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RESEARCH ARTICLE

Motor Adaptation Effect of Active Joint Movement During Cycling Generated by Ankle Assist Ergometer: A Proof-of-Concept Study in Healthy Participants

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This work involved human subjects or animals in its research. Approval of all ethical and experimental procedures and protocols was granted by the Research Ethics Committee of Tokyo Kasei University.

ABSTRACT To promote multi-joint movement considering limb coordination, we introduce a portable ankle assist ergometer (AAE) to deliver passive ankle joint motion at a specific crank angle per cycle during cycling. This study aimed to verify the motor adaptation effect using the developed AAE in healthy young participants. We converted the pedals of a commercial ergometer into an ankle-assisting tool driven by a motor and brake. The AAE delivered ankle dorsiflexion assistance to young adult participants from 90° after the top dead center until the bottom dead center of the crank angle during cycling. We obtained 1) a significant difference between the ankle dorsiflexion angular velocity at the bottom dead center before and immediately after assistance ($p < 0.05$) and 2) an increase in dorsiflexion muscle activation ($p < 0.05$). These effects persisted after the removal of the assistance. Our results suggest that the AAE activates the central pattern generator and induces motor adaptation of ankle movement by limb coordination during a motion cycle. We believe that the proposed AAE contributes to eliminating barriers to the clinical implementation of the rehabilitation tools delivering cyclic motion.

INDEX TERMS Ankle assistance, cyclic rehabilitation, ergometer, motor adaptation.

I. INTRODUCTION

In aging societies, walking obstacles for the population must be overcome to prevent people from becoming bedridden. Rhythmic and cyclic movements such as walking employ at least partially separate control mechanisms in the motor system [1]. From the perspective of use dependence, interventions for walking and other movements are required to improve locomotion [2]. Human bipedal walking can adapt to various environments by eliciting usable motor commands through the nervous system. During walking, limb movement is generated by motor commands coordinated between the legs (interlimb coordination) and between the joints in each leg (intralimb coordination) [3]. Such coordination allows us to control the movements of multiple joints. By applying the underlying mechanisms of interlimb and intralimb coordination, previous studies using split-belt treadmills (separated double-belt treadmills) have clarified the effect of changes in joint movement during walking on non-disabled people [4], [5] and patients suffering from movement disorders [6]. Ergometer training through cyclic movements such as walking can promote walking-like muscle activity and is

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associated with walking function improvement [7], [8], [9]. The effectiveness of ergometer exercises as an alternative to walking is supported by the common core hypothesis proposed by Zehr [10]. This hypothesis claims that upper- and lower-limb movements during walking, stepping, and ergometer exercises are controlled by a common rhythm and central pattern generator.

Rehabilitation using a double-belt or body-weight-supported treadmill promotes the adaptation of rhythmic and cyclic movements during walking. However, these treadmills are sometimes difficult to apply in clinical practice owing to problems with capital investment and the patient's ability [11].

We developed a portable ankle assist ergometer for promoting adaptation of rhythmic and cyclic movements that shared neural control with walking and verified its effectiveness. A portable ergometer is affordable and easy to transport and install. We believe that this is a promising rehabilitation tool that can be easily introduced into clinical practice. To provide stimulation equivalent to a split-belt treadmill using an ergometer, it is necessary to provide the same passive ankle joint motion at a specific crank angle per cycle during cycling movement. The purpose of this study was to develop a system to link the ankle joint motion with crank angle and to confirm the corresponding motor adaptation effect on healthy young subjects.

II. MATERIALS AND METHODS

A. ANKLE ASSIST ERGOMETER

Fig. 1 shows a measurement chain indicating all the electrical flows in the ankle assist ergometer.

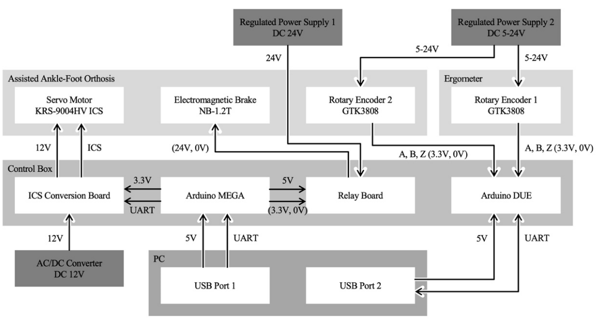

FIGURE 1. Measurement chain of ankle assist ergometer.

