Interactive Gait Rehabilitation System with a Locomotion Interface for Training Patients to Climb Stairs

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Abstract

This paper describes the development of a gait rehabilitation system with a locomotion interface (LI) for training patients to climb stairs. The LI consists of two 2-DOF manipulators equipped with footpads. These can move the patient's feet while his or her body remains stationary. The footpads follow the prerecorded motion of the feet of healthy individuals. For gait training, the patient progresses sequentially through successively more advanced modes. In this study, two modes, the enforced climbing of stairs and interactive climbing of stairs, were developed. In the interactive mode, foot pressure sensors are used to realize interactive walking. Comparisons were made between the modes for healthy individuals and a patient. The effectiveness of the system was examined using electromyography (EMG) and foot pressure data.

1   Introduction

In gait rehabilitation, physiotherapists describe the motion of walking to patients verbally or by physically manipulating the patient's body. However, there is an insufficient number of physiotherapists for the number of patients to be treated, which means their workloads are high. It is difficult to teach patients to walk with a satisfactory motion, even if the patients have the capability. In addition, the development of motion assist systems is becoming more crucial as we plan for an aging society.

To solve these problems, some systems using robotics have been developed. Hitachi Ltd. commercializes rehabilitation systems using two belt-type treadmills. These are already being used in many hospitals. However, the motion they generate is only in the horizontal direction. Colombo et al. have developed a gait training system with a treadmill and 4-DOF robots, which they have given the name Lokomat (Colombo, Joerg, Schreier, & Dietz, 2000). This system can be used to teach patients the correct motion; however, it is complicated and cumbersome to install and remove the patient. Hesse et al. have developed a gait trainer (Hesse & Uhlenbrock, 2000) and the Haptic Walker (Schmidt, Werner, Bernhardt, Hesse, & Krüger, 2007). These systems have two footplates which move the patient's feet. They used the Haptic Walker to develop a training system for climbing stairs (Hussein, Schmidt, Hesse, & Kruger, 2009), as
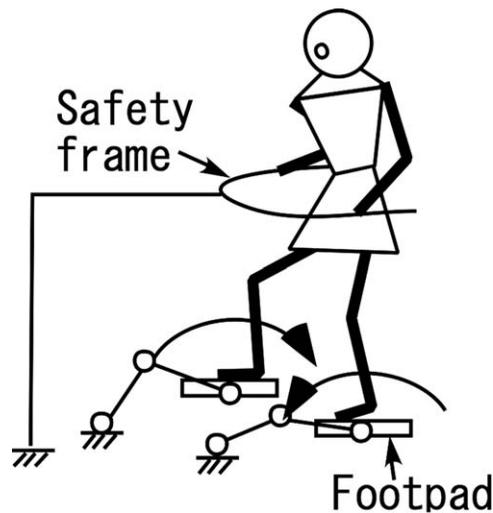
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Figure 1. *Basic concept of the system.*

well as one for walking on flat surfaces. Currently, their system, which is called the G-EO-System, is sold by Reha Technologies (Hesse, Waldner, & Tomelleri, 2010).

Such systems help physiotherapists to reduce their manual labor and allow them time to provide quality support while helping patients get more exercise in anticipation of attaining a higher level of recovery. Since 2000, the authors have been developing an efficient training system employing a locomotion interface that uses virtual reality technology to directly move the patient's feet using a manipulator (Yano, Noma, Iwata, & Miyasato, 2000). We have applied this system to various patients and have confirmed its effectiveness (Yano, Kasai, Saitoh, & Iwata, 2003).

Usually, in rehabilitating patients to ascend and descend stairs, real stairs, a mock-up of real stairs, or a specialized treadmill such as the StepMill (SM916 Step-Mill, 2012), are used. Since there is a risk of falling with all these methods, they are performed only with patients who have acquired stable walking skills. Our system, however, can support the patient's body safely by using footpads and a safety bar placed around the system (see Figure 1). A safe training system for ascending/descending stairs can be realized with this feature. The G-EO-System (Hesse et al., 2010) can also be applied to stroke patients. In this system, the patient is suspended using a body weight support system and thus he or she can repeatedly follow the foot trajectory in the same way as

with our system. In climbing stairs, one has to move one's legs in such a way as to avoid tripping on the staircase. Because the footpads of the above systems move a patient's feet involuntarily, the patient tends to exert some pressure on the footpad in the swing phase since there is no feedback on this provided to the patient. However, previous systems did not offer the interactive walking mode described in this paper.

In this study, gait rehabilitation methods for climbing stairs are proposed. A compact gait rehabilitation system that can be used with a commercial counterweight system was developed. Using this system, haptic rendering methods for training patients to climb stairs, especially in an interactive mode, were developed. An evaluation was carried out to examine the effectiveness of the proposed methods.

2 System Configuration

2.1 Design Concept

Gait rehabilitation is usually applied to patients who develop problems walking, due to damage to bones, joints, nerves, or muscles through accident, injury, or disease. In general, gait rehabilitation begins with training to distribute the weight evenly on each foot. Then, the patient learns how to move his or her body and how to walk by himself or herself. Usually, the physiotherapist describes the walking motion to the patient verbally or by physically manipulating the patient's body. These methods are based on the concept of relearning and also on the idea that alternative nerves can be activated even though the nerves usually used are injured. However, it is difficult for physiotherapists to repeat the same motion for a long period of time, since they become tired and also have insufficient time to thoroughly train each patient. If this movement can be realized using equipment, the physiotherapist can regulate the motion of the patient's body with less fatigue. The other advantage of such equipment is that physiotherapists who are not strong enough to manipulate the patient's body can train them using the equipment. To realize these goals, the equipment should be able to repeatedly move the patient's legs through any trajectory and to do so for a long period of time.

The motion presented by the system can be belt-driven, such as in a treadmill, or can be an exoskeleton, such as in the Lokomat (Colombo et al., 2000). The belt-driven technique has more advantages in that it allows a more liberal selection of walking styles; however, it provides very little restraint for the legs and demands certain skills in order for the patient to keep a constant position on the belt. On the other hand, exoskeletons, with actuators arranged along the physical joints, can accurately repeat the same motion, but they need to be adapted to each individual because of differences in walking style. Furthermore, attaching and detaching the devices that restrain the joints is a complicated operation.

In our research, we have adopted a system to move each foot with a manipulator with 2 DOF (back and forth and up and down) that allows repeated walking cycles, can easily be attached and detached, and moderately restrains the body. Given the range of movement of human joints, the device is designed so as to move only the feet, leaving the patient to choose the movement of the joints in the legs, hips, and other parts of the body. If the patient tries to stay upright on the footpads, his or her body will be jarred by the motion of the footpads. To avoid this, the patient must move his or her whole body. Since the trajectory of the footpads is prerecorded from healthy individuals, the body movement adopted by the patient will resemble the movement of a healthy individual. With this design, we have achieved compatibility in the amount of exercise and provided moderate restraint.

