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Leg-dependent Force Control for Body Weight Support by Gait Cycle Estimation from Pelvic Movement

Takao Watanabe, Eiichi Ohki, Yo Kobayashi and Masakatsu G. Fujie

Abstract— **A movable body weight support (BWS) system for overground walking has limitations with respect to power and space for actuators. To solve this problem, a novel method by which to estimate the gait cycle from pelvic movement and feed forward control for leg-dependent force control are proposed. Based on an experiment on gait cycle estimation, a method of estimating heel contact timing from pelvic rotation is proposed. In addition, based on an experiment on leg-dependent force control, the developed feed forward control method is evaluated based on the vertical floor reaction force.**

I. Introduction

THE ability to walk is fundamental in order to live an independent life and maintain physical and mental health. Body weight support (BWS) systems applied to patients who are undertaking rehabilitation from spinal cord injury, stroke, or a fracture of the femur, while walking on a treadmill or over ground can facilitate rehabilitation.

There are a number of different methods for providing the unloading force for BWS. According to a previous study [1], a number of commercially available BWS systems use a winch [2], a counter weight [3], or an elastic spring [4], and lift the body weight from above by means of a harness connected to a wire. These systems have one significant shortcoming in that once the amount of support is set, it cannot be adjusted during the gait cycle. Moreover, due to friction and the nature of the spring or mass used in these systems, the support force varies in an undesirable manner.

In order to solve this problem, several motor-actuated devices have been developed. Glauser *et al*. [5] designed and constructed a partial BWS system that constantly monitors the provided support force and takes immediate action to maintain the support force to be very close to a specified support force profile. The support force can be maintained constant during the entire gait cycle or can be varied in a controlled manner within the gait cycle in order to facilitate walking and to alter various gait parameters. Based on the gait cycle, the desired support level can be changed simultaneously with events or conditions that occur during walking, such as heel strikes, single limb support, or double limb support. By monitoring thresholds on the vertical ground reaction forces, the support force variation is synchronized with the stride of the patient. To allow a faster transition from low to high or high to low supporting force, a time optimal control, featuring a high motor shaft velocity and duration adaptation, replaces the error feedback term for a short time.

However, in the movable BWS system for overground walking, there are power and space limitations with respect to the actuators. As one solution to this problem, feed forward control is effective even when using an actuator with limited performance of motor shaft velocity and output torque.

The purpose of the present study is to develop leg-dependent force control for body weight support by gait cycle estimation for feed forward control. The proposed approach involves the application of a pelvic support mechanism for body weigh support to a sensing device for gait cycle estimation.

Many BWS systems lift the body weight from above by means of a harness connected to a wire. These systems allow unrestricted movement only in the vertical direction but restrict pelvic rotation, pelvic obliquity, and horizontal translation of the pelvis [6]. Recently, it has been demonstrated that patient-specific torso motions may be useful for generating desired gait patterns [7], and that pelvic movement also plays an important role in normal locomotion. Figure 1 shows the developed BWS system with pelvic support, which allows for natural walking motions of the pelvis and sensing of pelvic rotation. The developed system consists of the BWS system (A) and the pelvic support mechanism (B). This system will be integrated into a mobile base system to develop a movable BWS system for overground walking.

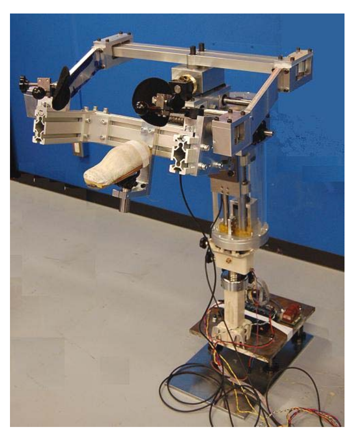

Fig. 1. Developed BWS system (A) and pelvic support mechanism (B)

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Takao Watanabe is with the Graduate School of Advanced Science and Engineering and the Faculty of Science and Engineering, Waseda University, Tokyo 169-8555, Japan (e-mail: takawo@toki.waseda.jp).

Eiichi Ohki is with the Graduate School of Advanced Science and Engineering, Waseda University, Japan.

Yo Kobayashi is with the Faculty of Science and Engineering, Waseda University, Tokyo, Japan.

Masakatsu G. Fujie is with the Faculty of Science and Engineering, Waseda University, Tokyo, Japan.