1) ERGOMETER

The ergometer was modified from a commercial product (PBE-100-(2); Meisei Co., Tokyo, Japan; Fig. 2a and Table 1). The crank had a resistive torque of 0.51 Nm at rest, and its angle was obtained using a rotary encoder (GTK3808; Oumefar®; Fig. 2b and Table 1) fixed to the crankshaft using a pulley and belt (Fig. 2c). The crank angle was measured at a 100 Hz sampling frequency using a microcomputer (Due; Arduino®, Somerville, MA, USA), mathematically processed by a software application, and stored in the CSV format.

TABLE 1. Specifications of proposed ankle assist ergometer.

Ergometer	Model Number	PRE-100-(2)
	Manufacturer	Meisei Co., Ltd.
	Weight	7100 (g)
	Size	W:420 × L:400 × H:310 (mm)
	Maximum Static Torque (Measured value)	0.51 (Nm)
Encoder	Model Number	GTK3808
	Manufacturer	Oumefar Co., Ltd.
	Weight	180 (g)
	Size	W:55 × L:38.7 × H:45 (mm)
	PCD	46 (mm)
	Shaft Diameter	8 (mm)
	Resolution	1000 (pulse / r)
	Operating Supply Voltage	DC 5–24 (V)
Servo Motor	Model Number	KRS-9004HV ICS
	Manufacturer	Kondo Kagaku Co., Ltd.
	Weight	104 (g)
	Size	W:51.5 × L:32 × H:39.5 (mm)
	Maximum Torque (11.1 (V))	8.91 (Nm)
	Maximum Velocity (11.1 (V))	0.23 (sec / 60 deg)
	Operating Supply Voltage	DC: 9–24 (V)
Electromagnetic Brake	Model Number	NB-1.2T
	Manufacturer	Sinfonia Technology Co., Ltd.
	Weight	650 (g)
	Size	W:88 × L:88 × H:27 (mm)
	Breakaway Torque	12 (Nm)
	Rated Voltage	DC: 24 (V)
	Power Consumption	11 (W)
Ankle-Foot Orthosis	Weight	2025 (g)
	Size	W:230 × L:255 × H:350 (mm)
	Foot Size	Free
	Maximum Assist Torque	8.91 (Nm)
	Maximum Resistance Torque	12 (Nm)

2) ASSIST DEVICE

The ankle assist device was constructed in the shape of an ankle joint orthosis for rehabilitation (Fig. 1d and Table 1). The assist device had three functions: 1) ankle joint angle acquisition, 2) dorsiflexion assistance, and 3) ankle joint fixation. Angle acquisition was performed using the same rotary encoder used to acquire the crank angle, and the output was synchronized with the crank angle (100 Hz). A servomotor (KRS-9004HV, ICS; Kondo Kagaku, Tokyo, Japan; Table 1)

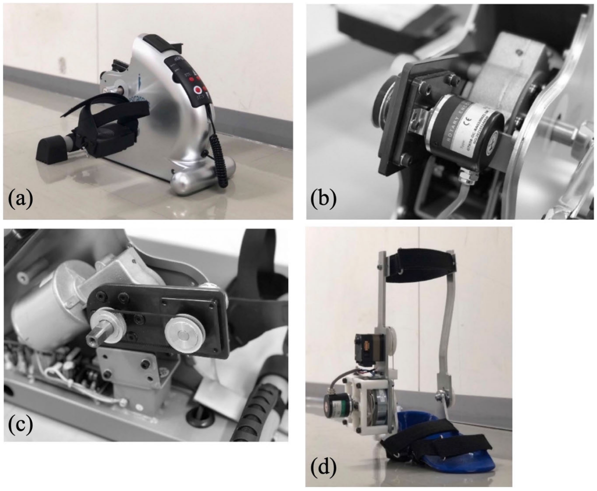

FIGURE 2. Ankle assist ergometer comprising (a) ergometer, (b) rotary encoder, (c) pulley and belt, and (d) ankle-foot orthosis.

