Effects of Assisted Dorsiflexion Timing on Voluntary Efforts and Compensatory Movements: A Feasibility Study in Healthy Participants

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***Abstract*— In previous research, we found that modulating the assistance timing of dorsiflexion may affect a user's voluntary efforts. This could constitute a focus area based on assistive strategies that could be developed to foster patients' voluntary efforts. In this present study, we conducted an experiment to verify the effects of ankle dorsiflexion assistance under different timings using a high-dorsiflexion assistive system. Nine healthy and young participants wore a dorsiflexion-restrictive device that enabled them to use circumduction or steppage gaits. On the basis of the transition from the stance to the swing phase of the gait, the assistance timings of the high-dorsiflexion assistive system were set to have delays, which ranged from 0 to 300 ms. The index results from eight out of nine participants evaluated compensatory movements and revealed positive strong/moderate correlations with assistance delay times ($r = 0.627$–0.965, $p < .001$), whereas the other participants also performed compensatory movement when dorsiflexion assistance timing was late. Meanwhile, the results from tibialis anterior surface electromyography from six out of nine participants showed positive strong/moderate correlations with dorsiflexion assistance delay times ($r = 0.598$–0.922, $p < .001$), indicating that tuning the assistance timing did foster these participants' voluntary dorsiflexion movements. This result indicates that there should be a trade-off between ensuring voluntary dorsiflexion movements and preventing incorrect gait patterns at different assistance timings. The findings of this feasibility study indicate the potential of developing an adaptive control method to ensure voluntary efforts during robot-assisted gait rehabilitation based on assistance timing modification. A new assistance mechanism should also be required to stimulate and motivate a patient's voluntary efforts and should reinforce the effects of active gait rehabilitation.**

***Index Terms*— Gait rehabilitation, assist-as-needed, voluntary effort, powered ankle–foot orthosis, compensatory movement.**

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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 Ethics Committee of Waseda University under Approval No: 2017-044.

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I. Introduction

THE poststroke population has been increasing rapidly. This has led to increased requirements for rehabilitation programs and follow-up care, which is a critical issue around the world [1]. A stroke survivor with hemiplegia experiences an enforced drop foot, "toe-down" posture on their paralyzed side during the swing phase of their gait [2]. This leads to insufficient minimum toe clearance (MTC), which is believed to create a high-stumbling risk [3]. To ensure sufficient MTC, patients with hemiplegia must use compensatory gait patterns, such as circumduction and steppage gaits [4], [5].

A powered ankle–foot orthosis (PAFO) is currently used in medical facilities and rehabilitation centers for assisting gait rehabilitation. Research studies on PAFO have sought to allow patients to relearn ankle movements during walking. To prevent drop foot, assistance of dorsiflexion movement during the swing phase in gait is especially important [6]. A few PAFO studies have focused on providing sufficient dorsiflexion in the swing phase to ensure foot clearance [6]–[8]. It is hoped that through the intervention of PAFOs, patients can relearn how to practice sufficient MTC and thus lower their stumbling risk and prevent compensatory gaits. For instance, Soft Exosuit and RE-Gait provide ankle assistance based on the detection of gait anomalies and have shown potential for improving patients' gaits, including dorsiflexion movements following intervention [9], [10]. However, two major challenges have arisen in the course of recent PAFO research.

The first of these challenges is the facilitation of patients' voluntary efforts during PAFO intervention. In recent years, various studies have noted the importance of ensuring patients'

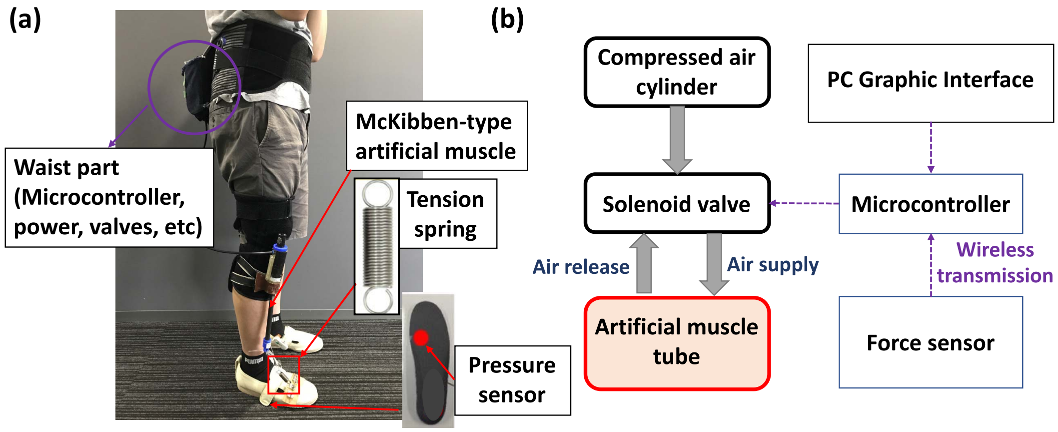

Fig. 1. The improved high-dorsiflexion assistive system; (a) appearance of the whole system; (b) configuration of the system.

voluntary efforts while being assisted by robotic devices during rehabilitation [11]. Robotic rehabilitation systems have been reported to have positive effects on gait recovery based on early intervention by implementing trajectory control strategies utilizing predefined repetitive gait patterns [12], [13]. Current studies of lower-limb exoskeletons, such as those of Lokomat and ALEX, however, aim to further enhance voluntary hip and knee movement efforts in patients with chronic and subacute stroke via assist-as-needed control (AAN) strategies [14]–[17]. For current PAFO devices, position and force control remain the primary control methods [7]–[10]. Although ReWalk ReStore^{TM} can manually modify assistance force [18] and Anklebot and Stevens Ankle–Foot Electromechanical orthosis applied impedance control [19], [20], very limited quantitative evaluations of voluntary ankle efforts during over-ground walking have yet been reported in the literature.

The second challenge is ensuring an optimal gait pattern while implementing a control strategy that facilitates voluntary ankle movements with a PAFO system. Regardless of the method used, increasing a user's voluntary effort also entails decreasing the extent of assistance. Lower-limb exoskeletons utilizing AAN control are usually accompanied by bodyweight support to reduce the patients' weight-bearing burden and are often worn by patients on speed-adjustable treadmills to create stable conditions [14], [15]. For PAFO devices, however, the main usage is envisioned as over-ground walking assistance. Insufficient assistance can result in stumbling or compensatory gait patterns in patients. In other words, when assisting dorsiflexion in the swing phase, there should be a trade-off between ensuring voluntary efforts and maintaining of safety and optimal gait patterns.