The remainder of the present paper is organized as follows. Section II describes the equipment used in the experiment. Section III describes an experiment on gait cycle estimation. Section IV describes an experiment on the evaluation of leg-dependent force control. Section V presents a summary and outlines future research.

II. Mechanical Setup

A. BWS system

In the partial BWS system mentioned above, linear motor with a peak force of 3,300 (N) [5] is used. As another example, the Lokolift system is composed of a passive elastic spring element to take over the main unloading force and an active closed-loop controlled electric drive to generate the exact target force. Its drive unit can generate a peak force of up to 1,500 (N), and the desired subject force is adjusted within 100 (ms) at an unloading step of 20 (kg) [1].

Figure 2 shows the developed BWS system. Unlike the BWS systems introduced above, this BWS system unloads body weight from below because this system will be integrated into a mobile base system to develop a movable BWS system for overground walking.

This system unloads body weight on a pedestal attached to a motor-actuated device (Fig. 2). The mounted load cell (UNIPULSE, RSCM, Rated capacity: 1,000 (N)) measures the unloading force from the pedestal that supports the body weight. The DC motor (Maxon, EC 45, Brushless DC, Rated power: 250 (W), Voltage rating: 24 (V), Max. rotating speed: 1,660 (rpm), Reduction gear ratio: 4.3), which is compact and has high torque, rotates a ball screw via reduction gears and moves the plates connected to the pedestal. The peak force is approximately 700 (N). Constant force control of this system has been investigated [8].

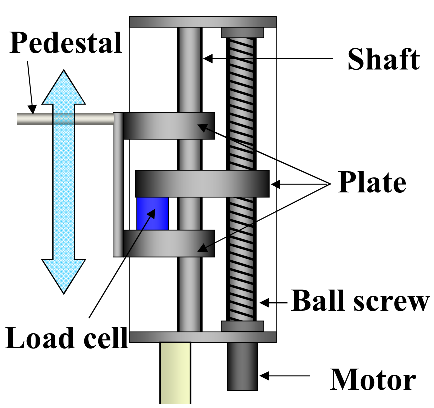

Fig. 2. Components of the BWS system

B. Pelvic support mechanism

As mentioned above, the pelvic motion significantly affects the human gait. Human gait locomotion consists of six determinants, which are pelvic rotation, pelvic tilt, stance phase knee flexion, knee mechanisms, foot mechanisms, and lateral displacement of the pelvis [9]. Three out of six of these determinants are pelvic-related movements, which highlights the importance of pelvic control during gait rehabilitation.

Figure 3 shows the developed pelvic support mechanism, which can be mounted on the BWS system. This mechanism supports the pelvis by holding the left and right parts of the anterior superior iliac spine and the ischial bone and has three passive rotational DOFs, as defined in Fig. 3. This mechanism allows rotation about the yaw-axis, bending movement about the pitch-axis, and hip rotation about the roll-axis, which allows for natural walking motions of the pelvis. The encoder that is connected to the shaft, which is the center of rotation of the parallel linkage, can measure the angle of pelvic rotation. Furthermore, the DC motor (Maxon, EC 40, Brushless DC, Rated power: 120 (W), Voltage rating: 24 (V), Max. rotating speed: 840 (rpm), Reduction gear ratio: 12.3) attached to the pelvic support mechanism can control the swing motion of the pelvis.

In the present study, this pelvic support mechanism is used not only for body weight support but also for sensing the gait cycle. The gait cycle can be estimated by a correspondence relation between pelvic rotation and the gait cycle because both pelvic rotation and the gait cycle are continuous motions during walking.

In order to verify that this pelvic support mechanism could follow the rotation movement of the pelvis, a preliminary experiment to measure the deviation in rotation angle about the axis of yaw between pelvis and the pelvic support mechanism was conducted. We measured the positions of right and left iliac bone of pelvis and right and left edge of pelvic support mechanism (Fig. 4) using a VICON 612 (VICON MOTION SYSTEM; sampling frequency: 100 (Hz), accuracy: less than 1 (mm)) when moving the pelvis repeatedly. The measured positions were converted into rotations with equation of (1). The result shown in Fig. 5 is typical for this experiment. As a result, the maximum deviation was less than approximately 1.0 (deg), which is considered to be negligible compared to the range of movement. Based on this result, we concluded that pelvis rotation can be measured by the encoder mounted on the pelvic support mechanism.