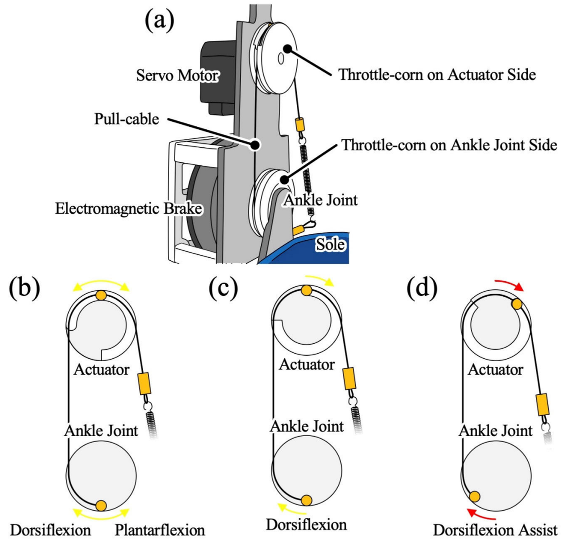

FIGURE 3. (a) Pull cable mechanism consisting of a pull cable and throttle corns. The pull cable is a stainless-steel cable with clasps at one end and in the middle. A throttle corn is placed on the ankle joint side, and another one is placed on the actuator side. The ankle joint side has a groove carved to fit the clasp at one end of the cable, and the actuator side has a groove carved to fit the clasp in the middle of the cable.

was used as the actuator, and a pull cable mechanism was used to transmit the rotational torque (Fig. 3a). The pull cable mechanism consisted of a pull cable and throttle corns. The steel pull cable had clasps at one end and in the middle. Two throttle corns were used, one on the ankle joint side and the other on the actuator side. The ankle joint side had a groove carved to fit the clasp at one end of the cable, and the actuator side had a groove carved to fit the clasp in the middle of the cable. When the clasp was within the range of motion of the throttle-corn groove on the actuator side, no assistance was applied, and the ankle joint was unrestricted (Fig. 3b). When the actuator rotated forward at a certain angle, the clasp reached the limit of the range of motion (Fig. 3c), and further forward rotation transmitted the rotational torque to assist the ankle joint in the dorsiflexion direction (Fig. 3d). The device also included an excitation-type electromagnetic brake (dry single-plate thin electromagnetic brake, NB-1.2T; Sinfonia Technology, Tokyo, Japan). The actuator and electromagnetic brake were controlled using a microcomputer (MEGA 2560; Arduino$^{\tilde{o}}$) according to the crank angle.

3) CONTROL SYSTEM

The two microcomputers and a software application written in Visual Studio® (Microsoft®, Redmond, WA, USA) constituted the control system with the control flow detailed in Fig. 4a. The rotary encoders used to measure the crank angle and ankle joint angle were controlled using the Due microcomputer. The resolution of the rotary encoder was 1000 pulses and increased to 2000 pulses by multiplying by two through programming. Thus, the angle could be measured in 0.18° increments. The data from the two angles were transmitted to the Due microcomputer and output as time-series data with a 100 Hz sampling frequency in the CSV format. The CSV data were sent in real time to the application running on a computer. The application could send a command to the MEGA microcomputer according to the crank angle. The MEGA microcomputer controlled the servomotor and electromagnetic brake installed in the assist device according to the commands sent from the Visual Studio application. In other words, the crank angle was monitored only during the assist period, and the servomotor and electromagnetic brake were controlled via the MEGA microcomputer according to the set crank angle to assist the motion of the right ankle joint. The timing and on/off state of the assist and brake commands for one crank revolution are shown in Fig. 4b. To activate the assist device at 90° after the top dead center, the command was sent from the application to the MEGA microcomputer at 80° after the top dead center to consider the communication delay. Assistance was provided for 0.45 s of rotation in the dorsiflexion direction, and the electromagnetic brake was immediately activated afterward. To ensure that the ankle joint was completely free at the bottom dead center, the command for releasing the servomotor and electromagnetic brake was sent at 10° before the bottom dead center to consider the communication delay. Given that the cycling velocity was not constant, this release command was prioritized in any case, and both the servomotor and electromagnetic brake were released at the bottom dead center.

System verification is described in appendix.

B. EXPERIMENTAL PROCEDURE AND TASKS

In this study, 16 healthy young adult participants (age, 22 ± 1 years; weight, 62.7 ± 8.6 kg; height, 1.70 ± 0.09 m) without a history of neurological or musculoskeletal disorders performed tasks after receiving sufficient practice and rest. All the participants provided written informed consent before the experiments. The study was conducted in accordance with the Declaration of Helsinki. The research ethics committee of the researchers' institution approved the experimental procedures (SKE2021-01).