2.2 Hardware

We developed a locomotion interface which we call the GaitMaster5 (GM5; see Figure 2). The size of the GM5 is 965 mm (H) × 885 mm (W) × 1200 mm (D). It is a manipulator-type locomotion interface with two footpads that trace a virtual floor beneath the feet. When the patient moves one of his or her feet forward, the footpad under the foot in the swing phase follows it like a shadow on the virtual floor. At the same time, the other footpad, the one beneath the foot in the standing phase, moves back the same distance as the other foot
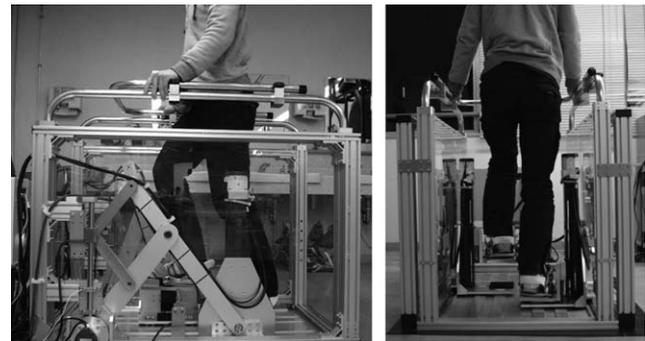

Figure 2. *Overview of the system (Left: side view, Right: back view).*

moves forward. By repeating this motion, the patient can walk on an infinite uneven virtual terrain, while his or her body remains stationary in the real world.

The GM5 consists of two 2-DOF motion platforms (see Figure 3) which have a linear guide and a slider-crank mechanism. To effect walking at 1.0 m/s on a flat surface, the footpads need to move back and forth at a maximum speed of 2.0 m/s. Since it is difficult to realize the 2.0 m/s velocity using a commercially available linear actuator, we use a crank-slider mechanism for the back-and-forth motion. To increase the stiffness of the slider-crank linkage, a parallel linkage is attached on the driving link of the slider-crank. The linkage is driven using an AC servomotor (MQMA041P1S, Panasonic), and gears with a reduction ratio of 160:1 (CSF-32-160-2UH, made by Harmonic Drive Systems). The footpads can move at a maximum speed of 2.0 m/s horizontally and generate a maximum force of more than 540 N.

A linear actuator with vertical movement is fitted at the end of the mechanism. Because the maximum velocity of each footpad is 0.5 m/s, we use a compact linear actuator that can safely support the patient's body. The actuator consists of a linear slider, KR45H10A+440L (THK), and an AC servomotor, MQMA041P1S (Panasonic), and has a vertical working range of 315 mm. It can generate a maximum vertical force of 1225 N and move at a maximum speed of 0.5 m/s.

On top of each motion platform there are 300 mm × 270 mm footpads. The footpads can move 700 mm horizontally and 315 mm vertically. The maximum walking speed on the GM5 is 1.0 m/s and the maximum payload
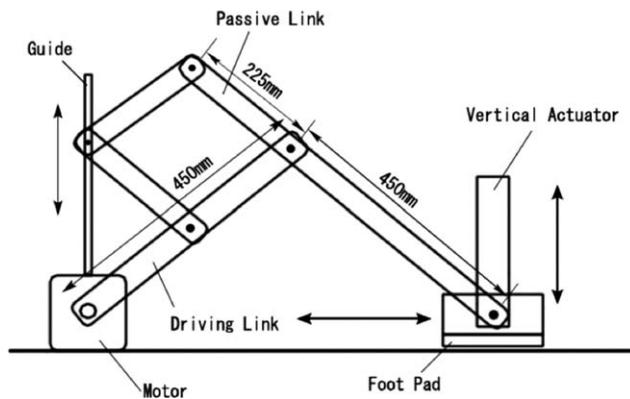
Figure 3. *Mechanical configuration of the GM5.*

of each motion platform is approximately 80 kg. The position of each footpad is measured using an optical rotary encoder on the AC servomotor, an MQMA041P1S made by Panasonic. To realize real-time operation, the Realtime Express controller of the AC servomotor is used. This is an embedded motor controller, controlled from a PC. The controller is connected to a PC board (HCRTEXsd, made by HPtech) on the PC via a LAN cable. The control program on the PC is written in Visual C++ and the HCRTEXsd SDK is on Windows Vista. The data for the trajectory of the footpads described in Section 3 are cyclic with a frequency of 1.0 Hz, and consist of 30 data points. These data points are subdivided into 1,000 points (1 ms position data) using a linear interpolation between each adjacent point. The program on the PC inputs 50 of these data points to the HCRTEXsd for the next 50 ms motion of the footpads. The Realtime Express controller reads the data in sequence from the HCRTEXsd and controls the AC motor changing its destination each millisecond. We provide the patient with the position/trajectory of the feet, since we think that the correct trajectory enables the patient to develop the correct motion. Although force control or impedance control can be used as alternatives for reasons of safety, positional errors can occur. Also, the controller of the AC servomotor only supports position control. Another consideration is the cost, so position control, which is cheaper to implement, is used in this system. In the GM5, since the patient's feet are on the footpads and the space above is clear, there is little
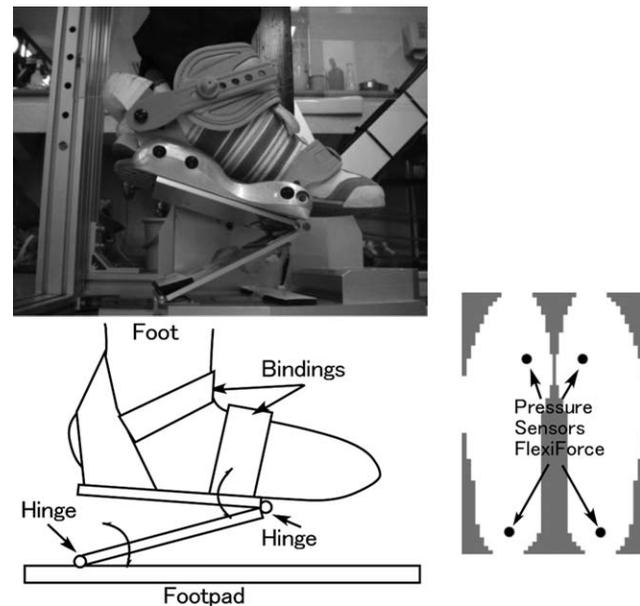

Figure 4. *Overview of the footpad of the GM5.*

risk that the patient's feet will collide with anything on the GM5. However, patients who have problems with their knee joints are excluded from using this system.

