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# Functional Resistance Training With Gait Phase-Dependent Control Using a Robotic Walker: A Pilot Study

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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 the School of Engineering, Nagoya University (under Application No. 21-10), and performed in line with the Declaration of Helsinki.

**ABSTRACT** Strength training can contribute to the improvement of the lower limb muscles, which does not always transfer to the improvement of walking ability. It results from the lack of training for the muscles which are responsible to generate propulsion force during walking, such as plantar flexors and hip extensors. This study aims to examine and evaluate the effects of a gait phase-dependent control strategy on the kinematics and muscular activation of users and to assess whether it can contribute to enhancing the strength of lower limb muscles in a task-specific manner, especially for plantar flexors. Eight healthy, young, male subjects participated in our experiment. We recorded EMG data from the muscles of the lower and upper limbs during walking with a robotic walker under different conditions, as well as participants' lower limb joint angles, to investigate the effects of the gait phase-dependent control. We found that gait phase-dependent control can lead to a high activation level of plantar flexors and a high angular velocity during the pre-swing phase and that the activity in triceps brachii is lower than that with constant resistance. Furthermore, an experiment with 9 elderly subjects verified the training effects of gait phase-dependent control on the plantar flexors. Therefore, we conclude that the gait phase estimation method can train plantar flexors with high effectiveness and efficiency. The long-term effects of gait phase-dependent control on improving the walking ability of elderly people will be investigated in the future.

**INDEX TERMS** Functional resistance training, surface EMG, kinematics, rehabilitation robot, gait rehabilitation.

## I. INTRODUCTION

One in six of the world's population suffer from neurological disorders (Alzheimer's disease, Parkinson's disease, and stroke), according to a report issued by the World Health Organization [1]. Neurological disorders often lead to gait disorders, especially in the elderly; gait disorders can severely damage walking ability and quality of life [2], [3].

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Muscle weakness frequently occurs with neurological impairments [4], often caused by structural and functional changes in the neuromuscular system or muscle fibers. Furthermore, lower limb muscle strength is highly related to balance [5], risk of falling [6], and walking ability [7]. Strength training such as leg press can improve the lower limb muscle strength, but does not always result in the improvement of walking ability [8], [9]; task specificity is crucial for strength training to restore the walking ability of users [9]. It has been demonstrated that it is better to

improve walking ability by using a combination of strength training and functional training [8], a training termed functional resistance/strength training. Some researchers have attempted to utilize functional resistance/strength training by increasing the physical load during walking, these studies have revealed the effects of different types of physical load on the kinematics, kinetics, and muscular activations [10]–[12]. The most straightforward way of increasing the physical load of walking is by placing additional weight on subjects' ankle joints; experimental results with subjects who suffer from chronic hemiparesis show this process can improve the hip and knee joint power bursts [10]. Kubinski and Higginson used a weighted vest weighing 1/6th of the subject's body weight during walking and found the double stance phase duration significantly increased with the weighted vest. Other researchers have tried to utilize the external horizontal force applied at the center of mass of a user, and have found that the activation of lower limb muscles increases with resistance exerted to users, especially for the hip extensors [12].

Using robots to conduct rehabilitation of elderly people is becoming increasingly common as this can reduce the physical load for physical therapists (PTs) [13], [14]. Moreover, robots can automatically monitor and assess the status of users during the training process, which allows the PTs to adjust the training plans or intensity of training accordingly. Several robots have been adopted for strength training in a task-specific manner. Wearable robots are modified to exert resistance to the knee and/or hip joints during walking, to train muscles involved in coordinating the swing of the legs [15], [16]. The activation of knee flexors increased while the range of movement of the knee was restricted owing to the applied resistance [15]. Mun *et al.* [17] proposed an overground robotic walker for resistance training. The proposed walker is attached to the user's pelvis to provide resistance and support. The experimental results with young healthy subjects have shown that the device can contribute to stimulating the knee joint muscles and hip extensors.

Although the above-mentioned functional resistance training methods can stimulate parts of the lower limb muscles, few of them can contribute to improving the strength of the plantar flexors. It was demonstrated that the key factor for strength training to achieve improved walking performance is focused on the muscles which are responsible to generate propulsion during walking [9]. The lack of training for plantar flexors can limit the translation from muscle strength to walking ability because plantar flexors are the main power generators for propulsion during walking [18], [19]. Moreover, plantar flexors also contribute to providing vertical support and swing initialization [20]. Therefore, the muscle strength of the plantar flexors is important for walking speed, which is a critical indicator of mortality [21], [22]. Furthermore, the strength of the gastrocnemius (GAS) and soleus (SOL) is highly correlated with one's ability to balance [23]. A high amount of muscle torque of

plantar flexors is needed when people try to recover from a loss of balance, especially when falling forward [24]. Elderly people who have weak plantar flexors are more likely to fall in their daily lives [25], and the strength of plantar flexors substantially decreases with aging [26]. It is desired to develop a rehabilitation robot that can restore the strength of plantar flexors for elderly people to achieve better rehabilitation outcomes during walking training.

The wearable robots are only able to exert resistance to the knee and hip joints [15]; therefore, it is difficult to stimulate the muscles of the ankle joint. The over-ground training walker proposed by Mun *et al.* [17] exerts constant resistance when walking through a body support system attached to the users' pelvis, which does not stimulate the activation of the plantar flexors. Instead, the knee extensors and hip extensors exerted a higher force during walking. Locomotion by gait consists of a series of cyclic motions. Several functional objectives should be achieved within one gait cycle, which requires different activation patterns of muscle groups to maintain a stable and safe gait [27]. A constant external resistance on the pelvis of users changed the walking patterns [17] instead of stimulating plantar flexors which mainly activated at the late stance phase. It is evident from their kinematic data that knee joints are bent at the moment of heel strike. We propose a gait phase-dependent control strategy to regulate the level of resistance exerted to a user during walking, which allows the rehabilitation robots to exert resistance only at some gait phases so that the muscles for generating propulsion force can be trained effectively and efficiently [28]. In this paper, we will examine the effects of gait phase-dependent control on the kinematics and muscular activation of the user and assess the effectiveness and efficiency of the proposed method on functional strength training for lower limb muscles, especially on plantar flexors.

## II. METHOD

### A. ROBOTIC WALKER

The walker used in this study was developed by Panasonic Corp. [29] and the structure of which is shown in FIGURE 1. It takes the structure of the common four-wheel walker, which consists of a mobile platform with four casters (two motor-actuated rear casters and two front passive omnidirectional casters). A loop-shaped handle serves as an interactive interface between the user and the walker. A six-axis force sensor (WEF-6A500-10-RCD-B, WACOH-TECH Inc., Toyama, Japan) mounted under the handle was used to monitor the interaction force between the user and the walker, which was applied to estimate the gait parameters of the users. The force/torque sensor and the motor controller are connected to a computer, in which a customized program monitors the interaction force and regulates the level of resistance generated by the motors. The height of the handle can be adjusted to meet the demands of users of different heights.