In our previous work, we developed a high-dorsiflexion assistive system that aimed to assist over-ground gait rehabilitation of patients with poststroke hemiplegia [21]. Although a pilot study in this work indicates the potential for improving the dorsiflexion angles of poststroke hemiplegia patients, another study indicated that our system cannot ensure even a healthy user's voluntary dorsiflexion movements, that resulted in significantly decreased surface electromyography (sEMG) in the tibialis anterior [22]. The reason was an excessively early activation of dorsiflexion assistance timing. Meanwhile, reducing assistance levels by reducing assistive torque leads to an increase in the sEMG signals of the targeted muscles [23]. Therefore, we speculated that voluntary movements could be facilitated by setting a delay in the activation of assistance. In this present feasibility study, we would conduct an experiment to assess the effects of our improved high-dorsiflexion assistive system on healthy participants using compensatory gait patterns due to restriction of their swing phase dorsiflexion movements. We assumed that later assistance timings may not only lead to increase in the extents of voluntary movements but also decrease MTC, and the participants' extents of compensatory movements could also be enhanced. We aim to assess whether modifying dorsiflexion assistance timing is an option of modifying assistance level and whether it could facilitate voluntary efforts and affect physical kinematics.

II. Materials and Methods

A. Overview of the High-Dorsiflexion Assistive System

In this study, we used a high-dorsiflexion assistive system that was improved from our previous prototype [21], [24]. The system and its configuration are shown in Fig. 1. Compared with the previous prototype, the weight of the lower-limb segment was reduced to 0.35 kg. Although lightweight, the system could provide dorsiflexion torque up to 75 Nm, which is much larger than the 13.5 Nm torque required for assisting dorsiflexion movement from the drop-foot posture [26]. This further enhances the portability of our system and reduces weight-loading burden on patients.

The hardware elements were partially upgraded compared with the previous version. The sensing unit consisted of insoles with force sensors (Flexiforce A301 Sensor, Tekscan Inc., USA) placed at the forefeet, wirelessly transmitted with wireless modules (nRF24L01, Nordic Semiconductor, Norway). The actuation unit contained a McKibben-type artificial muscle (DMSP-10, Festo Inc., Germany) and a tension spring that was displaced in series between the forefoot and knee (Tokai Spring Industries Inc., Japan). Active dorsiflexion movement was assisted by contracting the artificial muscle to lift the forefoot. For the control unit, a microcontroller with an embedded wireless module (nRF52832, Nordic Semiconductor, Norway), a portable multifunction jump starter as power source (Portec Inc., USA), and a portable air cylinder containing 74 g of carbon dioxide gas were included. For the convenience of this study, we used a 38 L air tank to provide compressed 38 L air source (AST-40, Fujiwara Sangyo Co., Japan). The microcontroller manipulated the solenoid valves

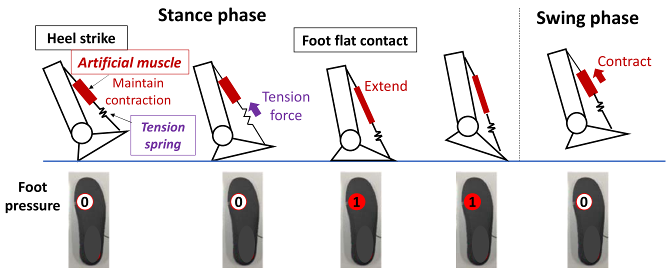

Fig. 2. Control of the high-dorsiflexion assistive system during the gait cycle.

(BV214A, MAC Valves Inc., USA) for air injection based on detected gait events. A graphical interface on a computer was used to manually tune delay of dorsiflexion assistance. The manipulation command was also transmitted through a wireless module (nRF52840, Nordic Semiconductor, Norway).

Fig. 2 depicts the high-dorsiflexion assistive system's assistance in the gait cycle. The dorsiflexion support was activated by a toe-off event, with a gait transit taking place from the stance to the swing phase in the instances in which no foot pressure was detected. The pressure sensor was positioned at the first metatarsal point of the insole so that the artificial muscle did not extend from the flat part of the foot until mid-stance. During the loading response phase, the tension spring supports a resistive dorsiflexion mechanism known as the heel rocker function. The spring is replaceable according to the required assistance level of heel rocker function of the user. The minimum selection of the spring coefficient was 2.65 N/mm, and the displacement during the swing phase was negligible [26]. In this study, we focused on the swing phase of dorsiflexion assistance provided by the artificial muscle.

B. Assumptions on the Effects of Assistance Timing on Gait

As noted earlier, we established in our previous work that the early assistance timing of dorsiflexion in the swing phase could hinder the user's voluntary efforts [22]. Because of the toe-off event detection with pressure from the first metatarsal bone, early assistance timing altered the original ankle dorsiflexion pattern, as shown in Fig. 3. This caused the user to fail to dorsiflex voluntarily prior to activation of our high-dorsiflexion assistive system. Studies with event-triggered dorsiflexion monitoring of the foot pressure at the first metatarsal joint also reported the intervention results of a possible early dorsiflexion or the decrease in the push-off plantarflexion angle [27], [28].

Meanwhile, our previous results indicated that the sEMG from the tibialis anterior significantly decreased when our system was providing its assistance compared with that of normal walking. However, reducing the levels of assistance, such as the levels of torque, resulted in an increase in the sEMG signals of the targeted muscles [23]. Therefore, we speculated that voluntary movements could be facilitated by setting a delay for the activation of assistance.

Different assistance timings may influence the extent of voluntary movements, but their relationship with safety and gait kinematics should not be neglected. By increasing the extent of voluntary dorsiflexion efforts in the swing phase with

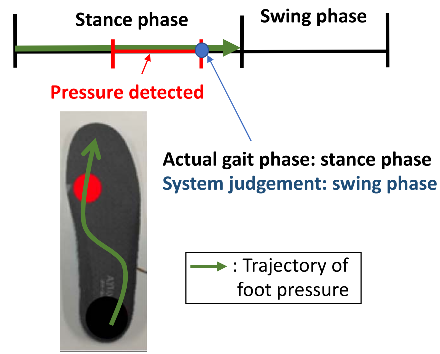

Fig. 3. Detected swing phase toe-off event and actual gait phase transition.