To fix the patient's foot to the footpad, we use the bindings from a snowboard. The foot joints are fixed in such a manner that the toes and heels can move freely by 290 mm vertically and 145 mm horizontally but are constrained laterally (see Figure 4), so that the dorsal and plantar flexion of the joints are free. In order to detect a shift in weight, each footpad is equipped with 2 Flexi-Force pressure sensors, made by NITTA. The FlexiForce is a pressure sensor made from a pressure-sensitive ink. It consists of a special, proprietary piezoresistive material sandwiched between two pieces of flexible polyester with silver conductors on each side. The FlexiForce sensor is basically a resistor whose conductance varies linearly with force under an applied load. With no force applied, the resistance is of the order of mega ohms; as the applied force increases, the output resistance drops, eventually reaching about 10 kΩ or lower, depending on the application. The sensors are placed under the heels and toes, based on the assumption that the patient puts his or her weight on these when the foot is placed on the stairs. The output format is 0–1023 digital data proportional to 0–5.0 volts of the sensor using an A/D converter.
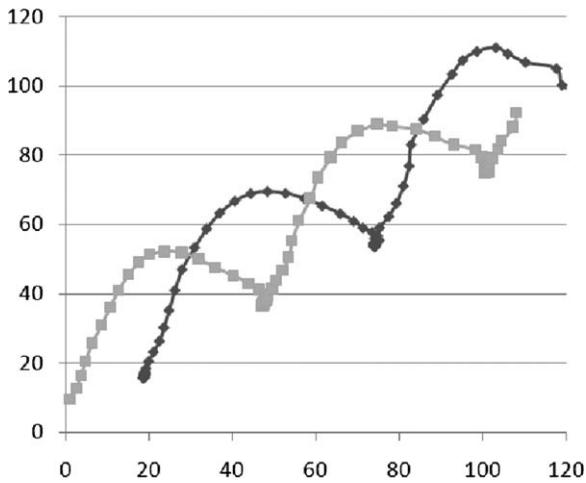
Figure 5. *Original trajectories of both feet in climbing stairs.*

Figure 6. *Trajectory of both feet in climbing stairs.*

3   Gait Rehabilitation for Climbing Stairs

We developed a gait training system for climbing stairs using the GM5. In climbing stairs, patients are expected to acquire the skill of moving their foot up from a lower step, and placing it safely on the next step. By repeating this motion, the body moves up. In this process, the $180°$ phase-shifted movement of the body and voluntary walking skills should be learned.

The GM5 was first used to follow the trajectories of the feet of healthy individuals. The trajectories of the ankles of three healthy individuals climbing stairs were recorded using a motion capture system (Stereo Labeling Camera, made by CyVerse), which can capture position every 33 ms. Each of the healthy individuals had no bad walking habits and climbed the stairs 10 times while keeping pace with the tempo of a metronome set at 1.0 Hz. Although the trajectory of the footpad should be obtained from the position of the heel, which is the first part of the foot to come into contact with the step, we used the trajectory of the ankle because it was difficult to attach the marker of the motion capture system to the heel.

Figure 5 shows a typical example of the captured data. To obtain the trajectory for one cycle of the footpads of the GM5 (see Figure 6), we considered the following: (1) When climbing stairs, it is important that the feet
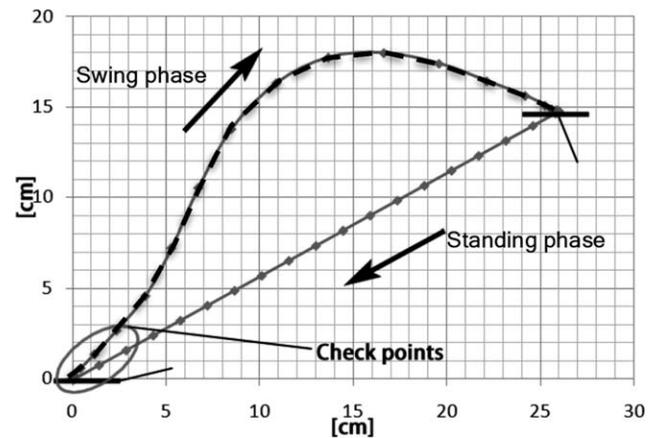
avoid colliding with the stairs, and (2) The impression of standing on a rigid floor is conveyed. The trajectory is composed of a swing phase (the dashed curve in Figure 6) and a standing phase (the continuous line in Figure 6). For the swing phase, the absolute positions from averages of the data from the three individuals were used. Since the user's waist is not fixed in this system, the trajectory of each foot is based on its absolute position, not its position relative to the waist. The data were smoothed to avoid instabilities in the control. In the smoothing method, the position data were manually adjusted to avoid sudden changes in acceleration and velocity. Also, to avoid any unnatural acceleration, which can cause the unwanted impression of standing on an unstable floor, the footpads move back at a constant speed in the standing phase. In addition, the footpads rest for 0.166 s before the step up, which is double the support time on the RS (real stairs). To include this, the five check points that have the same position on the footpad, shown in Figure 6, are inserted into the trajectory data. This adds to the impression of climbing solid stairs.

In this study, two training modes were developed. One was the enforced gait mode (EF mode for short), in which the patient's feet involuntarily follow a prerecorded trajectory. The EF mode can be used for novices or less active patients. Using EF mode training, the patient can acquire the $180°$ phase shift of the walking motion. The other is the interactive gait mode (Int mode for short), in which the footpads move depending on

the patient's intention to climb the stairs. The Int mode can be used for more active patients. By using the Int mode, they can acquire not only the 180° phase-shifted movement of the body, but also the skill of voluntary walking. To detect the intention to climb the stairs, the movement of the patient's body to the upper step or the recoil motion of the lower foot should be measured. A motion capture system can be applied for this purpose. However, the system may detect motion after the foot has taken off from the footpad and the latency in our motion capture system, due to the fact that the 33 ms between data points may cause the GM5 to respond slowly. Therefore, to detect the motion before the foot had left the footpad, pressure sensors were used to detect the shift in weight. The change in load on the lower step was used to detect the intention to climb in this system. When the patient's weight was applied to the footpad on the upper step, the values of the pressure sensors on the lower step became almost zero. Then the footpad moved forward with the foot in the swing phase and the other footpad began to move backward. Therefore, the timing of the movement is determined by the patient. This function is almost the same as the semivoluntary gait mode, which was proposed in our previous paper (Yano, Tamefusa, Tanaka, Saitou, & Iwata, 2010). The pressure sensor data, however, are also used to detect the load on the footpad in the swing phase. Since the patient tends to leave some of his or her weight on the footpad in the swing phase, information of this unwanted load in the swing phase should be fed back to the patient. We think the most effective way to communicate this tendency is to alter the speed of the footpads. When some weight (more than 0 kg using the FlexiForce pressure sensor) is detected, the velocity of the footpads is decreased every 0.05 s by 104 mm/s in the horizontal direction and by 60 mm/s in the vertical direction. The rate of change in the velocity of the footpads was determined by the rate at which most patients could sense it. If the patient pulls his or her foot up and the remaining weight falls to zero, the speed of the footpads is increased to the original speed at the same rate. This event persuades the patient not to put weight on the footpad. In this way, a patient using the Int mode can acquire the skill of voluntary walking.