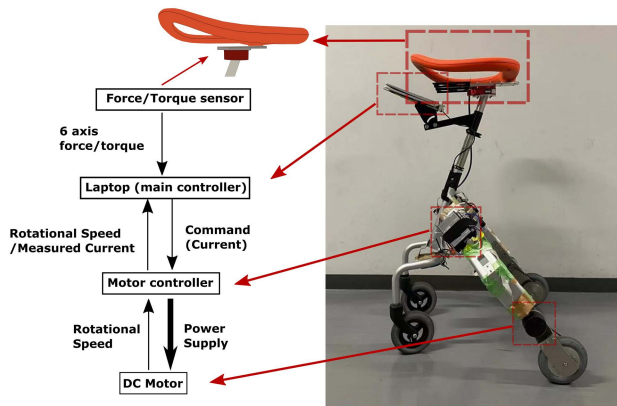


FIGURE 1. Robotic walker.

### B. GAIT PHASE-DEPENDENT CONTROL

In a gait cycle, some muscles generate braking force in the early stance phase, and some other muscles generate propulsion force in the late stance phase [30]. A constant external resistance along the horizontal direction can improve the activation of the muscles that are responsible for generating propulsion, though this may not be effective for training the muscles during braking. Additionally, resistance exerted during the wrong gait phase can change the gait patterns, which allows the muscle to generate propulsion during the braking phase. Thus, we try to exert resistance to users only when the target muscles are activated, while simultaneously releasing resistance when the muscles are at rest, to achieve effective and efficient training of the muscles in generating propulsion.

Gait phase-dependent control aims to train the muscles that are responsible for propulsion, especially the plantar flexors, namely the gastrocnemius (GAS) and soleus (SOL). The onset timing of the activations of the GAS and SOL is approximately 10% to 60% of the gait cycle, and the activation reaches a peak at 40% of the gait cycle [27]. In a previous study [28], we found that the duration of resistance was better set to 30%, rather than 20%, leading to a higher GAS activity during walking. In this study, we, therefore, set the duration of resistance as 30% of the gait cycle and changed the starting time of exerting resistance. We established three different start timing conditions for exerting resistance to participants, hereon mentioned as GPD1, GPD2, and GPD3:

- 1) 10%–40% of the gait cycle (GPD1)
- 2) 20%–50% of the gait cycle (GPD2)
- 3) 30%–60% of the gait cycle (GPD3)

The same pattern of gait phase-dependent resistance was imposed for both legs, so there were two time periods in one stride when external resistance was exerted to users. Taking the GPD1 condition as an example, resistance is exerted to users during the periods of 10% to 40% and 60% to 90% of the gait cycle. The resistance generated by motors under the constant resistance condition and a GPD condition are shown in FIGURE 2.

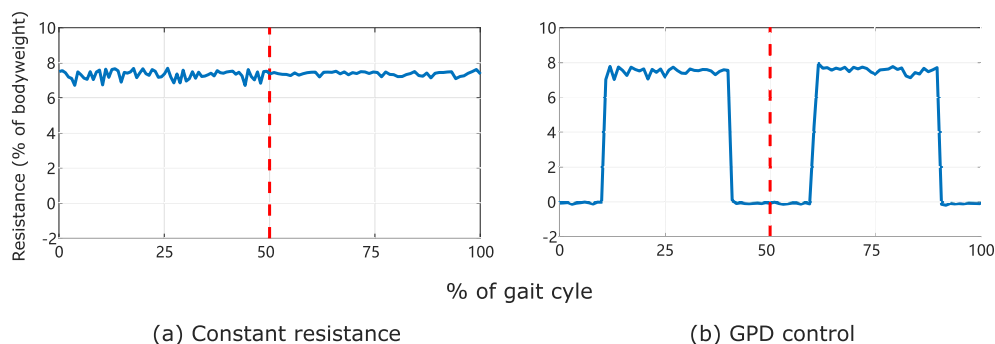
We found out that the GPD control elicited higher activation in the GAS in comparison with training with zero external resistance, which is similar to that with constant resistance (resistance remains the same throughout a gait cycle) in a previous study [28]. Furthermore, we also found muscle activity reduction in the rectus femoris and tibialis anterior in GPD conditions, in comparison with that of the constant resistance condition. In this study, we expect to find that the activity in plantar flexors in the GPD, and constant resistance conditions should be similar to each other. And we assume that the physical load on subjects in GPD conditions during walking should be lower than that in the constant resistance condition since the duration of resistance in GPD conditions is shorter.

To implement the proposed gait phase-dependent control, it is necessary to estimate the gait phase information in real-time. We applied a previously proposed online gait phase estimation method, which uses the interaction force between a user and a walker [31].

### C. EXPERIMENTAL PROTOCOL

In the experiment, muscular activation and kinematic data were monitored to assess the effectiveness and physical load of the proposed method. EMG data were measured using surface EMG bipolar sensors (Noraxon U.S.A. Inc., AZ, USA; Sampling rate: 1500 Hz) from eight lower limb muscles and an upper limb muscle: gastrocnemius lateralis (GAS), soleus (SOL), tibialis anterior (TA), biceps femoris (BF), semitendinosus (SEM), rectus femoris (RF), vastus lateralis (VL), gluteus maximus (GM), and triceps brachii lateralis (TRI). Participants' skin was shaved and prepared using alcohol before the sEMG electrodes were attached. All EMG electrodes were attached to the left legs of the participants, and the placement of the EMG electrodes was based on surface electromyography for the non-invasive assessment of muscles (SENIAM). Kinematic data (ankle, knee, and hip joint angles) were noted by a commercialized inertial measurement unit (IMU) system (Xsen, Enschede, Netherlands. Sampling rate: 60 Hz). Xsen is an IMU system with 23 IMU sensors that detect the acceleration, angular velocity, and orientation of 23 body segments (upper arms, lower arms, lower legs, upper legs, L5, S1, etc.). The data captured by the IMU units combined with a biomedical model can estimate the joint angles.

The study was approved by the ethics committee of the School of Engineering, Nagoya University (approval number: 21-10). Eight participants, who were all young male students (Age:  $25.0 \pm 2.1$  years. Height:  $176.4 \pm 4.9$  cm. Bodyweight:  $67.6 \pm 7.1$  kg), were recruited to join the experiment, and none of them suffered from lower limb impairments. Before the experiments, the participants were informed of the experimental protocol and signed a consent form. First, each subject walked with a walker with different levels of resistance (GPD and constant) for 10 min so that they could acclimate to walking with a walker. Each participant then walked with the walker for 30 m at a self-selected speed



**FIGURE 2.** Mechanical resistance in a gait cycle. Blue lines show the resistance generated by two motors in one gait cycle, and the level of resistance is 7.5% of the user’s bodyweight. Red dashed lines show the heel strike of another side of the leg.

for each trial with an upright posture and a stable cadence, they rested for 1 min between the two trials. They were instructed to walk twice in each condition, and the sequence of application of the different resistances was randomized. The following control strategies were applied to the walker: three types of GPD (GPD1, GPD2, and GPD3), constant resistance (Const), and no additional resistance (Zero) as the control group. No additional resistance means that the power of the robotic walker is turned off while the participant was walking. The resistance level in all conditions, except for the no additional resistance condition, was set at 7.5% of each subject’s body weight. In this paper, the level of external resistance means the resistance generated by two motors instead of the resistance exerted on users.

