A Performance Evaluation of Overground Gait Training with a Mobile Body Weight Support System using Wearable Sensors

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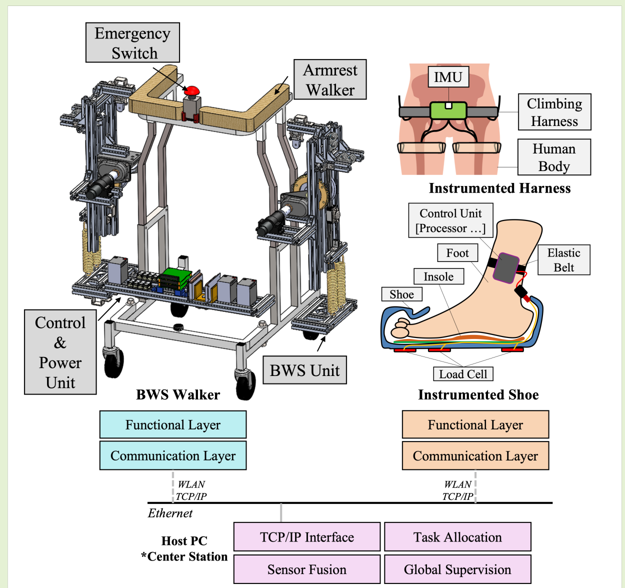
***Abstract*— Overground gait training under body weight support (BWS) for patients who suffer from neurological injuries has been proven practical in recovering from walking ability. Conventionally, skilled therapists or additional robots are required to assist the patient's body weight and pelvis movement, making the rehabilitation process physically and economically burdensome. We investigate if a BWS walker using only two actuators can support the user's body weight and simultaneously protect/assist the transverse pelvis rotation, improving natural gait with minimal motion compensation. In this paper, a BWS strategy called transverse pelvis rotation support (TPRS) is proposed to enable the BWS system to generate cable tension in the forward direction, as a purpose to support transverse pelvis rotation in addition to our previously proposed static or variable BWS. Wearable sensory devices, including instrumented shoes and harness, were developed to provide real-time ground reaction force and pelvis rotation signals simultaneously. Ten non-disabled participants were unloaded with 0% $\sim$ 15% BWS under four different controls. Vertical ground reaction force, transverse pelvis kinematics, and user experience were compared using proposed controls. One-Way repeated measures ANOVA analysis assessed if control strategies generally affect the performance. All proposed controls enable the walker to support part of the user's body weight. SBWS-TPRS and VBWS-TPRS control enable users to achieve a significantly improved pelvic motion and prolonged single support phase than pure static BWS or variable BWS, although users perceive a higher workload under them. The proposed BWS controls show the potential to become a complementary method in gait rehabilitation.**

***Index Terms*— Partial body weight support, wearable sensors, physical assistive device, human-robot interaction, robotic rehabilitation**



I. Introduction

TRAUMATIC brain injuries like stroke or cerebral palsy are among the most common and severe diseases worldwide that are often associated with disorders of posture and movement due to lingering damage or lesions in the brain [1]. As a result, people who suffer from traumatic brain injuries can have locomotor deficiencies, including lowered muscular strength and endurance, reduced pelvis rotation, and irregular foot landing motion, resulting in spatiotemporally asymmetrical and unstable gait patterns [2]. Furthermore, this can increase the risk of falling and developing other secondary musculoskeletal injuries [3].

The recovery of gait mobility based on locomotion rehabilitation has been considered an important goal for patients with gait impairments in the last several decades [4]. A successful gait recovery can be strongly associated with a higher degree of motivation and engagement [5], improve the individual's ability to perform activities of daily living and his/her capacity to participate in family or social life [6] [7].