Fig. 4. Dorsiflexion-restrictive device.

decreasing assistance, a decrease in MTC and the patient's utilization of compensatory movements could occur. Although the relationship among ankle dynamics, kinematics, MTC, and compensatory gait patterns is considered a logistical theory in the clinical field, only a few studies on conventional rigid ankle–foot orthoses and PAFO systems have been conducted [28]–[30]. To our understanding, no systematic evaluation of PAFO systems has been performed to verify whether theory applies to other different assistance levels.

C. Dorsiflexion-Restrictive Device

Fig. 4 shows an ankle–foot orthosis that restricts dorsiflexion movement by means of a tension spring set above the posterior parts of the calf. Contractures in patients with post-stroke hemiplegia include plantarflexion and the introversion posture [31]. However, given that the drop foot posture is the main cause of decreased MTC and compensatory movements, we simplified the restriction to dorsiflexion alone. The spring coefficient was 10.39 N/mm, and the nominal length was set to an angle of at least 10° with respect to the plantarflexion posture. A person wearing this orthosis must exert more dorsiflexion torque in the swing phase than usual.

This orthosis only restricts the dorsiflexion movements, enabling us to observe the overall gait effects from it. Corresponding to the strengths of individuals, different MTC levels decreased and compensatory movements in either the coronal plane or sagittal plane illustrated in Fig. 5 were anticipated.

D. Participants

Nine healthy young individuals without gait disabilities participated in the experiment, and their personal information is displayed in Table 1. In this experiment, assistance of our high-dorsiflexion assistive system was conducted while

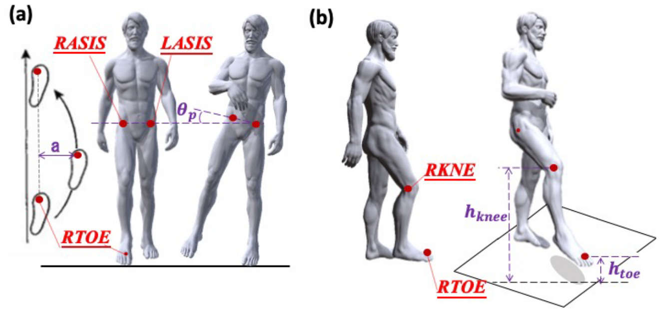

Fig. 5. (a) Compensatory movements in the coronal plane, evaluated with swing width a and lateral pelvis tilt angle θ_p; (b) compensatory movements in the sagittal plane, evaluated with maximum knee height in swing phase h_{knee}; minimum toe clearance evaluated with maximum toe height in swing phase h_{toe}.

TABLE I
PARTICIPANT INFORMATION

	Sex	Age (year)	Height (cm)	Weight (kg)
A	Male	31	172	74
B	Female	23	163	53
C	Male	23	177	70
D	Male	24	173	60
E	Female	24	167	57
F	Male	25	170	65
G	Male	23	179	64
H	Female	23	166.5	49
I	Male	25	175	95

the dorsiflexion responses of participants during the swing phase were restricted simultaneously. With our system's large assistive torque, even a participant wearing the dorsiflexion-restrictive device could still be assisted to the predefined dorsiflexion angles. This helped identify trends prior to implementation in patients.

E. Experimental Design

In this experiment, all participants were assumed to suffer from right-side paralysis. They were requested to walk 5 m 6 times at their own preferred speed in the following states:

1. Normal walk (NOR)

The participants wore the device during walking, but the dorsiflexion-resisting tension spring was not installed.

2. Simulated compensatory gait (SCG)

The participants wore the device during walking, and the spring was installed to restrict dorsiflexion movement.

3. Assistance without delay (AST0)

The participants simulated the compensatory gait, and the high-dorsiflexion assistive system supported dorsiflexion upon toe-off events, which were detected without delay.

4-9. Assistance based on the Δt delay (ASTΔt)

A Δt delay (in ms) after toe-off detection was set for activating the dorsiflexion assistance. Δt took the following values: 50, 100, 150, 200, 250, and 300 ms. That is, the conditions 4 to 9 are AST50, AST100, AST150, AST200, AST250, and AST300.

The latest delay condition AST300 was set in anticipation of observations of obvious differences in this feasibility study. As displayed in Fig. 3, immediate dorsiflexion assistance of the toe-off event AST0 was in fact an early assistance before the real transition to the swing phase of the gait. This timing difference among individual gait patterns was approximately 100 ms and was in agreement with our previous research [22]. The gait cycle time for a healthy person is approximately 1.3 s [32], and the time elapsed for swing phase dorsiflexion is equal to 20% of the gait period [33], that is, approximately 260 ms. Considering the early detection of the toe-off event and variation of gait cycle, we set a 300 ms delay as the longest time delay of assistance, with 50-ms intervals for shorter time delay conditions.

Ethical approval for this study was granted by the Ethics Committee of Waseda University (approval number: 2017-044).

F. Evaluations Indices

The experimental evaluation included three aspects: the extent of voluntary dorsiflexion efforts during the swing phase, risk of stumbling, and extent of compensatory movements, as summarized in Fig. 5.

The extent of voluntary dorsiflexion efforts was evaluated by means of sEMG of the tibialis anterior. The data were collected using the Trigno^{TM} Wireless System (Delsys Inc., USA). Signals were sampled at a rate of 2000 Hz. The collected data were analyzed using the software EMGworks (Delsys Inc., USA). According to our application of analyzing tibialis anterior sEMG signal, a fourth order Butterworth bandpass filter was used to filter the data, and the cutoff frequencies were set to values between 20 and 450 Hz. The envelope of the filtered signals were derived using root-mean-square (RMS) values with a 300 ms movable window [34], [35]. This evaluation index was derived based on the maximum percentage of amplitude analysis of processed data referencing the maximum voluntary contraction of the tibialis anterior during ankle dorsiflexion in the swing phase.