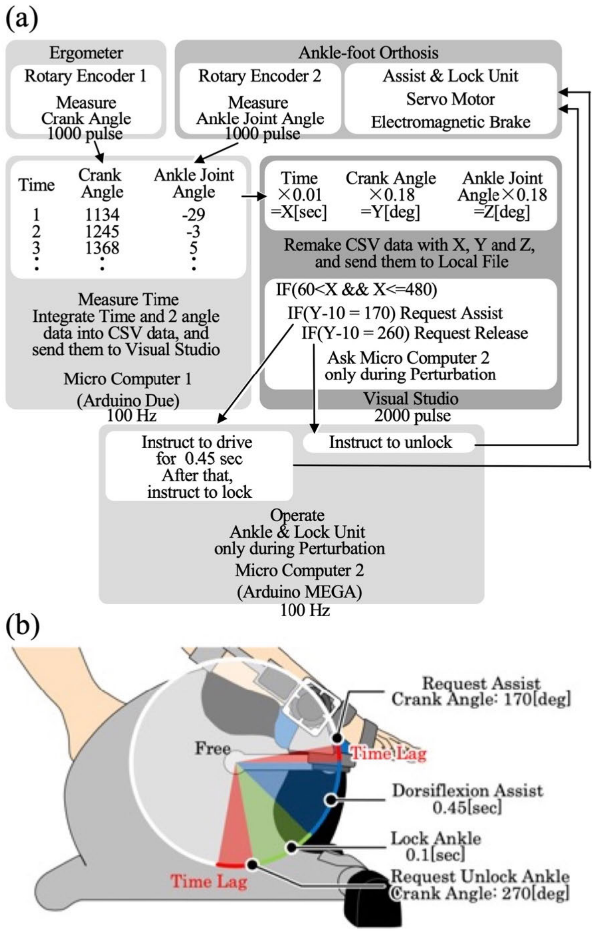

FIGURE 4. Control flow of ankle assistance. To activate assistance at 90° after the top dead center, the corresponding command was sent from the Visual Studio application to the MEGA microcomputer at 80° after the top dead center to consider the communication delay. Assistance was provided for 0.45 s of rotation in the dorsiflexion direction, and the electromagnetic brake was immediately activated afterward. To free the ankle joint motion at the bottom dead center, the command for releasing the servomotor and electromagnetic brake was sent at 10° before the bottom dead center to consider the communication delay.

Each participant pedaled with plantar flexion at 90° after the top dead center (90° before the bottom dead center) for 1 min with no assistance (baseline), for 7 min with assistance (assist), and again for 3 min with no assistance (washout) (Fig. 5a). The participants matched the timing of plantar flexion at 90° after the top dead center to the display of a metronome application running at 60 bpm (Fig. 5b). To prevent conscious assistance by the mechanical sound of the ergometer, each participant wore earplugs and headphones and pedaled while matching to the blinking shown on the metronome display.

Five of the participants (age, 24 ± 2 years; weight, 66.4 ± 12.8 kg; height, 1.69 ± 0.08 m) took part in an additional experiment (same protocol) to determine whether the observed increase in ankle dorsiflexion led to muscle activation by a consequence of ergometer assistance. Electromyography (EMG) activity of the soleus (SOL) and tibialis anterior (TA) muscles was recorded bilaterally using