Table 1. *Demographic Data of Healthy Participants*

	Age	Height (cm)	Weight (kgf)
Subject1	22	175	67
Subject2	24	170	54
Subject3	23	175	74
Subject4	24	173	50
Subject5	24	165	80
Subject6	22	171	60

By using these modes, the patient can learn the motion of healthy individuals for climbing stairs.

4   Evaluation of Training to Climb Stairs

4.1 Experimental Conditions

Experiments were conducted in order to assess the effectiveness of the proposed methods for training patients to climb stairs. We compared training in the EF mode, the Int mode, and real stair climbing (RS mode for short). To evaluate walking, the motion of the body, muscle activation, and the shift in weight can be considered. Since the frames and actuators partially hide the patient's body, video analysis could not be used in this study. From the observations of physiotherapists, no unusual motion was observed during climbing stairs on the GM5. Therefore, EMGs of the subjects' muscles were measured to record their activity. Also, the pressure applied to both feet was measured to observe the timing of the shift in weight of the subject.

The pressure applied by the feet and the muscle activities of the subject were measured during each trial. We used F-Scan II sensors to measure the weight shift between the feet. The principle of measurement of the F-ScanII sensor is same as that of the FlexiForce sensor. The sensors are 0.15 mm thick and were placed beneath each foot. They can measure the pressure at 955 points in real time. The time resolution of the measurement was 5 ms in this experiment. An electromyograph (EMG), the MEG-6108 manufactured by Nihon Kohden, which has 16 EMG channels, was used for measuring the muscle activity of the subjects. We meas-
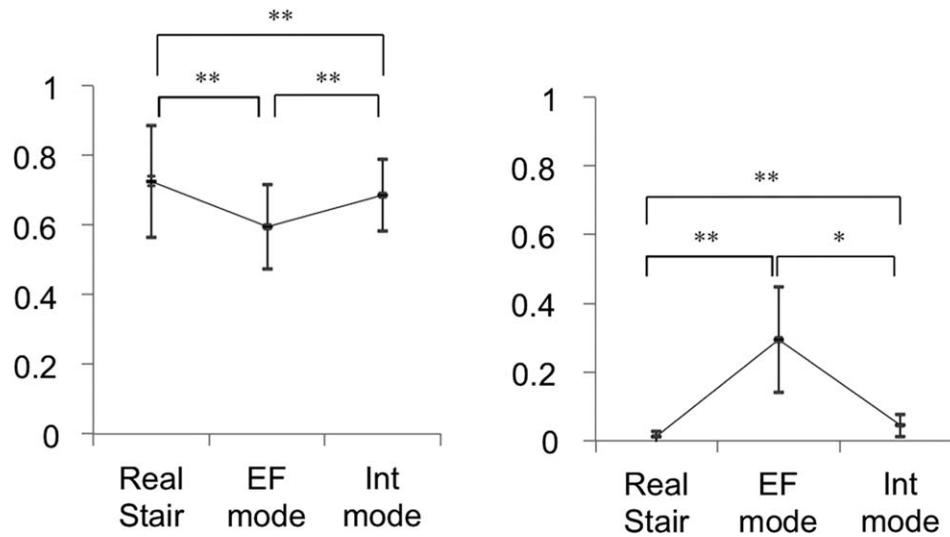
Figure 7. *Comparison of the average load of a typical healthy individual in each mode (Left: standing phase. Right: swing phase. ** refers to the 1% significance level, * refers to the 5% significance level).*

ured the electromyograms of four types of muscle. These were the tibialis anterior muscle, the hamstrings, the medial vastus muscle, and the gastrocnemius muscle. The tibialis anterior muscle is attached to the tibia, mostly located near the shin. This muscle is in charge of dorsiflexion of the foot. It allows the toe to be pulled up and held in a locked position. The hamstrings are the muscles at the back of the thigh, which are in charge of the extension of the thigh and rotation of the lower thigh. They decelerate the lower limb of the swinging side when the gait switches from the swing phase to the standing phase. The medial vastus muscle is a knee muscle, which absorbs the shock when the foot lands on the ground. The gastrocnemius muscle is a sural muscle that is in charge of flexion of the lower thigh and the springing motion used for acceleration (Rose & Gamble, 1994).

Three trials were conducted with six healthy individuals in their 20s, whose weights were between 50 and 80 kg (see Table 1).

A trial was also conducted on a 60-year-old hemiplegic patient weighing 80 kg who has had left hemiplegia, Brunnstrom Stage V, for the past 10 months. He can undertake a circumduction gait without canes or ankle foot orthotics in climbing stairs. He visits the hospital

regularly as an outpatient for rehabilitation. In the EF and Int modes, a rest for 10 s, climbing stairs for 30 s, and a further 10 s rest were applied. The RS task was limited to 11 steps by the length of the cable to the EMG used to measure the muscle activity of the subject. The size of the step in each mode was 260 mm deep and 150 mm high, the same as steps in an actual hospital staircase. The cadence in the stair-climbing test was 60 steps/min in the EF mode, 40–52 steps/min in the Int mode, and 43–65 steps/min in the RS mode. Considering the safety of the training, two physiotherapists who were familiar with the system determined these cadences based on the patients' experience. Since all the subjects were novice users of the GM5, they walked on it for about 90 s before the experiment. The following sections describe the results and analysis.