The risk of both stumbling and compensatory movements was evaluated using the motion capture system (MAC3D, NAC Inc. Japan), which consisted of eight high-speed cameras with the sampling rate set to 200 Hz. The recorded data were low-pass filtered with a cutoff frequency set to 6 Hz. The markers related to evaluation indices in this study are as follows:

1) RTOE: Between the first and second metatarsal bones
2) RKNE: Side of the knee joint
3) RASIS: Right anterior superior iliac spine
4) LASIS: Left anterior superior iliac spine

The stumbling risk was evaluated using MTC, which was determined by the minimum height of the RTOE marker h_{toe} in the swing phase subtracted by that in standing posture. The compensatory movements included those in the coronal and sagittal planes, referencing two typical pathological gaits of patients with stroke, circumduction and steppage, respectively [4], [5]. For the former, compensatory gait patterns of hip hiking and circumduction were exhibited, as depicted in Fig. 5(a). Hip hiking was evaluated with the lateral pelvis tilt angle θ_p and was calculated with respect to the angle between the floor and vector from the markers, LASIS to RASIS. Circumduction was evaluated with the swing width a defined as the horizontal length of marker RTOE in the swing phase. For the latter, an increase in the knee height was a representative characteristic of the steppage gait, as depicted

TABLE II
RESULTS OF PARTICIPANTS AS FUNCTIONS OF UTILIZED COMPENSATORY MOVEMENTS

Participant	Condition	Maximum knee height (mm)	Lateral pelvis tilt angle (°)	Swing width (mm)	Utilized compensatory gait
A	NOR[1)]	533.58 ± 1.38	2.85 ± 0.29	12.43 ± 2.08	Circumduction
	SCG[2)]	533.89 ± 2.10	6.32 ± 0.55	98.59 ± 11.45	
	NOR-SCG[3)]	p = .790	p < .001*	p < .001*	
B	NOR	550.75 ± 4.68	2.91 ± 0.72	33.23 ± 8.20	Circumduction
	SCG	544.21 ± 4.75	7.65 ± 0.83	144.93 ± 20.37	
	NOR-SCG	p = .037[(4]	p < .001*	p < .001*	
C	NOR	571.43 ± 5.56	1.42 ± 0.20	24.31 ± 5.77	Steppage
	SCG	583.99 ± 3.39	1.13 ± 0.45	26.56 ± 6.37	
	NOR-SCG	p = .002*	p = .175	p = .534	
D	NOR	522.35 ± 2.98	1.64 ± 0.44	19.37 ± 8.09	Steppage
	SCG	532.56 ± 4.24	2.45 ± 1.19	21.50 ± 4.66	
	NOR-SCG	p = .012*	p = .150	p = .589	
E	NOR	546.03 ± 4.30	4.29 ± 0.62	33.28 ± 10.92	Steppage
	SCG	572.57 ± 10.01	3.77 ± 0.48	34.02 ± 13.92	
	NOR-SCG	p < .001*	p = .135	p = .921	
F	NOR	547.17 ± 2.23	0.02 ± 0.25	24.91 ± 5.20	Steppage
	SCG	568.90 ± 5.79	0.84 ± 0.40	15.18 ± 5.20	
	NOR-SCG	p < .001*	p = .003[(5]	p = .015[(6]	
G	NOR	576.99 ±2.23	1.63 ± 0.29	27.23 ± 7.38	Steppage
	SCG	589.71 ± 3.70	2.83 ± 0.43	30.51 ± 11.12	
	NOR-SCG	p = .001*	p = .003[(5]	p = .641	
H	NOR	558.50 ± 3.55	-1.47 ± 0.75	38.11 ± 11.07	Circumduction + Steppage
	SCG	577.57 ± 4.87	9.04 ± 0.96	222.10 ± 27.62	
	NOR-SCG	p < .001*	p < .001*	p < .001*	
I	NOR	553.48 ± 2.39	0.773 ± 0.25	25.25 ± 7.24	Circumduction + Steppage
	SCG	564.28 ± 5.21	6.86 ± 1.62	243.43 ± 19.33	
	NOR-SCG	p < .001*	p < .001*	p < .001*	

*: p < .05, significant difference observed with Student's t-test
[1)]: NOR: Normal walk
[2)]: SCG: Simulated compensatory
[3)]: The significant difference between NOR and SCG conditions
[4)]: Significant decrease in maximum knee height in swing phase. It was not considered to that steppage gait was utilized
[5)]: Significant increase in lateral pelvis tilt angle, but the difference is smaller than 3°. It was not considered that circumduction gait was utilized
[6)]: Significant decrease in swing width, not determined as utilizing circumduction gait

in Fig. 5(b). Evaluation was conducted with the maximum RKNE marker's height h_{knee} in the swing phase.

G. Statistical Analysis

Given the expectation of different types and degrees of SCGs in participants, we opted to evaluate the results according to the individuals. To verify the compensatory movements used by each participant, the Student's t-test was applied on maximum knee height in the swing phase, lateral pelvis tilt angle, swing width between NOR, and SCG conditions with significant differences set as $p < 0.05$. A significant increase in maximum knee height for SCG is determined as utilization of steppage gait. In addition to fulfillment of significance level, we determined that a participant who used circumduction gait either increased the lateral pelvis tilt angle to values of $>3°$ or increased the swing width to values of >15 mm [36], [37].

For conditions from AST0 to AST 300, we applied Pearson correlation coefficient on timing of delay assistance and all evaluation indices (processed sEMG, MTC, maximum knee height in swing phase, lateral pelvic tilt, and swing width). Although conditions were set at 50-ms intervals, we believe that the interval is short as the gait cycle and Pearson correlation were appropriate to verify our assumption related to dorsiflexion assistance timing. Referencing a guide to correlation coefficients [38], we interpreted the strength of correlation r to be strong if $0.7 \leq |r| < 1$, moderate if $0.5 \leq |r| < 0.7$, and poor if $0.0 < |r| < 0.5$. The significance level of the correlation was set as $p < 0.05$. The analyses were conducted with SPSS (version 27). We expected that processed sEMG and used compensatory movements have at least positively moderate correlation with the dorsiflexion assistance delay time, while MTC has at least negatively moderate correlation with the delay time.

III. RESULTS

A. Used Compensatory Movements

Table II lists the results associated with the utilization of compensatory movements for all participants. Participants A and B used circumduction gait with significantly increased lateral pelvic tilt and swing width in the swing phase. Participants C to G used steppage gait with significantly increased knee height in the swing phase. Note that participants F and G's lateral pelvic tilt were significantly larger in SCG than in NOR. However, the average differences did not exceed 3°. Thus, we did not consider them as cases which used the compensatory movement. Participants H and I used all the compensatory movements, so they were determined as cases which used both circumduction and steppage gaits.