In the previous work [28], the resistance exerted on subjects was set at the level of 5% of bodyweight and no significant difference was found in upper limb muscle activation levels among all conditions. We assumed, based on the results, that the level of resistance was too low to induce a statistical difference in the upper limb activations. Therefore, we increased the level of resistance by 50% (7.5% of body weight) and 100% (10% of body weight). When the two levels of resistance were compared, the speed could not be obtained when the resistance was set at 10% of the bodyweight. The walker is designed to be a compact and light-weight training device for walking. The weight of the entire walker is about 20kg. The resistance at 10% is large enough that it causes the wheels of the walker to slip, which is not desired in walk training because it can lead to dangerous conditions. The slip on one side of the wheels can lead to unexpected movements of the robotic walker where it can collide with the user or cause the user to fall. Due to the aforementioned considerations, we set the resistance level at 7.5% of the bodyweight. In this experiment, the average resistance level was generated by two motors at  $7.46 \pm 0.22\%$  of body weight for all of the subjects under all conditions.

**D. EXPERIMENT WITH ELDERLY SUBJECTS**

The gait patterns of elderly people or those with gait disorders are less consistent than the gait pattern of younger/healthy

**TABLE 1.** Information of elderly subjects.

	Age	Height(cm)	Body weight(Kg)
Group 1	72 ± 1	163 ± 7	65.27 ± 14.84
Group 2	70 ± 3.61	161.33 ± 7.37	55.6 ± 15.15
Group 3	72.33 ± 4.65	164 ± 6.45	58.7 ± 14.21
Overall	71.44 ± 3.37	162.78 ± 5.37	59.86 ± 11.28

adults, which can lead to differences in the kinematics or muscular activation with rehabilitation intervention [32], [33]. These differences, even with the same resistance, are due to their weaker lower and upper limb strength. To evaluate the effectiveness of the proposed GPD control in the elderly, 9 participants (6 males, 3 females; all aged 66–77 years) were recruited. None of the subjects suffered from any neuromuscular conditions that could severely hinder their walking abilities. Moreover, they did not use any walking assistance device in their daily life. The participants were divided into three groups randomly, with each group consisting of two males and one female (TABLE 1).

This study was approved by the ethics committee of the School of Engineering, Nagoya University (approval number: 21-17). All the subjects were informed of the experimental protocol and signed a consent form before the experiment. In this experiment, the subjects were instructed to walk with the robotic walker in a circular corridor for 20 minutes at their preferred walking speed and in an upright posture. Different control strategies were applied to different groups. GPD1 control (with resistance set at 31 N) was applied for subjects of group 1; subjects of group 2 walked with constant resistance (resistance set at 20 N), and group 3 was a control group in which subjects walked with zero resistance. In contrast to the previous experiment, the same resistance was applied to all subjects within a group, irrespective of their body weight. This is because, unlike younger adults, the elderly do not necessarily have higher strength correlating to higher body weight. The level of resistance was set as 31 N for group 1 and 20 N for group 2 for two reasons. First, the level of resistance is reduced for elderly people since they have weaker strength. Differences in resistance between



group 1 and group 2 can be explained by the fact that the physical load exerted by the robotic walkers, defined as the resistance multiplied by time, during one gait cycle is similar for both groups. Since GPD1 only exerted resistance on the subjects during 60% of the experimental time, the level of resistance is increased by 50% (hence, 31 N) for that group.

Before the experiment, EMG sensors are attached to every user's lower limb muscles (TA, GAS, and SOL) and IMU sensors are used to monitor the joint angles. Since this experiment is conducted to verify the effectiveness of plantar flexors, only the muscular activation data and joint angle data from the lower leg are obtained for analysis. None of the subjects suffered from any conditions which can damage their walking ability or balance, and the walking symmetry of all of the subjects is visually verified by physical therapists.

Subjects are instructed to walk straight for 30 minutes, twice, without the robotic walker to obtain the muscular activation and kinematic data of normal walking. These data serve as a baseline to evaluate the changes in the muscular activation and kinematics when they walk with the robotic walker. Furthermore, to observe for any changes after walking with the robotic walker, the subjects are asked to repeat the normal walking (without the walker) after they have completed 20 minutes of walking with the robotic walker, which is used for evaluating the effects of fatigue on the muscular activities. Hence this experiment can be temporally distinguished into three phases: normal walking without the walker (Before), followed by walking with the robotic walker (Training) for 20 minutes, and then finally, normal walking again (After).

### E. DATA PROCESSING AND STATISTICAL ANALYSIS

The data from 20 strides of each trial were used for analysis, excluding the data from the start and end of the trial. In total, the data of 8 subjects (320 strides in total) in each condition were included for data processing and analysis.

A customized program in MATLAB (MATLAB R2019a, CA, USA) was used for data processing. The EMG signals were filtered using a band-pass filter (10–400 Hz) and rectified. The mean amplitude was calculated for each stride. Then, the root mean square (RMS), within a 100-ms time window, was applied to generate the linear envelope for each stride. The mean amplitude of the EMG of each stride and each muscle was calculated using the rectified signal. The resulting EMG envelope and mean amplitude of each muscle were normalized to the respective muscle's maximal voluntary contraction (MVC). The subjects were instructed to perform MVCs against manual isometric resistance with specific test positions for each muscle of interest [34]. Each test was repeated three times, and the peak value was used as the MVC value.

As for the experiment with elderly subjects, data from 40 strides in the middle of the Training (around the 10 minute mark), as well as from the Before and After periods, were used for analysis. Since the elderly subjects cannot always

perform maximum contraction of their muscle (due to pre-existing conditions) [35], each subject's maximum EMG value obtained among the experiments is used to normalize the data, instead of the previously used MVC.

A one-way analysis of variance was used to assess differences between the mean values of the EMG and kinematic data. Bonferroni's post hoc test was used to calculate the pairwise differences. All significance levels were set at  $P < 0.05$ .

## III. RESULTS

### A. EMG DATA

The mean values and standard deviations of normalized EMG data obtained from eight lower limb muscles and an upper limb muscle are shown in TABLE 2 and FIGURE 3. The linear envelopes of all nine muscles are shown in FIGURE 4. The average amplitude of the lower limb muscles is calculated using the data in the stance phase under the assumption that the external resistance has negligible effects on muscle activity during the swing phase.

During walking without additional resistance, EMG activities in the GAS, SOL, TA, and TRI were found to be significantly lower than those in any other conditions. Furthermore, the activation level of GM without resistance was lower than that in any other condition, except for GPD2. In the condition with constant resistance, the activity of all nine muscles was higher than that in the Zero condition. There was no significant difference in all of the muscle activities between the Const and the GPD conditions except for the GM and TRI.

### B. KINEMATIC DATA

The mean value and standard deviation of the joint angle parameters are shown in TABLE 3 and FIGURE 5. The profile of each joint during the gait cycle is shown in FIGURE 6.