4.2 Results in Stair Climbing

4.2.1 Foot Pressure of Healthy
Individuals. The average loads applied to the F-Scan II sensor by each foot during each task were compared. The left side of Figure 7 shows the results in the standing phase. The time series variation of the load for each subject was calculated by integrating the pressure data of all

points from the F-Scan II sensors. The average load data were obtained by averaging the integrated time series data. Then the average loads were obtained by normalizing the average load data by the body weight of each subject. There are significant differences between the modes at the 1% significance level, and the average load of the EF mode is the lightest in the standing phase. On the other hand, the average load of the EF mode in the swing phase is the heaviest (right side of Figure 7). This was due to the patient keeping his or her weight on the footpad during the swing phase. The average load in the swing phase in the Int mode is smaller than that in the EF mode, because the foot was lifted from the footpad in this mode. Also, the subjects held on to the safety frame in the EF and Int modes, so that their body weight was distributed between the footpads and the safety frame. These caused smaller average loads in the standing phase than in the RS mode. Figure 8 shows a typical time series variation of the weight shift of a healthy individual weighing 67 kg in each mode. Each value is normalized by the individual's weight. The gray colored regions are the double support phases. In the RS mode, there are pressure peaks when the subject lands on or leaves the ground. In the EF mode, the pressure peak on landing is delayed. As mentioned above, some weight remains on the footpad in the swing phase. In the Int mode, the timing of the pressure peaks at landing on and leaving the ground are similar to the timing in the RS mode. Also, in the Int mode, there is no load on the footpad in the swing phase because the subjects try to avoid deceleration of the footpad. These results show the effectiveness of the algorithm for detecting the time at which the foot leaves the lower step in the Int mode.

4.2.2 EMG of Healthy Individuals.

The muscle function was measured with EMGs to evaluate the effect of the training to climb stairs.

Figure 9 shows typical examples of the time series variations of the normalized root mean square (RMS) values of the electromyograms of the leg muscles of the six healthy subjects climbing stairs. The data were calculated using the following sequence. (1) RMS values were obtained from the time series variations of the EMG raw data for each muscle. (2) The RMS values were normal-
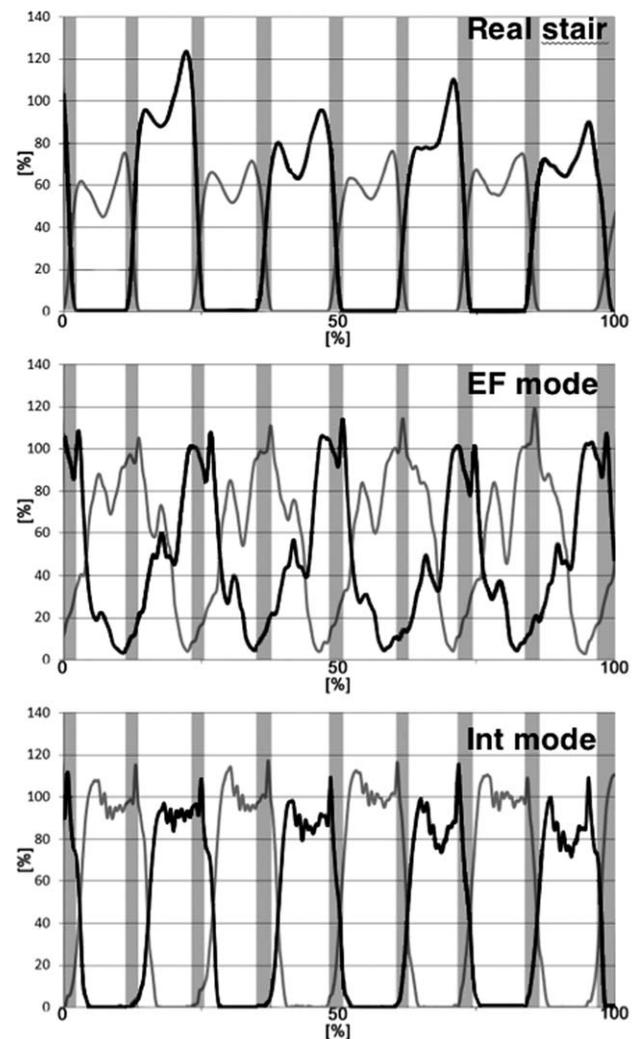

Figure 8. *Time series variation of weight on each foot of a healthy individual climbing stairs (Top: Real Stairs; Middle: EF mode; Bottom: Int mode; Gray: left foot; Black: right foot).*

ized by the subject's maximum voluntary contraction for each muscle in the RS mode. The regions in Figure 9 represent, from the left, the double support phase (left gray colored area), the standing phase of the right leg, the double support phase (right gray colored area), and the standing phase of left leg. The gray line shows the EMG of the left leg and the black line the EMG of the right leg. The tibialis anterior and gastrocnemius muscles in the EF mode are inactive for the whole period since the feet are moved involuntarily without dorsiflexion or flexion of the feet.
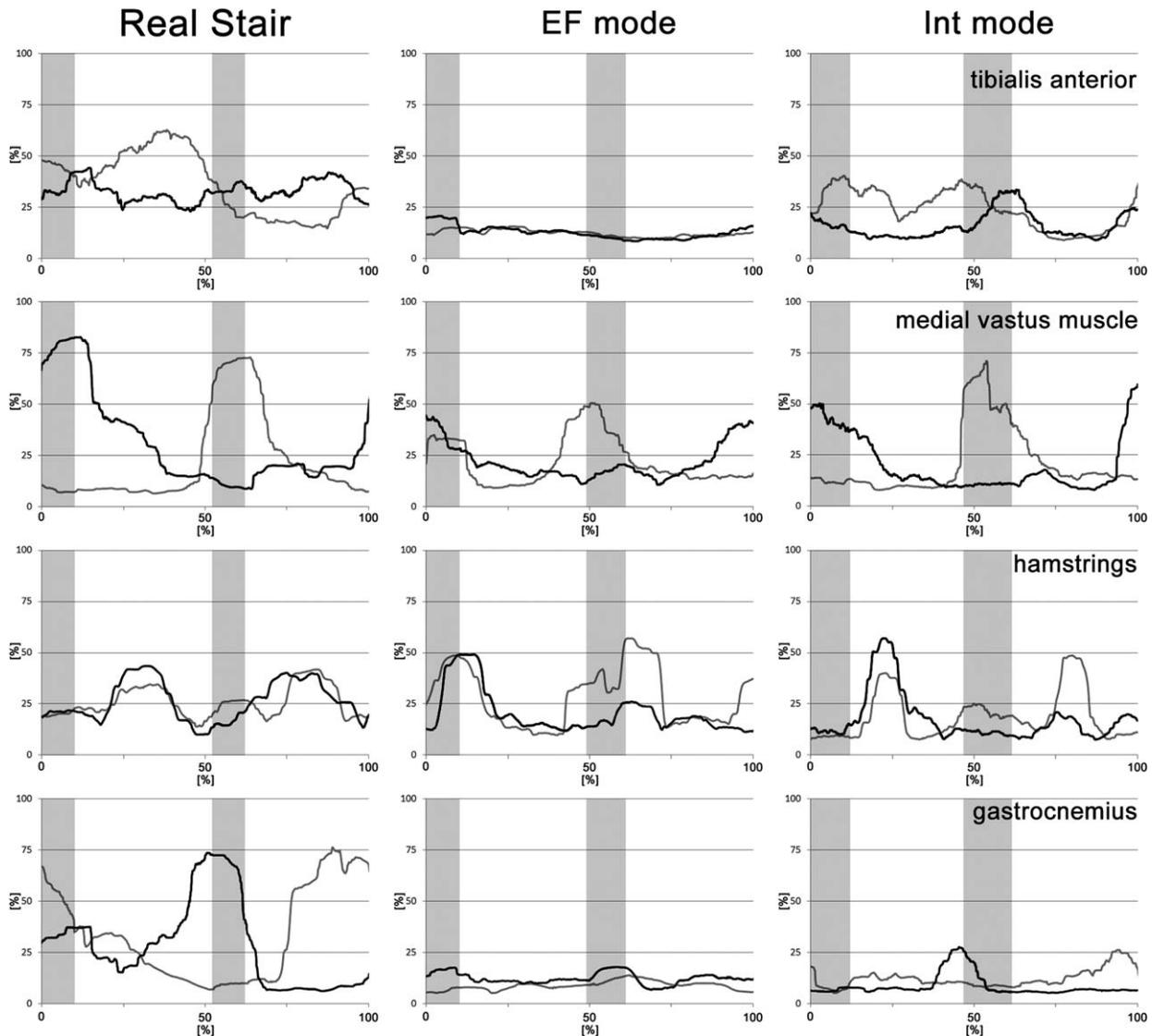
Figure 9. *Electromyography of muscles in the legs of typical healthy subjects in climbing stairs (Gray: left leg, Black: right leg).*