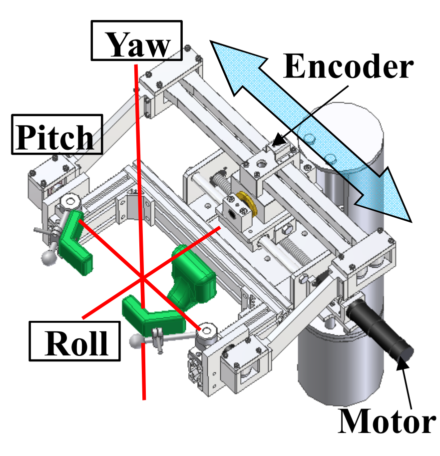

Fig. 3. Three-DOF of pelvic support mechanism and one active DOF for swing motion of pelvis

$$\theta_{yaw} = \tan^{-1}\left(\frac{x_{right} - x_{left}}{y_{right} - y_{left}}\right) \tag{1}$$

where x_{right}, y_{right} and x_{left}, y_{left} are right and left positions of pelvis or pelvic support mechanism.

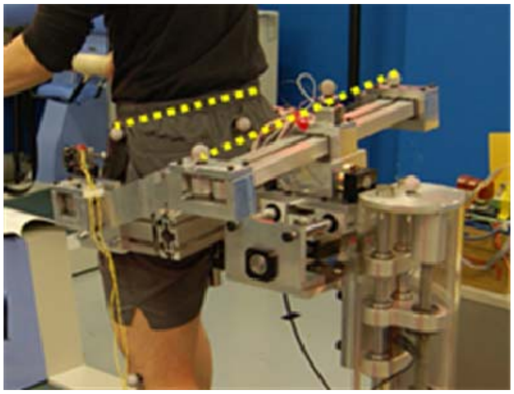

Fig. 4 Verification experiment for the following movement of the pelvic support mechanism

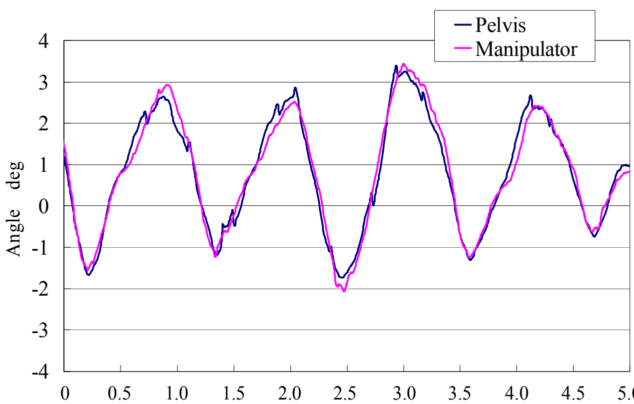

Fig. 5. Displacement of pelvis and pelvic support mechanism (manipulator)

C. Control Method

The objective of force control in this experiment is to change the unloading force depending on each leg. As mentioned above, feed forward control was adopted.

A block diagram of the gait-cycle-dependant force control is shown in Fig. 6. Here, θ_p is the pelvis rotation, F_{const} is the desired unloading force for the affected leg, $F_{mutable}$ is a variable by which to adjust the target force depending on the gait cycle, F_{target} is the target unloading force, F_r is the reference of the unloading force, F is the unloading force measured by load cell in developed BWS system, v_r is the motor rotating velocity.

The force control procedure is as follows. First, F_{const} is set to be F_{target}. During walking, the gait cycle is estimated from the measured pelvic rotation according to a prerecorded correspondence relationship between pelvic rotation and the gait cycle, and $F_{mutable}$ is derived. Then, $F_{mutable}$ is subtracted from F_{const}. The difference is F_{target}.

In order to realize this control system, it is necessary to establish a procedure by which to estimate the gait cycle from pelvic rotation and to define $F_{mutable}$ during the gait cycle. This is described in detail in the following section.

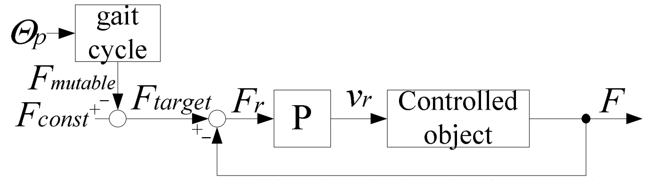

Fig. 6. Block diagram of the gait-cycle-dependant force control

Note that P is proportional controller. Its gain is set as 20 in consideration of rapidity and stability.