There were significant differences in the knee joint angle at the moment of heel strike (knee HS). Knee flexion was highest in the Const condition and lowest in the Zero condition. In the GPD2 condition, there was a significant increase in hip flexion and reduction in hip extension, while the range of movement of the hip joint was the same. With external resistance (both constant and GPD), the range of movement in the ankle and hip joints was significantly higher than that in the Zero condition.

### C. RESULTS OF AN EXPERIMENT WITH ELDERLY SUBJECTS

The kinematic data and EMG data from the experiment with elderly people are shown in TABLE 4 and FIGURE 7. GPD1 leads to higher activation in GAS, SOL, and a lower activation level in TA in comparison with walking without the robotic walker. Furthermore, the range of movement of the ankle joint also observably increased. In contrast, constant resistance increased the activation level in all

**TABLE 2. Average amplitude of normalized EMG data.**

	Zero	Const	GPD1	GPD2	GPD3
GAS	8.80 ± 2.72 *	10.44 ± 2.51#	10.02 ± 2.41#	9.55 ± 2.24 **	10.10 ± 2.49#
SOL	17.91 ± 6.39 *	22.70 ± 9.52#	23.41 ± 14.11#	21.93 ± 10.84#	24.70 ± 15.67#
TA	4.90 ± 1.49 *	6.27 ± 2.07#	5.51 ± 1.66*#	5.78 ± 1.56*#	5.97 ± 1.58#
BF	1.38 ± 0.71	1.54 ± 0.79	1.50 ± 0.72	1.56 ± 0.84 #	1.56 ± 0.78 #
SEM	7.36 ± 4.17	7.49 ± 3.23	7.43 ± 3.06	7.87 ± 3.10	8.22 ± 3.53 #
RF	1.86 ± 0.86 *	2.11 ± 0.95#	2.00 ± 0.73	1.56 ± 0.57 **	1.69 ± 0.73 *
VL	8.98 ± 4.47	9.25 ± 5.36	9.04 ± 4.99	7.30 ± 4.54 **	8.18 ± 4.53*
GM	7.99 ± 2.76 *	9.55 ± 3.09#	8.78 ± 3.04*#	8.54 ± 2.88*	8.85 ± 3.12*#
TRI	4.47 ± 2.69*	7.41 ± 4.13#	6.70 ± 3.01*#	7.37 ± 3.17#	6.48 ± 3.41*#

\* Statistical difference from Const condition.  
 # Statistical difference from Zero condition.  
 EMG data are normalized by MVC. Unit is % of MVC.

**TABLE 3. Summary of joint angles.**

	Zero	Const	GPD1	GPD2	GPD3
Ankle HS	-4.09 ± 5.13*	-2.53 ± 3.21#	-2.11 ± 4.84#	-1.41 ± 5.4*#	-1.61 ± 4.92#
Ankle MAX	14.34 ± 6.25	15.57 ± 7.56	15.23 ± 7.28	15.35 ± 7.28	15.92 ± 7.54
Ankle MIN	-26.76 ± 4.94	-27.75 ± 7.52	-29.53 ± 7.66*#	-27.83 ± 6.26	-28.77 ± 6#
Ankle ROM	41.10 ± 6.35*	43.32 ± 5.82#	44.76 ± 8.48#	43.18 ± 7.12#	44.69 ± 7.71#
Knee HS	7.75 ± 6.33*	13.70 ± 7.81#	10.78 ± 6.90*#	11.43 ± 7.38*#	11.55 ± 6.03*#
Knee MAX	59.43 ± 2.34	59.14 ± 2.42	58.63 ± 3.91#	58.4 ± 3.49#	59.96 ± 4.83*
Knee MIN	-2.13 ± 3.09*	-0.17 ± 2.00#	-0.37 ± 2.24#	-0.84 ± 2.14*#	0.10 ± 2.30#
Knee ROM	61.59 ± 3.19*	59.26 ± 3.23#	58.94 ± 3.77#	59.25 ± 3.75#	59.70 ± 4.37#
Hip HS	8.83 ± 8.43*	13.77 ± 6.72#	14.73 ± 6.86#	18.78 ± 5.95*#	13.31 ± 8.31#
Hip MAX	17.01 ± 7.11*	20.02 ± 7.06#	21.26 ± 6.34#	24.96 ± 4.75*#	22.06 ± 6.70*#
Hip MIN	-22.83 ± 7.89*	-20.83 ± 6.61#	-20.94 ± 5.23#	-16.58 ± 5.37*#	-20.61 ± 7.47#
Hip ROM	39.84 ± 2.95*	40.85 ± 3.39#	42.20 ± 3.11*#	41.54 ± 2.86#	42.66 ± 3.23*#
ST	1.34 ± 0.16*	1.49 ± 0.17#	1.38 ± 0.10*#	1.48 ± 0.14#	1.50 ± 0.19#

\* Statistical difference from Const condition.  
 # Statistical difference from Zero condition.

Ankle ROM, Knee ROM, and Hip ROM are the range of movement (ROM) of each joint. Ankle MAX is the maximum dorsiflexion angle and Ankle MIN is the maximum plantarflexion angle. Knee MAX and Knee MIN are the maximum and minimum knee flexion angles respectively. Hip MAX and Hip MIN are the maximum hip flexion and extension angle. ST is the stride time and the unit is second.

three muscles and the range of movement of the ankle joint. No statistical differences were found in the zero resistance condition, except for a marked decrease in TA activation.