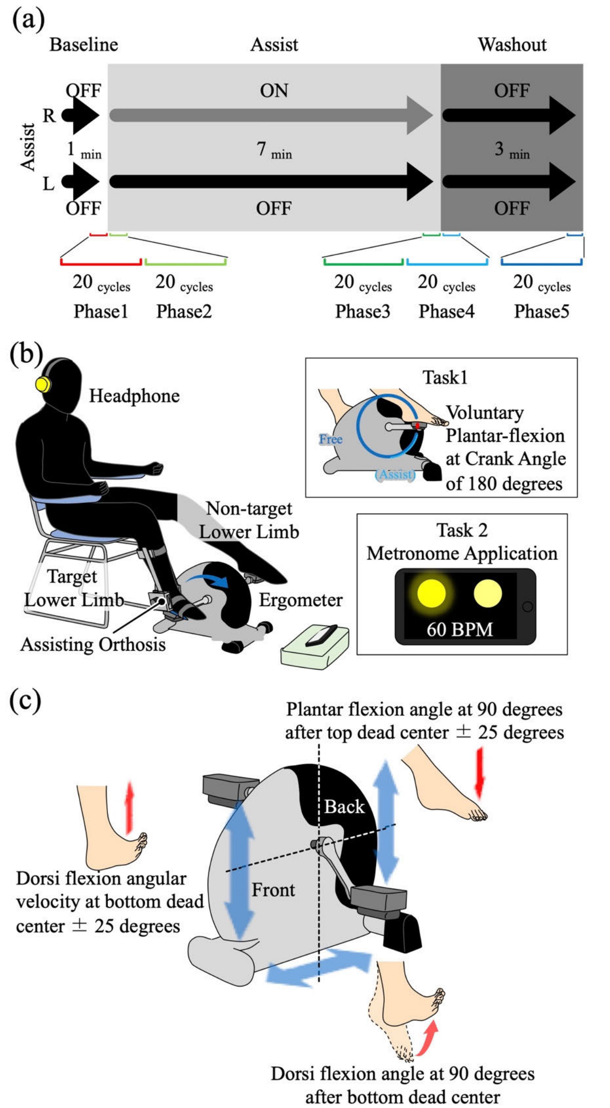

FIGURE 5. (a) Task protocol. Each participant pedaled with plantar flexion at 90° after the top dead center (90° before the bottom dead center) for 1 min with no assistance (baseline) and then for 7 min with assistance (assist), followed by 3 min with no assistance (washout). (b) Diagram of experimental setup and task. The participant matched the timing of plantar flexion at 90° after the top dead center with the display of a metronome application running at 60 bpm continuously. (c) Testing parameters using the right foot. 1) Plantar flexion at 90° after top dead center with ±25°, 2) dorsiflexion at 90° after bottom dead center, 3) dorsiflexion angular velocity at bottom dead center with ±25°, and 4) (additional experiment) root-mean-square EMG activity of TA/SOL muscles at bottom dead center with ±25°.

surface electrodes (Trigno® Wireless System; DELSYS®, Natick, MA, USA). The opposite ankle angle was recorded using a goniometer (FAC-GM150; 4Assist, Tokyo, Japan). The recorded EMG and goniometer signals were digitized simultaneously along with the ergometer crank data.

C. DATA ANALYSIS

Data analysis was defined as follows (Fig. 5a). Phase 1 involved the last 20 cycles of baseline, phases 2 and 3 involved the first and last 20 cycles of assist, respectively, and phases 4 and 5 involved the first and last 20 cycles of

