed-loop control system. All the control trials of low speed cycling control including turn were achieved maintaining the target speed without standstill. Increasing of cycling distance and cycling time by the closed-loop control was suggested in the repeated cycling trials. The developed fuzzy closed-loop control system was suggested to work reliably in mobile FES cycling from these results. Improvement of control method at the beginning of cycling, controlling the contribution of muscles to cycling for rehabilitation of both sides, and including variable stimulation timing control according to cycling speed would be desired for practical rehabilitation with mobile FES cycling with the pedaling wheelchair.

Acknowledgments

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