In the Int mode, the medial vastus muscle and the hamstring are less active than in the other modes. Although the intensities of the activity of all the muscles are lower than those in the RS mode, the timing in the Int mode is similar to the RS mode. Since the subjects held on to the safety frame during this test, they were able to walk easily on the GM5, so that the muscle activities for these are lower. From these results, we can conclude that the subjects were able to climb the stairs voluntarily in the Int mode.

Table 2 shows the results of comparing the average muscle activity of the healthy subjects during stair climbing. The average muscle activity was calculated by averaging the normalized RMS values. The values in parentheses refer to the standard deviation for each activity. The one-way ANOVA found a statistically significant difference in each muscle. Post hoc tests using the Scheffe Multiple Comparisons test reveals significant differences in each muscle. The table shows the same tendencies described above for Figure 9.

Table 2. *Comparison of Average Muscle Activity of Healthy Individuals Climbing Stairs in Each Mode (Units: Percent)*

Tibialis anterior	Real stair	>	Int mode	>	EF mode
	40.0 (9.2)	**	25.7 (6.2)	**	19.0 (7.7)
Medial vastus muscle	EF mode	=	Real stair	>	Int mode
	28.4 (13.2)		27.6 (6.8)	*	22.1 (8.2)
Hamstrings	EF mode	=	Real stair	>	Int mode
	32.8 (16.1)		31.7 (7.7)	**	25.4 (7.0)
Gastrocnemius	Real stair	>	Int mode	=	EF mode
	22.2 (9.2)	**	11.2 (8.2)		10.2 (9.5)

NOTE: * refers to the 1% significance level. ** refers to the 5% significance level.
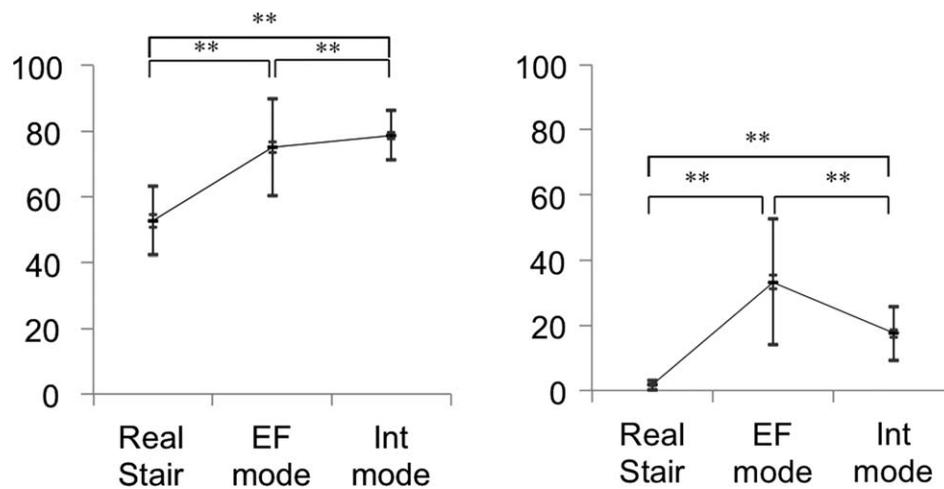

Figure 10. *Comparison of the average load from both feet of a hemiplegic patient in each mode (Left: standing phase. Right: swing phase. ** refers to the 1% significance level).*

4.2.3 Foot Pressure of the Hemiplegic Patient.

Figure 10 shows a comparison of the average load in each mode applied to the F-Scan II sensors by each foot during the task. The average loads are for both feet and are normalized by the patient's body weight. There is a significant difference between the Int mode and the other modes at the 1% significance level, with the average load in the Int mode being the heaviest.

In Figure 11, the time series variations of the weight applied by each foot of the patient in climbing stairs are shown. In the RS mode, there are peaks at landing on and taking off from the step. In the EF mode, the left leg does not leave the footpad. In the Int mode, the left leg is raised from the footpad for a longer time compared to the EF mode. Also in the Int mode, there is very little load on the right footpad in the swing phase due to the patient attempting to avoid deceleration of the footpad. The results show that the patient is able to learn the motion of the right foot correctly using the Int mode.