Body Weight Support (BWS) gait rehabilitation on a tread-

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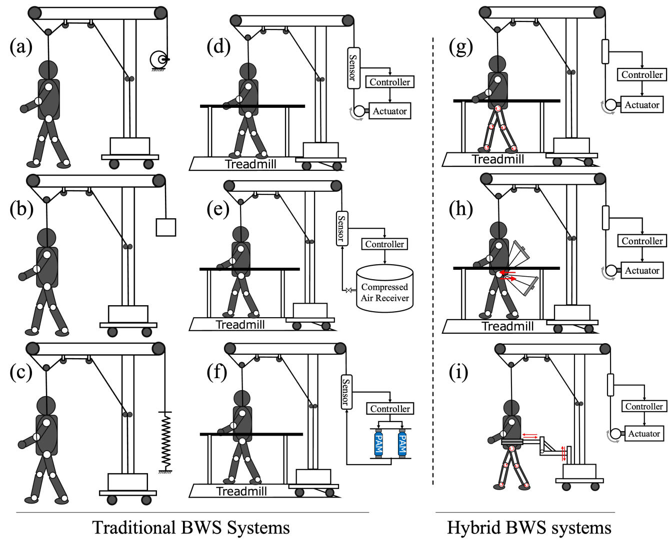


Fig. 1. State-of-Art of BWS systems: (a). static BWS system using a winch. (b). static BWS system using a counterweight. (c). passive BWS system using a spring. (d). active BWS system with an electric actuator. (e). active BWS system with a pneumatic actuator. (f). active BWS system with a hydraulic actuator. (g). Hybrid BWS system integrated with an exoskeleton. (h). Hybrid BWS system integrated with pelvic assist manipulator. (I). Hybrid BWS system integrated with robotic arms and orthoses.

mill has been widely studied and gained much attention for its ability to enable patients to start gait training in the very early stage in a safe manner while helping to repeat stepping sequences with high intensity [8] [9]. Additionally, studies have found that an overground BWS system can lead to equivalent [10] or more improvement in walking ability than that on the treadmill [11]. Overground walking while receiving BWS would represent a better and more generalizable practice for patients seeking to improve their walking ability in different environments, including their homes. Performing the training in the patient's comfort zone makes them more relaxed during the training [12].

In recent years, various kinds of BWS systems have been developed. The general types of the BWS systems consist of a harness system worn by the user, ropes that go through pulleys to connect the user's trunk with the actuators, carrying part percentage of user's body weight. Different types of actuators have been developed for the last two decades to be implemented with the BWS system (Fig.1 (a)∼(f) summarized the traditional BWS systems with different types of actuators). The primary purpose is to enable the system to generate the desired cable tension so that a patient can walk with a fixed amount of supported body weight [13]. Static and passive systems (Fig.1.a,b,c) using elastic springs (Fig.1.c) or counterweights (Fig.1.b) can be easily implemented and are convenient for operation [14] [15]. Active BWS systems with electric [16](Fig.1.d), pneumatic [17](Fig.1.e), hydraulic [18], or other types of actuators [19] (Fig.1.f) were later introduced, and they were useful for generating stable levels of cable tension. However, there is growing evidence that constantly unloading part of the body weight may limit several joint motions including lateral and transverse pelvic movement,

TABLE I
TABLE OF NOMENCLATURE

Symbol	Definition
NW	Natural Walking
TPR	Transverse Pelvis Rotation
SBWS	Static Body Weight Support
VBWS	Variable Body Weight Support
S(BWS)-TPRS	SBWS with TPR Support
V(BWS)-TPRS	VBWS with TPR Support
MPA-TPR	Average of largest amplitude in TPR
MA-TPR	Average of all amplitudes in TPR
PTM	Pelvis Total Movement
PL	Phase Lag of TPR relative to right foot
WL	NASA Workload
UE	User Experience

which are two of the six determinants of the natural gait [20]. Moreover, the duration of the user's stance and double-limb support may be decreased under relatively higher levels of BWS, which suggests that BWS can stimulate shorter periods of foot contact. This may be undesirable because the foot contact phase plays a key role in providing balance support for stable walking and muscular rehabilitation [21]. Therefore, strategies for simulating those movements under partial BWS gait rehabilitation training should be considered.