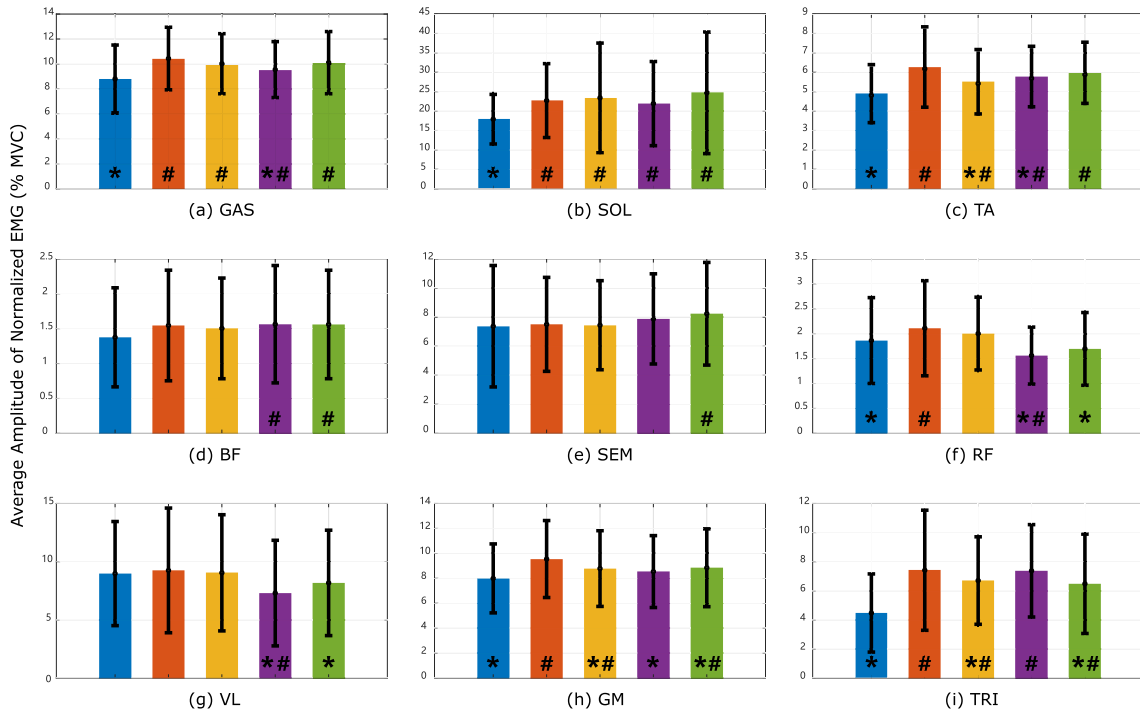
**IV. DISCUSSIONS**

**A. EFFECTS ON KINEMATICS AND MUSCULAR ACTIVATION**

From the results of the kinematic data, there were significant differences at the moment of HS among the conditions. Subjects tend to contact the ground with larger hip flexion and knee flexion at the moment of the HS in conditions with external resistance. Subjects try to contact the ground with their full feet rather than only the heel because external resistance makes users choose to walk with higher stability. The same change in kinematics can also be found in another

study that imposes horizontal external resistance on the user’s center of mass [17]. In the GPD2 condition, the hip peak flexion and extension angles were significantly higher, while the range of movement was almost the same, indicating that the users tend to lean forward in this condition, while the reason for this choice of posture is unclear. We assume that the walker exerts resistance to users when the user’s center of mass moves forward and passes their support leg and that the walker releases resistance at the moment when their swing leg contacts the ground, thus enhancing their walking stability and eventually leading to a lean-forward posture during walking.

Different types of resistance result in different activation patterns in the lower limb muscles. The imposed resistance (both GPD and constant) leads to higher activation levels in the TA, which is activated in the early stance phase to stabilize



**FIGURE 3.** Average amplitude of EMG signals. The blue, orange, yellow, purple, and green bars show the average amplitude of each muscle in the Zero, Const, GPD1, GPD2, and GPD3 conditions, respectively. \* Statistical difference from Const condition. # Statistical difference from Zero condition.

**TABLE 4.** Summary of experiment with elderly subjects.

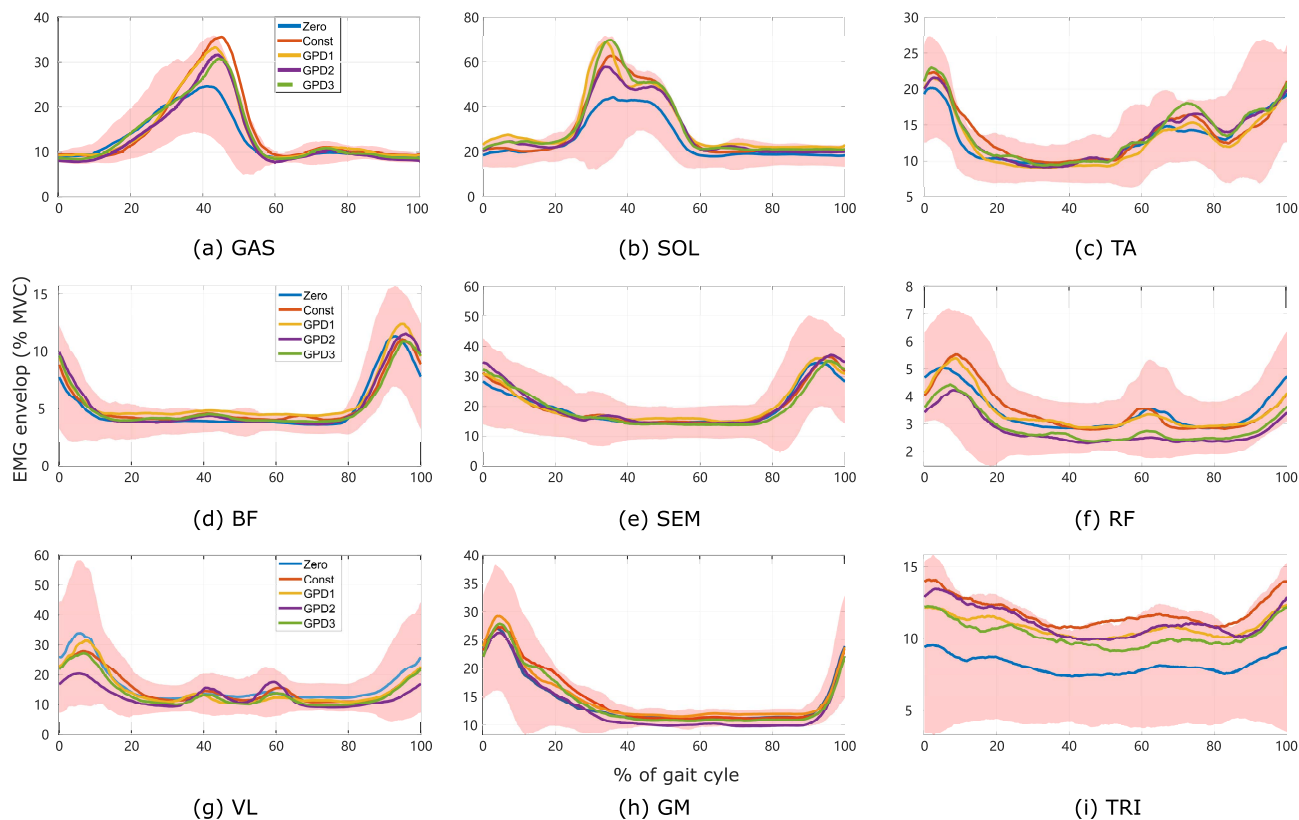
	No. of group	Before	Training	After
Ankle Max	1	10.52 ± 3	12.13 ± 1.91 *	9.84 ± 2.48
	2	11.99 ± 2.33	13.38 ± 2.92 *	10.69 ± 2.62 *
	3	14.91 ± 2.9	14.08 ± 2.37 *	14.18 ± 2.35
Ankle Min	1	-19.78 ± 3.8	-20.69 ± 3.24	-23.06 ± 2.98 *
	2	-23.22 ± 4.9	-24.51 ± 4.67	-25.88 ± 5.15 *
	3	-24.41 ± 11.23	-26.97 ± 9.53	-26.34 ± 11.03
Ankle ROM	1	30.53 ± 4.78	32.82 ± 4.53 *	32.85 ± 4.33 *
	2	35.37 ± 5.36	37.89 ± 4.7 *	36.56 ± 5.29
	3	39.24 ± 8.84	41.05 ± 7.68	40.63 ± 9.59
GAS	1	27.34 ± 9.15	30.27 ± 10.7 *	25.43 ± 9.09 *
	2	18.67 ± 7.55	20.86 ± 7.94 *	18.11 ± 6.01
	3	26.92 ± 10.25	27.93 ± 10.21	26.45 ± 9.18
SOL	1	25.46 ± 9.6	28.79 ± 10.14 *	25.88 ± 9.32 *
	2	26.72 ± 9.07	32.44 ± 11 *	28.95 ± 8.64 *
	3	29.51 ± 10.85	26.52 ± 10.31 *	28.33 ± 9.86
TA	1	27.75 ± 9.49	26.38 ± 7.44 *	25.24 ± 8.53 *
	2	30.02 ± 7.97	32.4 ± 8.75 *	28.87 ± 8.59
	3	31.26 ± 7.29	26.01 ± 7.89 *	27.68 ± 6.9

\* Statistical difference from Before (Walk without the robotic walker at the beginning).