We compared the ratio of the time spent in the standing phase between the left (Period A) and right (Period B) legs in each mode. Periods A and B are the times for which the pressure sensors detect that the weight is on the left or right foot, respectively. The measurement was
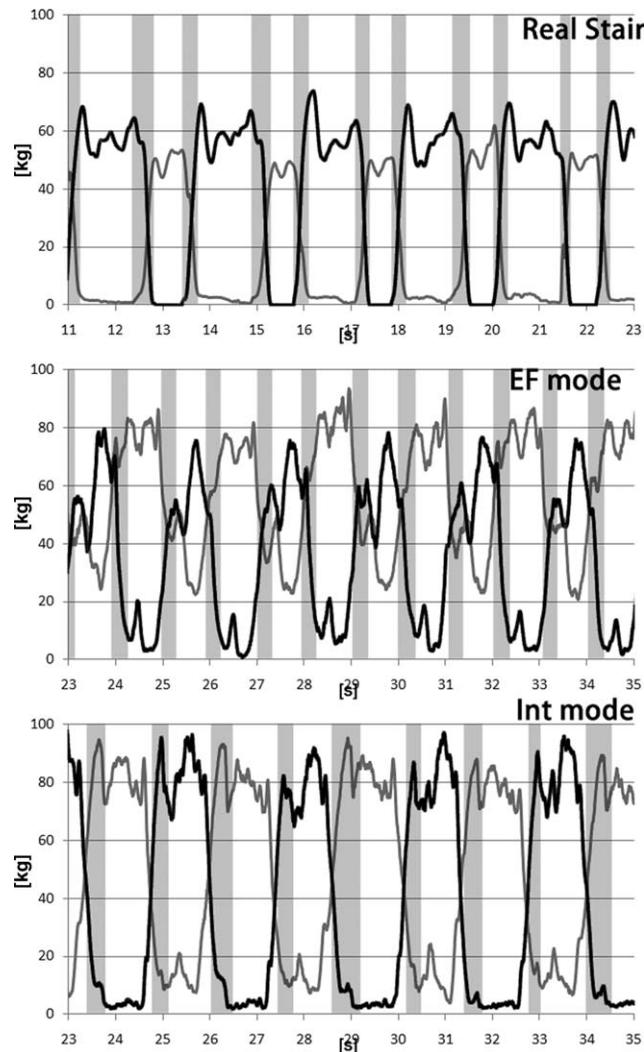
Figure 11. *Time series variation of the weight on each foot of the patient in climbing stairs (Top: Real Stairs, Middle: EF mode, Bottom: Int mode; Gray: left foot, Black: right foot).*

Table 3. *Comparison of the Average Ratio of Period A to Period B for the Patient in Each Mode*

Int mode	>	EF mode	>	Real stair
1.063 (0.048)	**	1.005 (0.012)	**	0.877 (0.046)

NOTE: ** refers to the 1% significance level.

one-way ANOVA and the post hoc tests using the Scheffe Multiple Comparisons test. In the RS mode, because of the circumduction gait of the patient, his left leg moved a greater distance than his right leg, so that the duration of the swing phase of his left leg was longer than that of his right leg. As a result, the standing phase of his left leg was shorter than that of his right leg. This is the reason for the small ratio in the RS mode. In the Int mode, since the patient had to keep his weight on his left foot during the swing phase of his right leg, it required a conscious effort to keep his body weight on his left foot. In addition, shifting his weight from his right leg to his paralyzed left leg takes more time than the inverse operation. Since a double support phase before a standing phase of his left leg became longer than that of his right leg, the Period A became longer than the Period B. Hence the ratio is more than 1.

The above results demonstrate that the EF mode can be used to the 180° phase-shifted movement of the body. However, since the foot is moved involuntarily by the footpad, the patient tends to lean against the footpad even when his foot is in the swing phase. On the other hand, in the Int mode, the patient can learn not only the 180° phase-shifted movement of the body, but also to raise his leg voluntarily. Thus, this mode is more beneficial than the EF mode.

4.2.4 EMG of the Patient in Stair Climbing.

Tables 4 and 5 show comparisons of the average muscle activity in the patient's legs during stair climbing. The values in parentheses refer to the standard deviation of each activity. The result of the one-way ANOVA and the post hoc tests using the Scheffe Multiple Comparisons test are also shown in these tables.

Figure 12 shows EMGs of the patient's legs in climbing stairs. The regions in the figure represent, from the

acquired for two steps, and so includes the double support phase. Thus, if the time for which the weight is on the left foot is greater than that on the right, the ratio is larger than 1.

The average ratios for the three modes are shown in Table 3. The values in parentheses refer to the standard deviation of each activity. In healthy individuals, the ratio of each mode is usually about 1.0 and we found no significant difference between modes. However, for the patient, a significant difference between the RS mode and the other modes was found from the result of the

Table 4. *Comparison of the Average Activity in the Muscles of the Left Leg of the Hemiplegic Patient in Climbing Stairs (Units: Percent)*

Left leg	Tibialis anterior	Int mode	=	Real stairs	=	EF mode
		40.0 (0.8)		37.6 (3.1)		31.8 (1.6)
	Medial vastus muscle	Real stairs	>	EF mode	=	Int mode
		25.0 (4.4)	*	16.4 (0.9)		14.9 (0.6)
	Hamstrings	Real stairs	=	Int mode	=	EF mode
		14.1 (3.4)		13.0 (1.2)		10.3 (1.1)
	Gastrocnemius	Int mode	>	EF mode	=	Real stairs
		35.1 (7.1)	*	25.8 (10.5)		17.6 (3.1)

NOTE: * refers to the 5% significance level.

Table 5. *Comparison of the Average Activity in the Muscles of the Right Leg of the Hemiplegic Patient in Climbing Stairs (Units: Percent)*

Right leg	Tibialis anterior	Int mode	=	Real stairs	>	EF mode
		48.3 (0.8)		45.1 (9.7)	*	34.9 (3.4)
	Medial vastus muscle	Real stairs	>	Int mode	=	EF mode
		25.3 (5.5)	*	14.6 (0.5)		12.6 (0.3)
	Hamstrings	EF mode	=	Real stairs	=	Int mode
		25.0 (0.4)		24.3 (7.0)		22.5 (0.7)
	Gastrocnemius	Real stairs	=	EF mode	=	Int mode
		13.9 (2.7)		10.0 (3.8)		9.1 (0.6)

NOTE: * refers to the 5% significance level.

left, the double support phase (left gray colored area), the standing phase of the right leg, the double support phase (right gray colored area), and the standing phase of the left leg. The gray and black lines are the EMGs of the left and right legs, respectively.

The gastrocnemius muscle is activated in all modes, particularly in the Int mode, where the activity is significantly higher than in the other modes. This is because the patient tends to raise his paralyzed leg higher in the Int mode in order to ensure the movement is detected by the sensors. This result is consistent with the result given in Table 4. The medial vastus muscle is activated in the 180° phase-shifted movement of the body in the Int mode. The medial vastus muscle in the RS mode is more activated than the other two modes, since the foot is guided to a soft landing by the footpad in the other two

modes. These results show that not only can the patient acquire the 180° phase-shifted movement of the body, but also voluntary walking skills using the Int mode.

Although there is no significant difference between each mode for the average activity of the hamstrings and the tibialis anterior of the left leg in Table 4, it is less active in the EF mode than in the other modes in Figure 12. Since the footpad involuntarily moves the left foot in the EF mode, the patient did not dorsiflex or extend the thigh but continued to exert pressure on the footpad.