Three methods have been proposed to support or preserve the natural gait motion under partial BWS based on the state-of-the-art research: therapist-based, mechanism-based, and control-based approach. The therapist-based approach, which has been the most widely used since the last two decades, requires several skilled therapists to give manual assistance for patient's foot placement, knee flexion and extension, and pelvis alignment during training procedures, as a purpose to help to generate a normal physiological gait pattern [22] [23]. Additional therapists may also be needed to observe the rehabilitation performance and progress objectively. Therefore, the BWS training procedure can be physically exhaustive for physical therapists [24]. Recent studies have proposed a novel recovery evaluation system by integrating various kinds of wearable sensory devices and hybrid intelligent computation to monitor the rehabilitation progress autonomously and objectively [25]. This shows great potential in minimal human intervention, especially for anterior cruciate ligament reconstructed subjects [26] [27]. Nevertheless, therapist-based physical assistance of the lower extremity is still required for subjects having lower limb impairments due to stroke or cerebral palsy.

The mechanism-based approach is to develop hybrid BWS systems controlled with other robotic devices like exoskeletons (Fig.1.g) and powered orthoses (Fig.1.h) to provide support directly to the patients. For example, Li et al. developed a robotic gait training system called PRPGT, which was integrated with a pneumatic BWS system, a pneumatic postural support system, and a pneumatic gait orthosis system for patients who suffer from weakened lower limbs [28]. T. P. Luu et al. developed a robot called NaTUre-gaits, a BWS gait training platform containing 14 DoFs (Fig.1.i). Additional robotic arms and orthoses are used for active assistance to hip, knee, and ankle joints in the sagittal plane [29]. Lokomat, one of the most widely used devices for robot-assisted gait therapy (RAGT), is combined with a treadmill BWS system and a bilaterally driven gait exoskeleton [30]. Literature suggests that such combined configuration may be effective in improving walking ability and balance in patients with chronic stroke [31]. Nevertheless, the widespread adoption of such robotic devices is still facing obstacles due to the high economic costs and poor portability.

Lastly, the control-based approach is to equip the BWS system to provide synchronously modulated BWS support, which requires the system to adjust the unloading force based on the user's walking kinematics or muscle activities, such as gait events, the center of mass acceleration, lower limb muscle activities, et al.. Depending on the walking environment, the control-based approach can be easily adapted into treadmill-based and overground BWS training. [32] For example, J.R. Franz proposed a gait synchronized force modulation during the stance period of one limb by using a BWS treadmill system. They found that modulated support caused improvement in knee flexion, hip flexion, and duration of single-limb support [33]. On the other hand, T.V. Thuc developed a BWS treadmill system using pneumatic muscles and controlling the force that tracks the center of pressure from the left to right and vice versa [19]. Through clinical experiments with the disabled patient, the researchers found that the Body Weight Support system with the unloading force modulation showed more advanced and better behavior than the traditional Counter Weight system [34]. Munawar developed a series elastic active BWS system called GRAVITY-ASSIST. By utilizing emulated inertia compensation scheme, the robot can actively compensate for the inertial forces detected by the human body, improving lateral stability during walking [35]. Hidler et al. created an overground BWS training system called ZeroG, which attempts to provide BWS by dynamically controlling the cable tension through a custom-series elastic actuator. The authors suggested that more natural ground reaction forces and gait characteristics can be achieved using this method [35]. The benefit of such a strategy is that no additional robots or structures are needed, only required to detect the user's walking status like gait cycles or system dynamics, and apply to appreciate control of unloading force. However, these systems are usually very cumbersome, limiting their usage to treadmills or overhead tracks only.