Ankle ROM is the range of movement (ROM) of the ankle joint. Ankle Max is the maximum dorsiflexion angle and Ankle Min is the maximum plantarflexion angle.

the ankle joint against the impact of HS and smoothly lowers the forefoot to the ground [36]. The highest activation level of TA is found in the condition of constant resistance because TA needs to be activated for a longer duration to resist the external resistance throughout the early stance phase. This can also explain why the activity with GPD1 resistance is significantly lower than that with constant resistance because there is no resistance exerted to users around the moment HS in the GPD1 condition. At the moment of HS, RF is activated to generate a braking force to slow down the progression

of the lower limb [19], which makes RF highly related to walking speed [37]. This is the reason why the RF activation level is higher in the conditions when subjects walk with a higher walking cadence (GPD1 and Zero). However, there is an exception in that the RF average amplitude in the Const condition is the highest among all of the conditions, while the walking cadence is very low. Considering that the swinging leg contacts the ground with a significantly larger knee flexion angle and follows with a knee extension movement in the Const condition, we assume that the knee



**FIGURE 4.** Linear envelope of nine muscles. The pink shadow shows the standard deviation in the Zero condition. The blue, orange, yellow, purple, and green lines show the average amplitude of each muscle in the Zero, Const, GPD1, GPD2, and GPD3 conditions, respectively.

extensors, both RF and VL, contribute to generating an extra propulsion force at the early stance, to resist the external resistance. As for the hip joint, the GM activity in all GPD conditions was significantly higher than that in the Zero condition and significantly lower than that in the Const condition. GM is the main contributor in the early stance to generate propulsion force [30], and the walker releases resistance at a part of the early stance in all GPD conditions, which results in a significant reduction in GM activity.

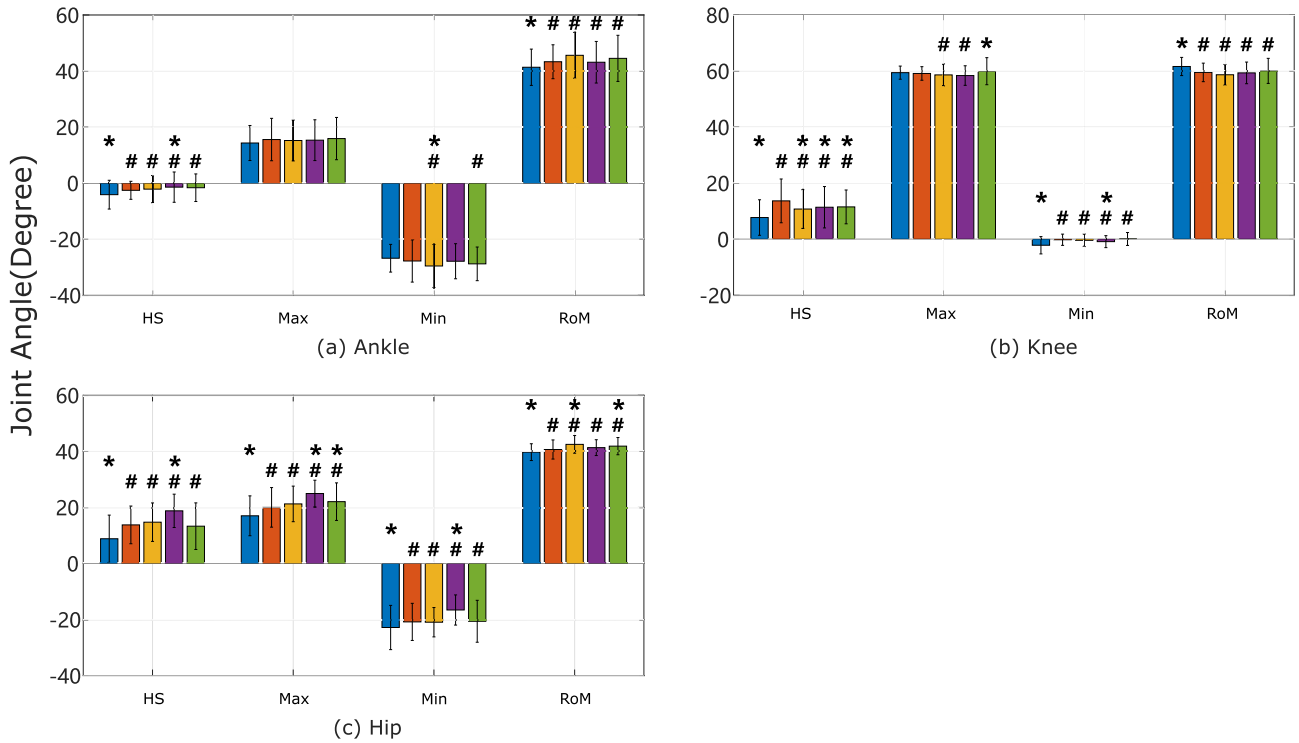
During the experiment, subjects were given the instruction to “Try to keep the upright posture during walking”. As shown in TABLE 5, participants walk with similar inclination angles (78-84 degrees); however, the GPD 2 condition results in a larger inclination angle compared to GPD 1, GPD 3, and Zero conditions. However, the different inclination angles (GPD2 vs. other conditions) did not lead to different muscular activation patterns as shown in FIGURE 4, in which the activation profile of the muscles monitored under different conditions have similar patterns. The different inclination angles did change the activation levels of some muscles, as the gastrocnemius muscular activation level decreases while the triceps brachii muscular activation level increases for the GPD 2 condition. Among all of the eight subjects, five of them walked with a smaller inclination angle (lean more forward) in comparison with that under other conditions, and the muscular activation

levels of the gastrocnemius of four of the five subjects are lower than that under other conditions. While the other three subjects who walked with a similar inclination angle as that under other conditions had a similar level of muscular activation as others. Another important point shown in TABLE 5 is that the lean forward angle with the GPD2 control has a large variance in comparison with that of the other control strategies. This results from the inter-subject difference instead of the intra-subject difference. There was one subject whose lean forward angle was much smaller than the others, reaching 62.64 degrees. As a result, the activation level of the gastrocnemius is much smaller than that of the others.