III. Experiment on Gait Cycle Estimation

A. Goals

In order to estimate the gait cycle from pelvic rotation, a correspondence relation between pelvic rotation and the gait cycle must be obtained in advance. In particular, since leg impact is maximized immediately after heel contact, the timing of heel contact must be detected from pelvic rotation in order to adjust the unloading force by $F_{mutable}$.

In this experiment, we simultaneously measured pelvis rotation and the gait cycle. In addition, we discuss how to estimate heel contact from pelvis rotation and define $F_{mutable}$.

B. Methods

Pelvis rotation was measured by the encoder mounted on pelvic support mechanism. Heel contact during the gait cycle was detected from the floor reaction force. The separated floor reaction force was measured using a floor reaction force meter (AMTI, OR6-7 2000, cutoff frequency for the low-pass filter: 10.5 (Hz)) and a separated treadmill (developed in our laboratory [10]), which allows separate measurement of the right- and left-side floor reaction forces (Fig. 7)

The subject of this experiment was a healthy, young male, who agreed with the objectives and the risks of the experiment. The subject's body weight was 65 (kg). The target unloading force was set as 250 (N), in order to compare the difference in results as compared to previous research on leg-dependent force control. The walking speed was chosen as 2.0 (km/h) based on the average walking speed of healthy, elderly people.

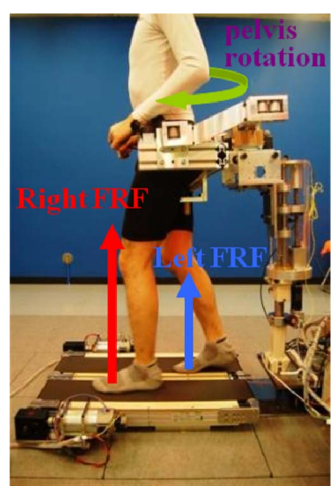

Fig. 7. Separated treadmill on force plates

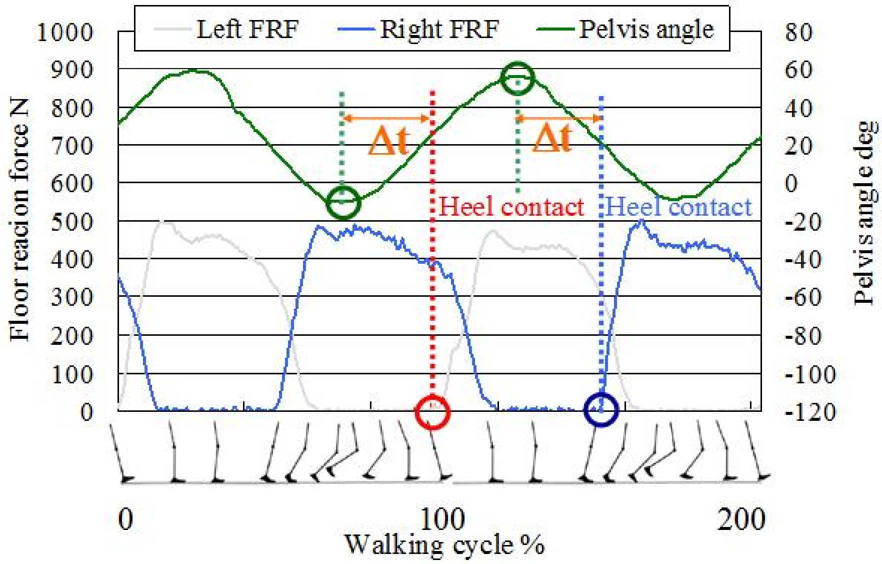

Fig. 8. Relationship between floor reaction force and angle of pelvis rotation

C. Results

Figure 8 shows the results of measured pelvis rotation and floor reaction force. Heel contact was detected based on the floor reaction force.

The results shown in Fig. 8 indicate that the angle of pelvic rotation was similar to a sine curve and had a local minimum and a local maxim before heel contact of the left and right legs, respectively. The time lag between the peak of the pelvis rotation and heel contact is defined as Δt. The average Δt for 20 gait cycles is shown in Table 1.