B. Correlation Between Tibialis Anterior sEMG and Assistance Delay Time

According to Tables III–V, processed sEMG of tibialis anterior for participants A (r = .802; p < .001), C (r = .733; p < .001), D (r = .831; p < .001), E (r = .733; p < .001), and H (r = .922; p < .001) showed positive, strong correlations with assistance delay time. Participant B's sEMG

TABLE III

CORRELATIONS BETWEEN EVALUATION INDICES AND ASSISTANCE DELAY TIME FOR PARTICIPANTS BASED ON THE UTILIZATION OF CIRCUMDUCTION GAIT

Participant	Condition	Amplitude of sEMG[1] (%)	MTC[2] (mm)	Maximum knee height (mm)	Lateral pelvis tilt angle (°)	Swing width (mm)
A	AST0	7.34 ± 1.01	22.19 ± 1.94	527.81 ± 2.63	3.28 ± 0.33	15.70 ± 4.17
	AST50	8.95 ± 1.03	20.13 ± 1.72	526.33 ± 1.64	3.26 ± 0.70	28.28 ± 7.30
	AST100	9.98 ± 1.67	20.29 ± 2.34	527.36 ± 2.76	3.01 ± 0.68	41.30 ± 10.04
	AST150	9.67 ± 1.36	20.14 ± 2.76	527.89 ± 1.97	5.21 ± 0.45	68.79 ± 11.84
	AST200	10.60 ± 1.29	20.84 ± 2.85	527.74 ± 2.62	5.38 ± 0.31	78.49 ± 10.47
	AST250	13.25 ± 1.20	21.30 ± 2.37	526.55 ± 1.04	5.39 ± 0.32	92.97 ± 9.60
	AST300	20.58 ± 3.60	29.96 ± 2.85	533.41 ±1.71	5.63 ± 0.72	107.74 ± 4.47
	Correlation	r = .802**, p <.001	r = .467, p < .001	r = .427, p = .003	r = 809**, p < .001	r = .965**, p < .001
B	AST0	12.52 ± 2.69	47.09 ± 3.41	558.02 ± 4.06	2.87 ± 0.45	22.17 ± 9.30
	AST50	12.62 ± 3.90	35.36 ± 3.91	543.45 ± 6.04	2.41 ± 0.32	40.33 ± 16.92
	AST100	14.26 ± 1.68	37.38 ± 6.20	541.32 ± 2.92	3.29 ± 0.96	93.90 ± 28.07
	AST150	13.66 ± 2.34	43.70 ± 5.26	547.33 ± 1.82	5.22 ± 1.04	105.63 ± 8.79
	AST200	14.77 ± 1.83	41.15 ± 4.16	539.50 ± 3.59	5.15 ± 0.39	109.72 ± 26.96
	AST250	17.03 ± 2.42	44.68 ± 3.42	534.10 ± 3.76	5.88 ± 1.48	152.26 ± 29.05
	AST300	17.69 ± 2.44	38.13 ± 5.97	548.16 ± 6.66	6.15 ± 1.01	141.33 ± 22.30
	Correlation	r = .598*, p <. 001	r = −.055, p = .732	r = −.437, p = .004	r = .814**, p < .001	r = .874**, p < .001

**: $0.7 \le |r| < 1$, strongly correlated; *: $0.5 \le |r| < 0.7$, moderately correlated
[1] sEMG: surface electromyography
[2] MTC: minimum toe clearance

TABLE IV

CORRELATIONS BETWEEN EVALUATION INDICES AND ASSISTANCE DELAY TIME FOR PARTICIPANTS BASED ON THE UTILIZATION OF STEPPAGE GAIT

Participant	Condition	Amplitude of sEMG[1] (%)	MTC[2] (mm)	Maximum knee height (mm)	Lateral pelvis tilt angle (°)	Swing width (mm)
C	AST0	9.78 ± 2.45	23.50 ± 2.92	N/A[3]	0.65 ± 0.28	12.78 ± 1.87
	AST50	37.98 ± 10.99	11.33 ± 2.08	563.77 ± 1.82	1.08 ± 0.49	20.46 ± 6.73
	AST100	49.11 ± 10.30	17.65 ± 3.02	569.80 ± 3.69	0.95 ± 0.15	12.92 ± 0.65
	AST150	54.72 ± 9.81	22.61 ± 2.65	570.08 ± 3.02	1.01 ± 0.27	12.05 ± 2.66
	AST200	54.69 ± 11.46	16.55 ± 2.31	568.05 ± 1.82	1.24 ± 0.26	26.55 ± 10.89
	AST250	61.62 ± 14.54	19.49 ± 4.02	572.05 ± 3.09	1.31 ± 0.28	17.49 ± 7.86
	AST300	56.54 ± 12.61	25.17 ± 4.41	576.05 ± 1.68	1.12 ± 0.30	15.60 ± 6.20
	Correlation	r = .733**, p < .001	r = .274, p = .079	r = .727**, p < .001	r = .449, p = .003	r = .154, p = .330
D	AST0	19.67 ± 5.37	8.67 ± 4.02	504.33 ± 4.24	1.72 ± 0.78	15.30 ± 3.01
	AST50	22.59 ± 5.12	8.56 ± 3.12	504.37 ± 1.89	2.38 ± 0.22	20.30 ± 5.32
	AST100	24.63 ± 3.73	12.11 ± 2.47	508.04 ± 3.01	2.65 ± 0.53	22.68 ± 13.93
	AST150	24.66 ± 3.36	16.04 ± 2.76	506.52 ± 1.51	2.40 ± 0.15	19.34 ± 3.24
	AST200	36.33 ± 6.17	13.88 ± 4.71	511.86 ± 5.46	1.99 ± 0.43	19.05 ± 5.37
	AST250	40.72 ± 3.79	14.06 ± 2.04	513.98 ± 3.16	2.59 ± 0.90	40.85 ± 9.78
	AST300	38.11 ± 1.48	18.21 ± 6.08	519.48 ± 6.33	2.62 ± 0.65	21.81 ± 9.99
	Correlation	r = .831**, p < .001	r = .614*, p < .001	r = .772**, p < .001	r = .290, p = .066	r = .389, p = .012
E	AST0	31.47 ± 3.52	34.84 ± 3.66	553.75 ± 3.33	4.73 ± 0.67	36.83 ±28.80
	AST50	20.58 ± 6.77	32.44 ± 3.53	551.33 ± 1.95	5.56 ± 0.73	34.91 ± 14.48
	AST100	27.10 ± 5.68	33.07 ± 6.52	551.13 ± 5.37	4.71 ± 0.87	39.51 ± 14.88
	AST150	38.90 ± 12.94	28.59 ± 4.99	551.24 ± 4.67	4.06 ± 0.70	54.33 ± 9.19
	AST200	52.50 ± 6.95	43.41 ± 5.69	557.02 ± 8.17	4.87 ± 1.34	35.84 ± 32.72
	AST250	50.29 ± 5.00	38.87 ± 6.06	559.74 ± 3.19	4.99 ± 0.47	42.23 ± 20.45
	AST300	54.64 ± 9.11	40.42 ± 9.13	569.53 ± 10.95	5.44 ± 1.03	32.22 ± 12.08
	Correlation	r = .773**, p < .001	r = .404, p = .009	r = .626*, p < .001	r = .079, p = .622	r = −.023, p = .844
F	AST0	31.17 ± 6.84	16.65 ± 3.22	548.80 ± 4.59	2.09 ± 0.37	17.82 ± 5.26
	AST50	29.94 ± 5.25	13.45 ± 4.56	548.33 ± 2.10	1.97 ± 0.42	26.04 ± 4.48
	AST100	39.28 ± 3.41	18.28 ± 3.15	548.33 ± 4.29	0.88 ± 0.78	34.38 ± 3.15
	AST150	60.13 ± 8.04	13.46 ± 3.12	545.79 ± 3.05	1.94 ± 0.25	21.16 ± 5.78
	AST200	33.82 ± 4.82	17.41 ± 1.37	550.43 ± 1.99	1.84 ± 0.90	23.30 ± 5.56
	AST250	29.07 ± 9.25	16.40 ± 2.05	548.11 ± 0.88	2.07 ± 0.60	20.91 ± 2.10
	AST300	34.14 ± 3.86	10.51 ± 4.04	553.08 ± 1.15	1.97 ± 0.44	23.22 ± 5.17
	Correlation	r = −.002, p = .989	r = −.245, p = .121	r = .318, p = .071	r = .087, p = .605	r = −.065, p = .696
G	AST0	21.47 ± 1.20	12.18 ± 2.77	566.46 ± 0.71	6.51 ± 0.55	21.83 ± 6.39
	AST50	22.47 ± 2.20	31.26 ± 3.15	572.79 ± 1.78	4.12 ± 0.52	13.45 ± 1.27
	AST100	24.76 ± 2.88	21.19 ± 2.73	572.12 ± 0.15	2.21 ± 0.62	18.58 ± 3.81
	AST150	26.10 ± 5.25	15.85 ± 3.96	572.28 ± 2.89	2.50 ± 0.51	18.17 ± 3.66
	AST200	22.92 ± 3.81	15.85 ± 1.86	572.29 ± 3.26	2.78 ± 0.30	17.89 ± 5.95
	AST250	27.29 ± 1.78	20.34 ± 5.67	577.52 ± 2.24	3.07 ± 0.48	16.76 ± 4.69
	AST300	19.80 ± 1.70	15.80 ± 1.70	582.91 ± 3.52	5.89 ± 3.00	13.48 ± 5.11
	Correlation	r = .056, p = .705	r = −.180, p = .359	r = .858**, p < .001	r = −.127, p = .521	r = −.279, p = .151