**B. EFFECTIVENESS AND EFFICIENCY OF TRAINING PLANTAR FLEXORS**

The activity of the GAS in all GPD conditions was significantly higher than that in the Zero condition, and it was similar to that in the Const, except that the GAS activity in GPD2 was significantly lower than that in the Const condition. The reduction in GAS activity results from users’ forward-leaned posture in the GPD2 condition because walking with a forward-leaned posture allows users to place more bodyweight on the handle of the walker. Therefore, the plantar flexor activity decreases because they are responsible for supporting the bodyweight of users during walking [18]. This phenomenon can also be found





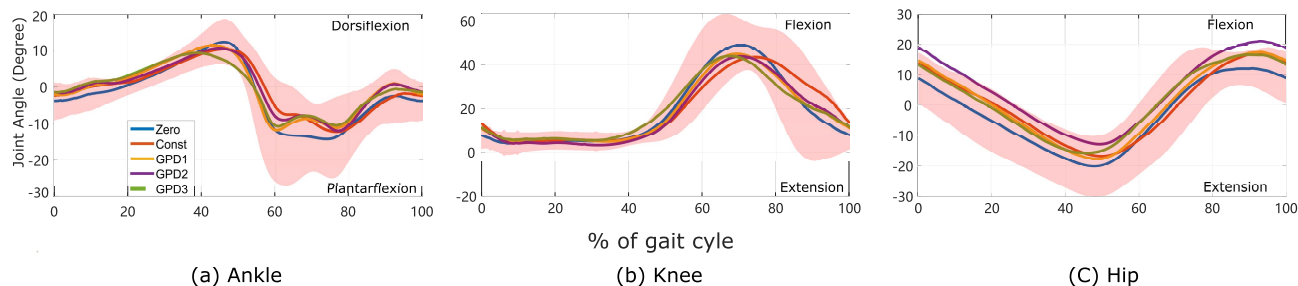
**FIGURE 5.** Mean amplitude of kinematic parameters. The blue, orange, yellow, purple, and green bars show the average amplitude of joint angles in the Zero, Const, GPD1, GPD2, and GPD3 conditions, respectively. HS stands for the angle at the moment of heel strike. Max means the maximum joint angle in a gait cycle. Min means the minimum joint angle in a gait cycle. RoM means the range of movement of a joint. \* Statistical difference from Const condition. # Statistical difference from Zero condition.

in a study by Suica *et al.* [38], in which they found that increased weight-bearing of the handle of the walker results in a decrease in lower limb muscle activation. The speed and range in which the muscles act is also important for achieving the task-specificity of strength training [9], [39]. It was found that high-speed resistance training is more effective for improving the functional performance and muscle power of elderly women in comparison with low-speed resistance training [40]. We found out that the maximum plantarflexion angular velocity, which is the mean value of the maximum plantarflexion angular velocity in each gait cycle, in the GPD1 ( $318.7 \pm 70.61$  degrees/second) and GPD3 ( $325.79 \pm 84.26$  degree/second) conditions are significantly higher than that in the Const condition ( $298.76 \pm 82.73$  degree/second) as shown in TABLE 6. The effectiveness of GPD1 and GPD3 in functional strength training for plantar flexors is better than that of constant resistance because of the higher speed of plantarflexion movement during walking.

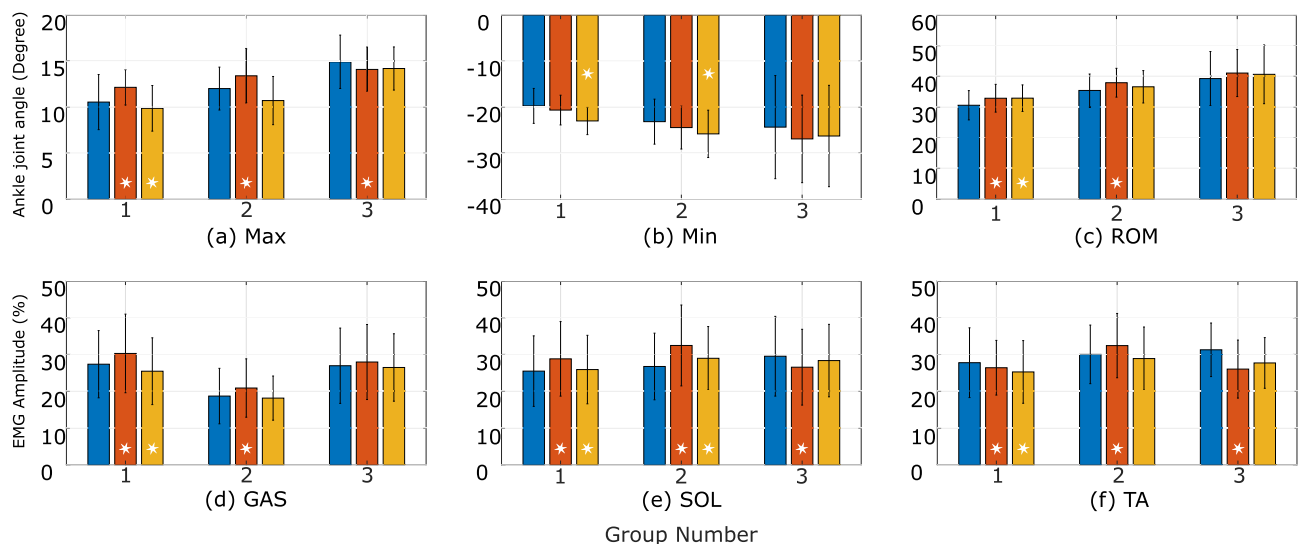
Soleus and gastrocnemius have similar functions during walking or standing; however, their contribution to adjusting posture to keep stable can be different due to their different compositions of muscle fibers. Soleus contains less type II muscle fiber (known as fast muscle fiber) than that of gastrocnemius [41]; therefore, the gastrocnemius can generate a higher muscle power because fast muscle fibers contract with higher speed. Furthermore, the muscle thickness of

gastrocnemius deteriorates faster than that of soleus, with aging, due to the fact that the fast muscle fibers deteriorate faster than slow muscle fibers. In a study conducted by Fujiwara *et al.* [42], where there were 800 subjects, elderly people were found to have a smaller gastrocnemius while there is almost no difference in the size of the soleus in people of all different ages. A weaker gastrocnemius causes the ability to react to disturbances during walking or standing to weaken which leads to a weaker ability to balance. A walk training that includes high-speed functional training can be more beneficial for elderly people to restore balance and walking performance. Therefore, it is more effective to provide a training strategy that induces a fast plantar flexion and high muscular activation in gastrocnemius, which can be achieved by choosing the GPD1 or GPD3 control strategies, to improve balance in elderly people.

The triceps brachii is the only muscle that can generate force to extend the elbows [43], and it is the main muscle that generates force to push the walker during walking. Therefore, we chose TRI as an indicator of the physical load from the walker to the users. We found a significant difference in the activation of the TRI in constant condition with that in any other conditions except for the GPD2 condition, while the high activity in the GPD2 condition results from the lean-forward posture, which puts higher weight on the walker through the arms. A lower physical load on users during training can allow the user to take longer



**FIGURE 6.** The ankle, knee, and hip joints. Plantarflexion is negative, and dorsiflexion is positive. Knee flexion is positive. Hip flexion is positive, and hip extension is negative. The shadow shows the standard deviation in the Zero condition. The blue, orange, yellow, purple, and green lines show the average amplitude of joint angles in the Zero, Const, GPD1, GPD2, and GPD3 conditions, respectively.