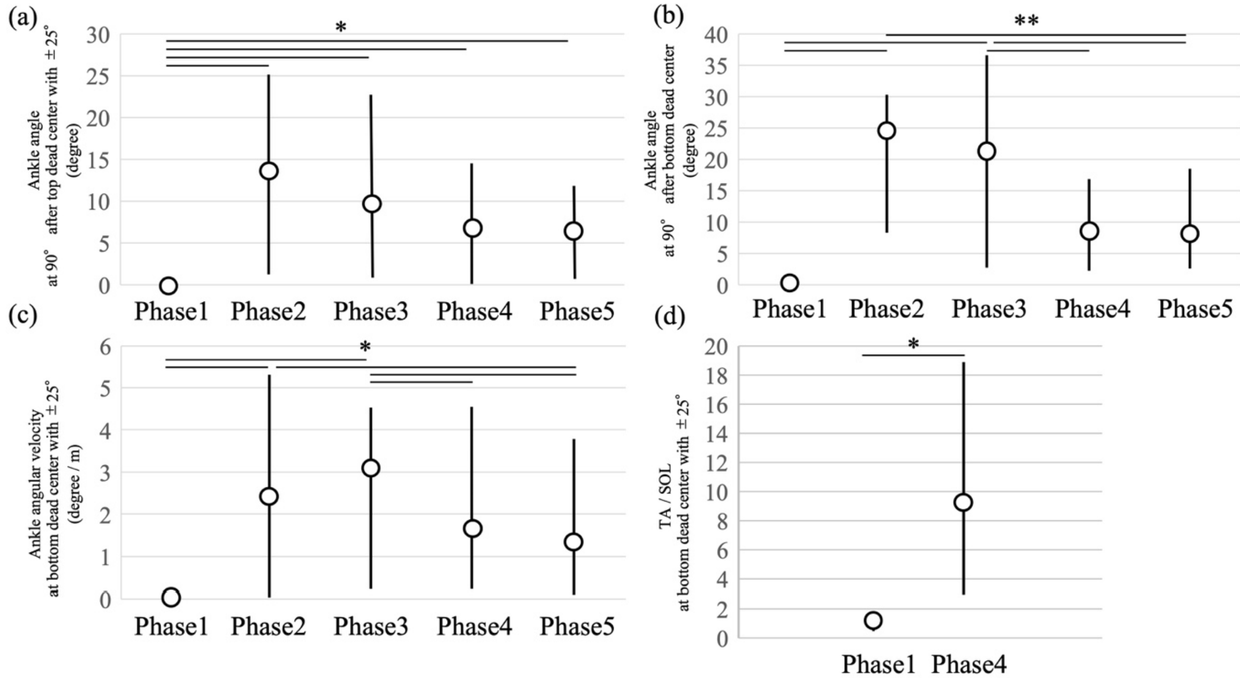

FIGURE 6. Statistical results of (a) plantar flexion at 90° after top dead center with ±25°, (b) dorsiflexion at 90° after bottom dead center, (c) dorsiflexion angular velocity at bottom dead center with ±25°, and (d) root-mean-square EMG activity of TA/SOL muscles at bottom dead center with ±25°. *$p < 0.05$, **$p < 0.01$.

washout, respectively. The testing parameters (Fig. 5c) were the following: 1) plantar flexion at 90° after the top dead center with ±25°, 2) dorsiflexion at 90° after the bottom dead center, 3) dorsiflexion angular velocity at the bottom dead center with ±25°, and 4) (additional experiment) root-mean-square EMG activity of the TA/SOL muscles at the bottom dead center with ±25°.

The joint angle and EMG were measured and processed through a lowpass filter at 5 Hz and bandpass filter from 20 to 450 Hz using a fourth-order zero-phase-lag Butterworth filter. To compare the results for each parameter, we used a repeated measures analysis of variance with a within-subject factor of phase. The significance level was set to $p = 0.05$ and adjusted using Scheffe correction for posthoc multiple comparisons. All analyses were performed using MATLAB® (MathWorks®, Natick, MA, USA). The experimental and statistical protocol followed a split-belt treadmill study, which is similar to motor adaptation research [3]. To compare the results of ankle dorsiflexion muscle activation before and after assistance, we used a paired t-test between phases 1 and 4.

III. RESULTS AND DISCUSSION

1) Plantar flexion at 90° after the top dead center with ±25° showed no significant differences across phases 2–5. The participant's voluntary plantar flexion movement was performed accurately (Fig. 6a). 2) Dorsiflexion at 90° after the bottom dead center was significantly different in phases 2 and 3 compared with that in phases 1, 4, and 5 ($p < 0.01$, Fig. 6b). There were no significant differences between phases 1, 4, and 5 ($p = 0.05$–7). After assistance, the dorsiflexion angle at the bottom dead center did not change without mechanical assistance at the bottom dead center. 3) The dorsiflexion angular velocity at the bottom dead center with ±25° was significantly different between phases 1 and 2–4 ($p < 0.05$, Fig. 6c). There was no significant difference between phases 1 and 5 ($p = 0.19$). After immediate assistance, the dorsiflexion angular velocity at the bottom dead center changed without mechanical assistance. 4) (Additional experiment) The root-mean-square EMG activity of the TA/SOL muscles at the bottom dead center with ±25° was significantly different between phases 1 and 4 ($p < 0.05$, Fig. 6d).

This study aimed to confirm the adaptation of joint movements during cycling by using the developed ankle assist ergometer. The results showed that the ergometer assists in a specific crank angle during cycling, leading to the generation of ankle dorsiflexion angular velocity and muscle activation. Our results are consistent with those of previous studies. Cleland et al. [12] compared the lower leg muscle activity and joint movement between coupled and uncoupled cranks. In coupled crank in healthy adults and stroke patients, TA activity was cohesive and efficient (i.e., the muscle activity range from maximum to minimum was small), and the synchronous of the joint movement was increased. Furthermore, a study using an ankle assistance device revealed that the timing of dorsiflexion assistance was related to TA activity in healthy participants during walking [13].