**: $0.7 \le |r| < 1$, strongly correlated; *: $0.5 \le |r| < 0.7$, moderately correlated
[1] sEMG: surface electromyography
[2] MTC: minimum toe clearance
[3]: Data not applicable owing to the failure to detect the RKNE marker

TABLE V

CORRELATIONS BETWEEN EVALUATION INDICES AND ASSISTANCE DELAY TIME FOR PARTICIPANTS WHO UTILIZED CIRCUMDUCTION AND STEPPAGE GAITS

Participant	Condition	Amplitude of sEMG[1] (%)	MTC[2] (mm)	Maximum knee height (mm)	Lateral pelvis tilt angle (°)	Swing width (mm)
H	AST0	26.92 ± 1.98	39.55 ± 7.51	577.90 ± 6.66	-0.52 ± 0.90	54.25 ± 9.28
	AST50	35.93 ± 8.77	58.81 ± 4.66	594.19 ± 4.46	-1.65 ± 0.86	40.78 ± 12.98
	AST100	38.23 ± 4.06	51.03 ± 9.20	586.59 ± 8.48	1.40 ± 1.12	47.89 ± 10.21
	AST150	45.15 ± 10.21	51.52 ± 11.62	587.10 ± 12.72	2.94 ± 1.26	49.51 ± 14.07
	AST200	62.31 ± 4.80	61.37 ± 3.98	594.39 ± 5.00	-0.21 ± 1.48	38.64 ± 16.68
	AST250	65.26 ± 6.83	55.37 ± 7.42	587.02 ± 5.38	-0.51 ± 1.26	113.44 ± 20.33
	AST300	77.02 ± 6.41	46.46 ± 7.63	588.63 ± 7.36	4.17 ± 1.24	102.84 ± 21.06
	Correlation	$r = .922^{**}, p < .001$	$r = .127, p = .436$	$r = .172, p = .288$	$r = .477, p = .002$	$r = .649^{*}, p < .001$
I	AST0	29.63 ± 9.41	27.46 ± 7.65	549.56 ± 12.83	6.26 ± 0.57	22.21 ± 4.76
	AST50	32.50 ± 2.96	37.27 ± 6.15	554.66 ± 4.49	6.27 ± 0.56	19.87 ± 4.95
	AST100	28.43 ± 2.99	32.61 ± 3.44	554.75 ± 12.96	6.26 ± 0.67	23.44 ± 7.00
	AST150	26.14 ± 3.21	41.18 ± 12.44	553.79 ± 8.52	5.85 ± 0.22	22.99 ± 6.44
	AST200	28.65 ± 7.86	39.70 ± 3.98	547.19 ± 4.05	7.10 ± 1.01	162.02 ± 10.01
	AST250	27.83 ± 4.19	37.29 ± 1.14	540.08 ± 2.34	7.56 ± 0.33	156.78 ± 24.14
	AST300	26.37 ± 5.68	41.08 ± 4.72	540.51 ± 5.20	8.01 ± 0.90	198.04 ± 2.02
	Correlation	$r = -.246, p = .112$	$r = .461, p = .005$	$r = .435, p = .010$	$r = .627^{*}, p < .001$	$r = .857^{**}, p < .001$

**: $0.7 \leq |r| < 1$, strongly correlated; *: $0.5 \leq |r| < 0.7$, moderately correlated

[1] sEMG: surface electromyography

[2] MTC: minimum toe clearance

also showed positively moderate correlation with the delayed time ($r = .598$; $p < .001$). The results of the previous participants fulfilled our assumption. Conversely, processed sEMG and delayed time were poorly correlated for participants F ($r = -.002$; $p = .989$), G ($r = -.008$; $p = .961$), and I ($r = -.246$; $p = .112$).

C. Correlation Between MTC and Assistance Delay Time

The MTC results totally differed from our assumptions. Even a negatively moderate correlation could not be observed between MTC and the assistance delay time. Conversely, as shown in Table IV, the MTC of Participant D even showed a positively moderate correlation with the delay time ($r = .614$; $p < .001$).