**FIGURE 7.** Results of experiment with elderly subjects. The blue, orange, and yellow bars show the experiment results from Before, Training, and After, respectively. Max means the maximum joint angle in a gait cycle. Min means the minimum joint angle in a gait cycle. ROM means the average range of movement of a joint. And the white star stands for a statistical difference from Before. (Walk without the robotic walker at the beginning.)

training without causing fatigue in their arms, which can lead to a more efficient method for improving lower limb strength.

In the experiment with elderly people, only GPD1 is applied to subjects because GPD1 shows the best training performance among GPD conditions for young subjects. In this experiment, we observed similar effectiveness of the training for plantar flexors as that in the experiment with young subjects. The activation levels and the range of movement for plantar flexors increased when GPD control or constant resistance was applied when compared to normal walking (without the walker). Furthermore, the maximum plantar flexion angular velocity increased when the GPD control was applied (Table 7), whereas, there seem to be no significant differences in Group 2, where the resistance was constant. In both young adults as well as elderly participants, constant resistance stimulated a higher activation level for TA by extending its activation duration (FIGURE 4). We assume that the activation of TA enhanced the stability of the ankle joint after heel contact while simultaneously limiting the plantar flexion motion of the joint, which eventually

led to a reduced angular velocity of the ankle joint. Our results indicate that, in terms of high velocity stimulation of the ankle joint, training of plantar flexors with the proposed GPD control is more effective than training with constant resistance. And high speed training is preferable for enhancing the power of muscles of elderly people [40]. Since the experiment requires subjects to walk for over 20 minutes, it could lead to fatigue of the lower limb muscles and fatigue is a factor that often changes the amplitude of an EMG signal [44], [45]. However, fatigue can be excluded here since the activation of GAS and SOL decreased right after the experiment, indicating that fatigue is not the reason for the observed higher activation level. In the reference group, the zero resistance did not increase the activation levels of the three muscles, since the lack of resistance does not contribute to strength training for plantar flexors. Moreover, there was no statistical difference among the range of movement of the ankle joint when they walk with no resistance. Thus, the results indicate that simply walking with a walker without any additional resistance is not beneficial for lower limb muscle training.

**TABLE 5. Inclination angle.**

Zero	Const	GPD1	GPD2	GPD3
84.30 ± 3.25 **	79.22 ± 5.80#	79.89 ± 5.33# **	78.79 ± 7.29#	80.01 ± 4.44# **

\*\* Statistical difference from GPD2 condition.

# Statistical difference from Zero condition.

The inclination angle is calculated as the angle between the ground and the line between the S1 spine joint and the T8 spine joint. The data in the above table is the mean value inclination angle of each gait cycle under different conditions.

**TABLE 6. Maximum plantarflexion angular velocity.**

Zero	Const	GPD1	GPD2	GPD3
331.38 ± 73.94*	298.76 ± 82.73#	318.7 ± 70.61*	292.72 ± 68.12#	325.79 ± 84.26*

\* Statistical difference from Const condition.

# Statistical difference from Zero condition.

The maximum plantarflexion angular velocity is calculated as the mean value of the maximum angular velocity of the ankle joint in the plantarflexion direction from each gait cycle.

The unit is degree/second.

**TABLE 7. Maximum plantarflexion angular velocity of experiment with elderly subjects.**

	Before	Training	After
Group 1	313.65 ± 36.8	325.41 ± 40.33 *	308.73 ± 40.33
Group 2	342.39 ± 66.45	334.58 ± 49.42	329.02 ± 43.93
Group 3	389.81 ± 38.4	404.95 ± 42.65 *	386.25 ± 58.51

\* Statistical difference from Before.

The unit is degree/second.

### C. CLINICAL IMPLICATION

Based on the results from the kinematics and muscular activity, we conclude that walking under the GPD1 condition reaches a rehabilitation outcome for plantar flexors with higher effectiveness and efficiency compared to those with constant resistance. The GPD3 condition leads to the highest plantarflexion angular velocity but did not lead to a significant difference in the plantar flexion angle statistically. The GPD1 condition leads to both a larger plantarflexion angle and faster plantarflexion angular velocity as compared to constant resistance with a statistical difference. Training with a high speed can result in the improvement of explosive muscular force [46], which is beneficial for making a quick recovery motion from a loss of balance [47], [48]. Also, we believe that the GPD control also has the potential to be used for training other parts of muscles. As an example, resistance exerted on users at the 0% to 20% of a gait cycle can contribute to enhancing the hip extensor without altering the knee joint angle, while maintaining a high speed of gait.

Constant resistance can improve rehabilitation outcomes for plantar flexors, knee extensors, and hip extensors. Plantar flexor and hip extensor improvements can contribute to better walking ability. Furthermore, the improvement in knee extensor strength can contribute to other movements such as sit-to-stand or climbing stairs [49], while the training for knee extensors does not contribute much to walking ability [9]. The external constant resistance results in a high physical load on users' upper limb muscles, which can cause fatigue in users' upper limbs that can limit the efficiency of strength training.

During training with a robotic walker, a leaning-forward posture reduces muscular activity in the lower limb muscles and results in fatigue in the upper limbs. Therefore, it is better to instruct users to maintain an upright posture during training to optimize the rehabilitation outcome for the lower limb muscles by using robotic walkers. It will help us to understand the mechanism of lower limb training by using a robotic walker.

### D. LIMITATIONS

As a next step, we will continue this research by recruiting both male and female elderly people to conduct a long-term experiment that lasts for two months so that we can verify the changes in kinematics and muscular activation and evaluate the rehabilitation outcome of the proposed gait phase-dependent control algorithm.

We also found that posture during walking influences the training effectiveness for the lower limb muscles. We will continue to investigate the parameters that can influence the effectiveness of lower limb strength training through the use of a robotic walker. The effects of different postures will be investigated by changing the postures (distance to the walker, body lean angle, etc.) during walking with a robotic walker.

### V. CONCLUSION

In this study, we evaluated the effects of the proposed gait phase-dependent (GPD) control on the muscular activity and kinematics of the lower limb muscles. By using GPD control, a robotic walker can exert resistance to users at a certain gait phase. We conducted experiments with young

subjects under conditions of zero external resistance, constant resistance, and GPD resistance. We found that the proposed GPD resistance can enhance the muscular activity of the plantar flexors and that the angular velocity of the GPD condition was higher. At the same time, upper limb muscular activity decreased compared to that with constant resistance. Specifically, GPD1 and GPD3, which provide resistance at the gait phase of 10% to 40% and 30% to 60% respectively, attain higher performance for training the plantar flexors and reduce the unnecessary burden on upper limbs supporting users themselves and push the walker forward, which is found by a significant difference from that in a constant resistance condition. Furthermore, we verified the training effectiveness of GPD1 for lower limb muscles by conducting an experiment with both male and female elderly subjects. We conclude that the proposed GPD control can achieve a rehabilitation outcome for plantar flexors with high effectiveness and efficiency.

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