5   Discussion

We have developed the EF mode and the Int mode for climbing stairs, which proved effective in training patients to acquire 180° phase-shifted movement of the
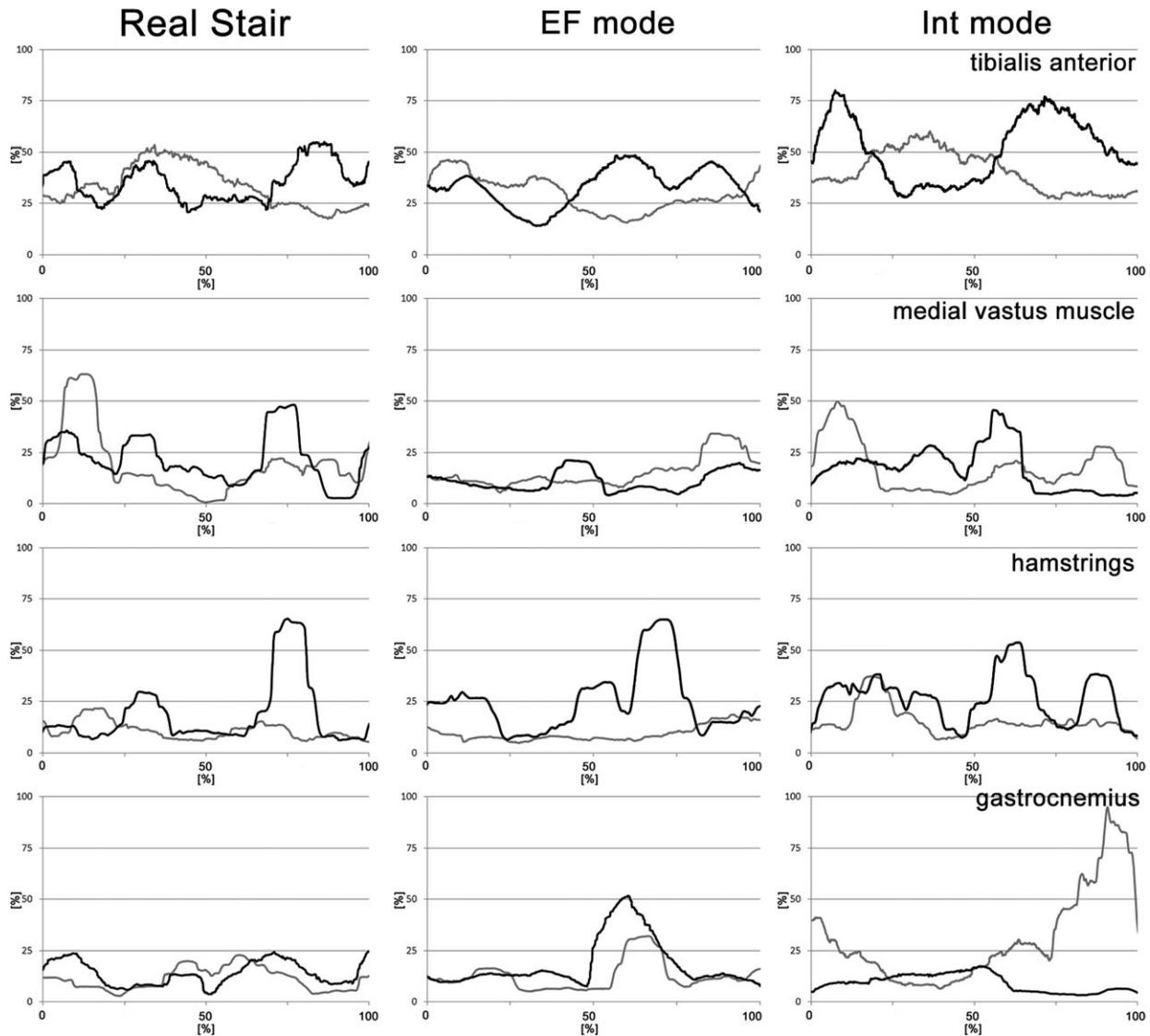
Figure 12. *Electromyography of the patient's legs in climbing stairs (Gray: left leg, Black: right leg).*

body. In the Int mode, interactive support and an interactive warning (deceleration when the weight remained on the footpad during the swing phase) proved to be effective in developing voluntary walking skills. The duration and difficulty differ between climbing stairs and walking on a flat surface. The duration for normal walking on a flat surface is longer than that for climbing stairs. The velocity is almost constant in walking on a flat surface, whereas in climbing stairs an up-and-down movement of the feet is required. To acquire these skills, interactive training is needed. In this study, the patient's

feet were moved along a trajectory, and an interactive support method was realized. Also, the interactive warning method achieved through deceleration when pressure on the footpad was detected in the swing phase was implemented. However, since climbing stairs is an advanced part of the training process in gait rehabilitation, the training should be done in stages. We think the following steps should be used in practical rehabilitation. First, the patient should learn the whole body motion using the EF mode. Secondly, he or she should learn to climb stairs voluntarily using the Int mode. After this,

the patient should learn the motion for real steps by removing the bindings on the footpads. We call this mode the LI mode. In this mode, by adding foot position sensors, the footpads can follow the motion of the feet, so that the patient can move his or her feet freely.

However, a limitation of our system is that it should only be used for patients who can walk unattended and have a low risk of heart problems.

There is also a possibility to change the walking pattern if the patient uses his or her hands for support. In this case, the biomechanics of the upper trunk have to be considered. In this study, the frame of the system can be used in the same way as the handrail on a staircase. The cadence used in climbing stairs was 1.0 Hz/step, which is slower than the usual speed of healthy individuals. The dynamics of the upper trunk have an important influence when the walking speed is increased. Consideration of the biomechanics of the upper trunk is left for future work.

6   Conclusion

In this paper, a gait rehabilitation system with a compact LI for training patients to climb stairs was developed. An enforced gait mode and an interactive gait mode were implemented for the training. In the experimental evaluation, which involved tests on six healthy subjects and one hemiplegic patient, it was shown that the correct walking motion could be learned with our proposed system. The EF mode can be used for novice/less active patients to learn the $180°$ phase-shifted movement of the body and the Int mode can be used for more active patients to learn voluntary walking. The key features of our training system are (1) two 2-DOF manipulators which move the patient's feet along a prerecorded trajectory obtained from healthy individuals; (2) the trajectory of the footpad is composed of absolute position data of the foot in the swing phase, and a constant return speed for the standing phase, (3) implementation of a double support phase for giving the impression of walking on a rigid surface, and (4) using pressure data from beneath the foot in order to detect the commencement of a step, and also to detect the remaining weight on the footpad in the swing phase.

We are now developing an interactive method for descending stairs. Also, the long-term effects of our rehabilitation method will be investigated. In future work, we plan to develop a combination of walking on a flat surface and climbing stairs. Since the stair climbing modes can be used to increase the training workload, these might shorten the training periods required for normal walking on a flat surface. We also plan to provide a more complex and realistic training environment with an immersive projection display.

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