D. Correlation Between Compensatory Movements and Assistance Delay Time

Table III shows results from participants who used the circumduction gait, Participant A's lateral pelvic tilt angle ($r = .809$; $p < .001$) and swing width ($r = .965$; $p < .001$) were positively strongly correlated with assistance delay time. Participant B's lateral pelvic tilt angle ($r = .814$; $p < .001$) and swing width ($r = .874$; $p < .001$) also showed the same results. Poor correlation was found between assistance delay times and maximum knee heights, so circumduction related movements apparently dominated the assistance effects. The results indicated that a longer assistance delay time could result in an increase in the extent of circumduction gait, which fulfilled our assumption.

Table IV shows results of participants who used steppage gait. Participants C ($r = .727$; $p < .001$), D ($r = .772$; $p < .001$), and H ($r = .858$; $p < .001$) showed positively strong correlation between their maximum knee height in swing phase and delayed assistance time. Participant E also showed a moderate correlation ($r = .626$; $p < .001$). Note that the data from Participant C were not applicable due to failure in detecting the RKNE marker. However, knee height results for other conditions should be sufficient to perform correlation. Meanwhile, poor correlations were observed for all the circumductions associated with related indices and assistance delay time pertaining to these participants. These results indicate that a longer assistance delay time could result in an increase in the extent of steppage gait, which fulfills our assumption. Conversely, no positively strong or moderate correlation was observed between the assistance delay time and maximum knee height of Participant F.

Table V shows the results of participants using both circulation and steppage gaits. Participant H's swing width and assistance delay time were positively moderately correlated ($r = .649$; $p < .001$). For Participant I, there was a positively strong correlation between the swing width and delay time ($r = .857$; $p < .001$) and a positively moderate correlation between lateral pelvic tilt angle and assistance delay time ($r = .627$; $p < .001$). However, only poor correlations were found between their maximum knee heights and assistance time delay. Although the participants used compensatory movements at both coronal and sagittal planes, effects of circumduction dominated the effects with assistance timing. The indices related circumduction movements still fulfills our assumption.

IV. DISCUSSION

In this study, we have verified the effects of different dorsiflexion assistance timings on tibialis anterior sEMG, MTC, and compensatory movements. The results showed positively strong or moderate correlations between tibialis anterior sEMG and dorsiflexion assistance delay times in six out of nine patients. Meanwhile, the results showed positively strong or moderate correlations between extents of compensatory movements and dorsiflexion assistance delay times. These results largely support our hypothesis.

A. Voluntary Dorsiflexion Movement

Positively strong or moderate correlations were observed in the tibialis anterior sEMG in six out of the nine participants'

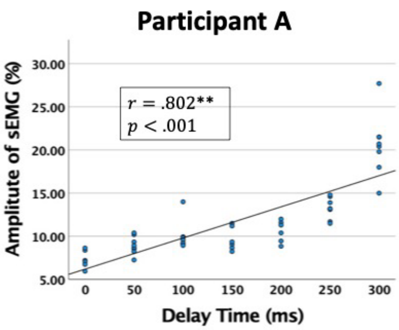

Fig. 6. Correlation of processed sEMG for participant A.

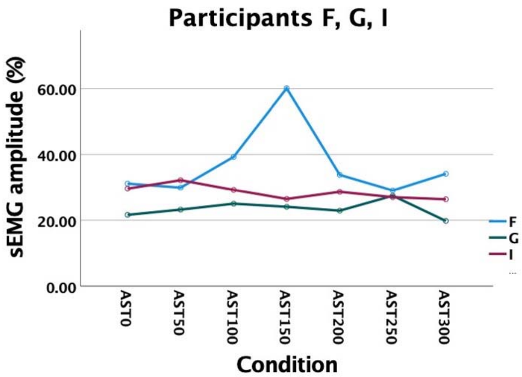

Fig. 7. Average of processed sEMG for participant F, G and I.

regarding different delay times for dorsiflexion assistance. Fig. 6 shows that with longer delay times, such as the case of Participant A, more dorsiflexion movements tended to be performed. This indicates that adjusting the assistance time for a powered device could be treated as tuning the level of assistance. However, poor correlation results were also found in the cases of participants F, G, and I. Fig. 7 shows their sEMG results apparently did not increase with longer assistance delay time. In fact, a review article has pointed out that progressively increased resistance for patients in training is important for patients with stroke [39]. Although the participants for this study were healthy individuals, it is understandable that they may also be subconsciously unwilling to exert voluntary dorsiflexion against the dorsiflexion-resistive device when our system's assistance was late.

B. Increase in MTC Results With Exaggerated Compensatory Movements

For all participants, no negatively strong or moderate results were observed for all participants. Although it seems logical that decreases in dorsiflexion assistance would result in decreased MTC [40], research has also shown that a patient's compensatory movements could adequate to maintain sufficient MTC [23]. The ankle movements of healthy participants were restricted, but the movements of other joints, such as the knees and hips, were not restricted. Therefore, it is reasonable that healthy participants have the abilities to prevent decrease of MTC with compensatory movements. The result of one participant's MTC (Participant D) even shows positively moderate correlation with assistance delay time possibly due to exaggeration of the compensatory movements.

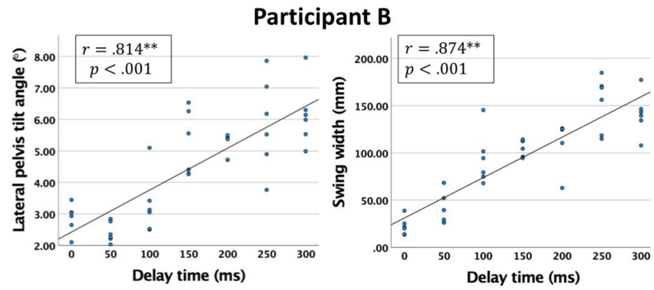

Fig. 8. Correlations of lateral pelvis tilt angles and swing phase for participant B.

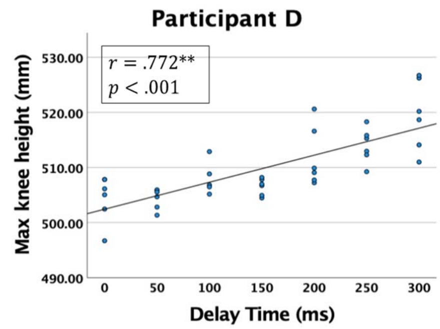

Fig. 9. Correlations of maximum knee height for participant D.