TABLE I
AVERAGE TIME LAGS BETWEEN LOCAL PEAKS OF PELVIS ROTATION AND HEEL CONTACT

	Ave. $\Delta t \pm$ SD s	Δt /gait cycle time %
Left	0.31 $\pm$0.04	24.7
Right	0.34 $\pm$0.07	27.2

D. Discussions

Based on the results of these experiments, the correspondence relation between pelvic rotation and the gait cycle at a certain walking speed was revealed. Since the standard deviation of Δt was considered to be negligible compared to the gait cycle time, Δt has repeatability under the condition where by the walking speed is constant. Therefore, it is possible to estimate heel contact with Δt and the timing at which the angle of pelvis rotation has a local peak at a constant walking speed. In this experiment, the subject was a healthy, young male without disorder of laterality, and the difference in Δt between the right and left sides was not notable. In case that this method is adapted to patients with disorder of laterality, the difference Δt between the right and left sides might be notable. However, this method is applicable even that case because Δt for right side and Δt for left side are measured respectively. In case that the walking speed is not constant, Δt is expected to vary over gait cycles. This problem remains to be solved. One possible solution is to obtain other information from the pelvic motion, such as swing motion or ZMP, to segment the gait cycle into several parts. If the gait cycle can be divided into several parts, such as immediately before or after heel contact, then the gait cycle can be estimated more precisely even for varying walking speeds.

Next, we define *Fmutable*. As an example, the unloading force for the affected leg is set as 250 (N), and the unloading force for the unaffected leg is set as 50 (N). The defined *Fmutable* is shown in Fig. 9. When switching from the affected side to the unaffected side, *Fmutable* must be increased so that the unloading force is decreased. Because of the characteristics of the actuator, the transition of the unloading force from 250 (N) to 50 (N) requires a time of less than Δt. Therefore, *Fmutable* must be increased gradually, as indicated by the slope in Fig. 9. When switching from the unaffected side to the affected side, *Fmutable* must be decreased so that the unloading force is increased. Because of the characteristics of the actuator, the transition of the unloading force from 50 (N) to 250 (N) require a time of more than Δt. Therefore, the transition of *Fmutable* is set as square wave simply.

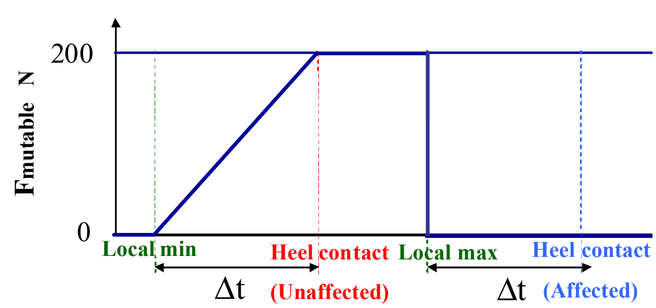

Fig. 9. Definition of mutable force for adjusting the desired force

IV. Experiment on Leg-dependent Force Control

A. Goals

The goal of this experiment is to evaluate the leg-dependent force control proposed. By measuring the vertical floor reaction force for the right and left legs during walking at body weight support, the unloading force was found to be adjusted properly.

B. Methods

In order to measure the vertical floor reaction force, the set-up used in the previous experiment was used in the present experiment. The subject of this experiment was the same as that of the previous experiment, and his body weight is 65 (kg). We assumed the case in which the target unloading force was 250 (N) for the affected leg (right leg) and 50 (N) for the unaffected leg (left leg). The walking speed was chosen as 2.0 (km/h), as in the previous experiment. Based on the results of the previous experiment, the time required for transition of the unloading force was set to be approximately 25 (%) of the gait cycle.

C. Results

Figure 10 shows the results for measured vertical floor reaction force when walking with constant 250 (N) BWS, and Fig. 11 shows the results for the measured vertical floor reaction force when walking with leg-dependent force control. Compared to the vertical floor reaction force when walking with constant 250 (N) BWS, the leg-dependent unloading force results in a different amount of transition for each leg. When walking with leg-dependent force control, the amount of vertical floor reaction force from the right leg (affected) was identical to that at constant 250 (N) BWS, and the amount of vertical floor reaction force from the left leg (unaffected) was larger than that at constant 250 (N) BWS by approximately 150 (N).

However, the shape of the vertical floor force reaction was different between the constant 250 (N) BWS and the leg-dependent force control. When walking with the leg-dependent force control, the shape of the vertical floor reaction force from the right leg (affected) had a second peak that was larger than the first peak.