During the split-belt treadmill task, leg multi-joint movement adapts to the continuous disturbance caused by the double belt, eliciting new leg movements as an adaptive change [14]. The total involvement of the cortical process is relatively larger at the beginning of adaptation to the split-belt-induced perturbation. In [15] and [16], error-based learning enabled a participant to update the internal model for walking, which presumably occurs in the cerebellum. This adaptive change is seen in interlimb parameters (e.g., step length and double-support time) as gradual adaptation [17]. In ergometer cycling, pulling up from the bottom dead center contributes to ankle dorsiflexion. In the present study, as the ankle assist ergometer was placed in front of the participant, dorsiflexion contributed to pulling up from slightly in front of the bottom dead center to 90° after the bottom dead center. During assistance, the participant was given a perturbation that was forced to dorsiflex before the pulling phase entered but after the maximum plantar flexion at 90° before reaching the bottom dead center. Therefore, assistance induced a perturbation, and thus the participant performed an error-based task. As a result, new ankle dorsiflexion movements were learned during the same phase.

Previous studies have demonstrated adaptive changes not only in interlimb parameters but also in TA activation during walking [18]. This is because predictive feedforward control is required to set the optimal ankle stiffness to prepare for the next perturbation in heel contact [18], [19], [20]. The participants in our study learned to start the dorsiflexion movement predictively early by being assisted in the dorsiflexion movement in a phase that was not necessary. As a result, motor adaptation increased the TA activity immediately after assistance. Previous studies have confirmed that repetitive passive movements increase the excitability of the motor area of the cortex [21]. Moreover, other studies have shown that the effect is constant and unaffected by the range of motion [22]. These findings may partly explain the neurological mechanisms underlying our results.

The developed ankle assist ergometer promoted central pattern generation and induced motor adaptation of ankle movements by limb coordination during a cycle. Previous studies have shown that ergometer exercise improves walking ability in patients with central nervous system diseases [23], [24]. Therefore, the ergometer may activate a set of similar residual neural pathways to strengthen interlimb and intralimb neuronal coupling to improve walking performance after stroke and in patients suffering from health conditions such as Parkinson's disease [8].

The applicability of our study to other tasks (walking) and subjects (stroke patients) is limited. Owing to use dependence, the direct effect on the improvement of walking ability could not be confirmed in this study because the walking ability was not measured before and after ergometer training. The application in patients with central nervous system diseases whose ankle joint motion is impaired remains to be verified. A potential limitation of our methodology is the tradeoff between the accuracy of the device's assistance and the naturalness of the cycling. Our AAE does not provide appropriate assistance at fast crank cycling velocity. A low velocity is inadequate due to the lack of interlimb coordination during the voluntary and phasic cycling motion. Therefore, a feasible, constant pedaling speed was ensured by assisting the participants with the metronome. Thus, we selected the application of a uniform speed over the participants' natural cycling speeds. Moreover, because our study only consists of a few young adults, our findings are limited in terms of their generalizability to other populations. However, in diseases with keeping ability to motor adaptation, the same effects with subjects of present study are expected.

IV. CONCLUSION

We developed an ankle assist ergometer that generates unilateral ankle joint motion during cycling. The ergometer provided accurate ankle dorsiflexion assistance to healthy subjects according to the crank angle. As a result, the ankle dorsiflexion angular velocity and dorsiflexor muscle activity improved. This effect persisted after assistance was removed.

We aimed to eliminate barriers to the clinical implementation of rehabilitation tools that deliver rhythmic and cyclic movements. The motor power and control software of the developed device facilitated accurate movement of the ankle joint of a healthy adult of normal body size. We succeeded in developing a rehabilitation ergometer tool (including a control computer) that can be used by a person with ease. In future studies, we will improve the ankle assist ergometer and further validate the effect of choosing a simple rehabilitation tool to improve the walking ability without using treadmills.

DECLARATION OF INTEREST

There are no conflicts of interest, financial or otherwise, to declare.

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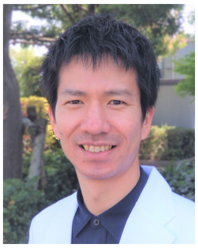

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