C. Compensatory Movements

With the dorsiflexion-restricted device, different types and extents of compensatory movements were realized by only limiting participants' dorsiflexion movements in the swing phase of gait. For Participants A–G, all of their used compensatory movements showed positively strong/moderate correlations with assistance delay time of our high-dorsiflexion assistive system except from Participant F. Fig. 8 shows examples of Participant B's circumduction-related results, and Fig. 9 shows Participant D's steppage-related result. In fact, on observing Participant F's maximum knee height data in the swing phase in Table IV, we found that the average of maximum knee height in AST300 abruptly increased compared with the other conditions. Considering different cadence and speed values for different individuals, we speculated that the steppage for this participant became observable after delay time was set longer than 250 ms. This result could still reveal utilization of compensatory movement with late dorsiflexion assistance. For Participants H and I, they both performed pseudo circumduction and steppage gaits. Both of their results also show positively strong/moderate correlation between only circumduction related movements and assistance delay time. Whether this result could relate to a real patient outcome requires future study. Even still, the results show increased extents of compensatory movements, with the dorsiflexion assistance time being prolonged.

In brief, the experiments results highlighted a potential feature that indicated that with longer assistance delay times of our high-dorsiflexion assistive system, the extent of compensatory movements would also increase. Moreover, the extents of compensatory movements in the same condition differed among participants. An example related to the swing widths of participants A, H, and I is shown in Fig. 10. An abrupt increase

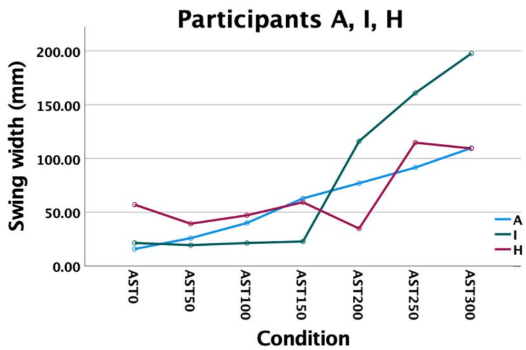

Fig. 10. Swing width variations for participants A, H, and I.

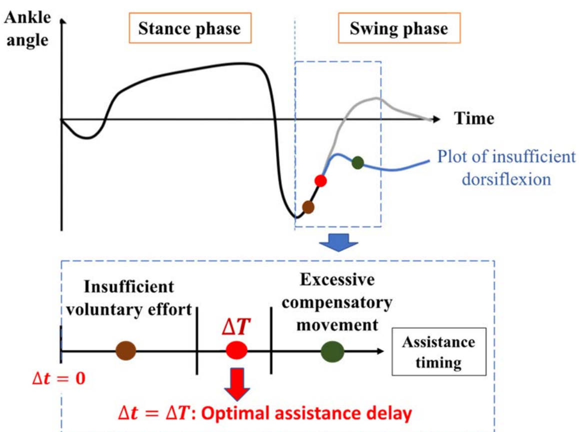

Fig. 11. Concept for optimal dorsiflexion assistance timing that maximizes voluntary efforts while minimizing compensatory movements.

in swing width was observed from AST200 to AST250 in the case of Participant H, whereas in the case of Participant I, the increase occurred from AST150 to AST200. However, in the case of Participant A, a tendency of continuous increase in swing width with later dorsiflexion assistance timing was observed. Selecting suitable assistance timing for ankle dorsiflexion should focus on referencing or suppressing the used compensatory movements to acceptable extents for individuals.

D. Modifying Assistance of Delay Time as a Modifying Assistance Level

According to the previous statements, we consider tuning assistance timing as an option for the modification of the assistance level. It is intuitive that by tuning the force, torque, assisted target angle of joint or assisted velocity, and required voluntary efforts from the user and his/her gait pattern should be affected. With the same concept, we believe that it is reasonable to regard the modification of assistance timing of an event-triggered control as a potential option of the modification of the assistance level during rehabilitation session.

This study sought to assess the effects on voluntary efforts and gait kinematics resulting from different assistance timings for ankle dorsiflexion. The ultimate goal, however, is to apply these findings to a rehabilitation protocol. A suitable setting for the activation timing for dorsiflexion assistance should differ among different patients considering their recovery conditions and gait characteristics. Based on the findings of this study we believe that this perfect help timing will allow viable voluntary efforts while suppressing compensatory movements to an acceptable level, as shown in Fig. 11.

E. Clinical Implications

Many patients with stroke have gait disturbances, and dorsiflexion movements of their feet are reduced during walking. Therefore, it is important to consider the muscle activity involved in dorsiflexion during the recovery process. In this study, we demonstrated the possibility of altering muscle activity and compensatory movements by changing the timing of dorsiflexion in healthy subjects. Given that patients with stroke have various gait patterns, it may be possible to control muscle activity and compensatory movements by controlling the timing of dorsiflexion for each patient in the future.

F. Limitations and Recommendations

This study's limitations include the use of healthy persons as subjects and small sample size. The experiment performed with healthy volunteers allows us to safely assess the impact of ankle dorsiflexion restriction on circumduction and steppage gaits. The duration of aid delay was shown to be associated with the tibialis anterior sEMG and the number of compensatory movements. Thus, we believe that our findings regarding the voluntary efforts and compensatory movements under various support timings can be applied to most patients. Future intervention studies should be undertaken to assess the impact on real patients, considering more individuals with problematic gait patterns because of low muscle strength, low balancing ability, perceptive paralysis, and recovery status. Assistance timing circumstances should be fine-tuned to ensure safe and effective clinical trials.

This study's experiments were conducted by manually modifying the predefined assistance delay time. Future research studies should also consider an adaptive control strategy using the highly suitable assistance timings for a patient's recovery status and gait characteristics. In addition, as aforementioned, decreasing assistance level could only ensure room for self-efforts. An improved PAFO system with a training protocol that motivates and facilitates ankle movements would also be considered as part of future research endeavors.

V. Conclusion

In this feasibility study, we verified the effects of different timings for assisted ankle dorsiflexion that used our high-dorsiflexion assistive system. Nine participants emulated compensatory movements while their capacities for ankle dorsiflexion was constrained. The results showed longer dorsiflexion assistance delay times, and a trend was observed whereby the extents of their compensatory movements increased; in most of the participants, their voluntary efforts associated with dorsiflexion movements increased. This indicates the potential of utilizing the adjustment of delayed timings as an adaptive control method to ensure voluntary efforts. Future research includes evaluating the effects of different assistance timings on real stroke survivors and developing a strategy that facilitates voluntary efforts and ensures optimal gait.

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