D. Discussion

The results indicated that the vertical floor reaction force from the right leg (affected) was reduced by the same amount as for the constant 250 (N) BWS and that the vertical floor reaction force from the left leg (unaffected) was reduced by only 50 (N) from the body weight of the subject. Moreover, the leg-dependent force control appears to have functioned effectively. However, from the viewpoint of the shape of the vertical floor reaction force, the vertical floor reaction force from the right leg (affected) was not reduced in the same manner as the constant 250 (N) BWS. Here, we adopted feed forward control to transition the unloading force gradually in advance. The possible cause is that the transition of the unloading force by $F_{mutable}$ affected the walking movement, such as the walking pattern and body mass transitioning. As related research, the Lokolift system allows force variations of up to 20 (kg) within only 0.1 (s), which is faster than that of the actuator used in the present study, but there is no information available regarding its effect on walking movement. Further investigations are necessary in order to determine whether parameters such as timing or speed of transition of unloading force and gait patterns have a positive or negative effect on the outcome of the rehabilitation process.

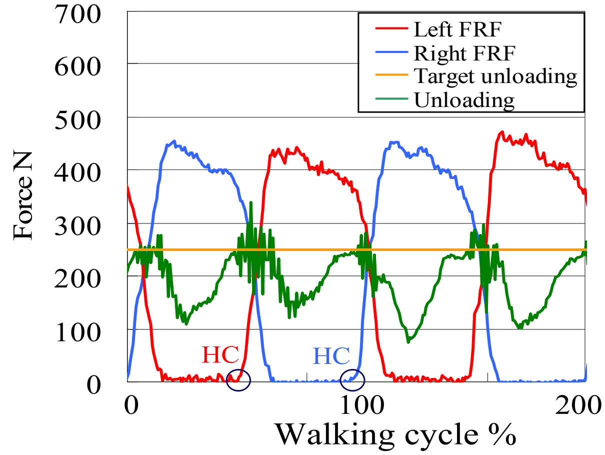

Fig. 10. Vertical floor reaction force when walking with 250 (N) BWS

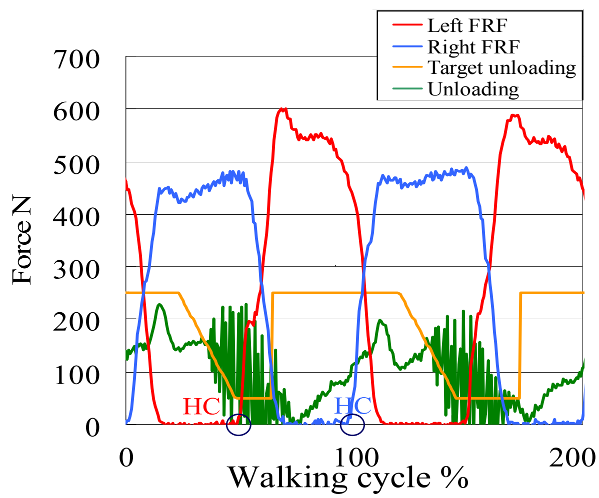

Fig. 11. Vertical floor reaction force when walking with leg-dependent force control (50 (N) BWS for the left leg, 250 (N) BWS for the right leg)

V. Conclusion and Future Research

A novel method by which to estimate the gait cycle from the pelvic movement and the feed forward control for leg-dependent force control are described in the present paper. Based on the experiments on gait cycle estimation, a method for estimating heel contact timing from pelvic rotation was proposed. Although the adaptation of this method remains to be demonstrated, other information determined from the pelvic motion, such as swing motion or ZMP, might be useful to divide the gait cycle and perform more precise estimation. Based on the experiments of leg-dependent force control, the developed feed forward control method was evaluated based on the vertical floor reaction force. In terms of the amount of vertical floor reaction force, the leg-dependent force control appears to have functioned successfully. However, from the viewpoint of the shape of the vertical floor reaction force, the vertical floor reaction force was not reduced in the same manner as the constant body weight support. Further investigations are necessary in order to determine whether parameters such as timing or speed of transition of the unloading force and gait patterns have a negative or positive effect on the outcome of the rehabilitation process.

Future research will focus on solving the remaining problems mentioned above and integrating the BWS system and the control method into a mobile base system that allows overground walking.

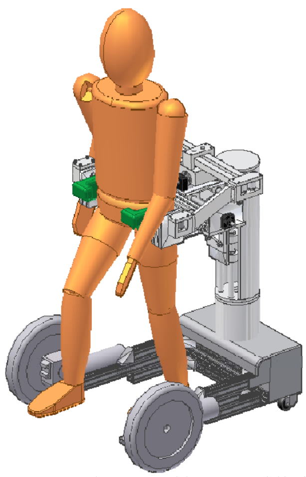

Fig. 12. BWS system integrated into a mobile base system

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