In our previous studies, we developed a compact and mobile BWS walker system with 2 DoFs [36] which is integrated with a pair of instrumented shoes that can measure users' gait events and communicate with the system over Wi-Fi. In addition, we proposed two BWS control strategies, static and variable BWS control. During static BWS walking, the user walks while a part of their weight is constantly supported by the walker. Under this strategy, we observed that users exhibited a gait that was not natural (i.e., small pelvic motion, reduced stance phase duration). However, under variable BWS walking, the controlled unloading force from the walker was synchronized with the user's gait, aiming to preserve their natural gait. Through experiments with non-disabled subjects, we found that subjects showed improved pelvis rotation and total pelvis motion, together with an improved feeling of comfort and naturalness under variable BWS [37].

Since pelvis motion plays a critical role in serving an optimal center of mass movement, producing smooth and energetically efficient overground gait locomotion [20], monitoring and supporting the user's pelvis movement simultaneously under partial BWS becomes essential [38]. To look into pelvis movement in more detail, in this research, we utilized the center of mass (CoM) model to represent human walking, and investigate the pelvis motion within the transverse plane. A 3D human link model is utilized, and we hypothesize that the CoM (pelvis) [39] movement could represent lower extremity activities without considering the cyclic contraction of numerous inner muscles [40]. The knowledge of foot the CoM movement together with foot placement could help identify rehabilitation practices for patients with balance disorders and has been widely considered by skilled therapists [41] [42]. Using this consideration, we need to make an extensive hardware upgrade to our previous system, enabling it to monitor pelvis movement and foot placement in real-time.

Many sensory devices have been developed and integrated with robotic systems so far to provide the precious measurement of pelvis movement in different ways [43]. For example, Electromyography (EMG), including surface, needle, or wire types, is considered one acceptable tool for real-time evaluation of the bioelectrical activity of the pelvic muscle activity by many researchers. Some researchers also combine the EMG evaluation with brain activity using the electroencephalograph (EEG) or the magnetoencephalogram (MEG) as a part of the gait evaluation [44]. Optical motion capture systems or force plates-based kinesiology analysis is another approach for evaluating pelvis movement [45], which is frequently combined with EMG or EEG in the clinical neurophysiological investigation of pelvis movement.

Another option for studying the pelvis movement would be using an inertial measurement unit (IMU), which contains three-dimensional accelerometers, gyroscopes, a magnetometer (optional), and GPS (optional) within a small, inexpensive chip. Considering some unsupervised at-home rehabilitation training does not require a fully comprehensive kinematic analysis of motion [46], using a single IMU for kinematics motion analysis provides an increasingly convenient, portable, and affordable way of monitoring human motion as a complementary monitoring method in addition to a clinical investigation [47].

In this research, we focus on developing an inexpensive, compact, and portable wearable motion monitor device that can provide a 3D measurement of pelvis motion with high sensibility and accuracy. Thus, our first upgrade for the BWS system is the implementation of a single IMU-embedded harness. The absolute yaw orientations are utilized as the kinematic information of pelvis rotation in the transverse plane.

Furthermore, we proposed two complementary control methods (SBWS-TPRS and VBWS-TPRS) for the developed walker system to support pelvis rotation on the transverse plane (defined as Transverse Pelvis Rotation, or TPR in this article) by utilizing the user's upper limbs strength. Nevertheless, the feasibility of BWS systems to support transverse pelvis rotation without using additional robots has to our best knowledge, not yet been investigated. It is therefore not clear if the proposed robot and controls can simulate an improved gait while always keeping the user in a safe condition with a better feeling of human-robot interaction. Validation experiments with non-disabled populations can effectively help us determine the system's usability and portability. Thus, before such control strategies can be clinically implemented in rehabilitation, we believe it prudent to assess the impact of such a system on the gait of healthy subjects.

In this paper, we discuss the pros and cons of the four proposed BWS control methods and summarize the possible use cases for each of them in this article. This investigation under our lab experiment shows the potential of the developed BWS walker's capability of supporting both BW and pelvis motion while monitoring the user's kinematics features simultaneously without using additional actuators or sensory devices. The experimental result also provides a reference for specialist therapists concerning further clinical investigation under health care facilities.

This article is organized as follows: In Section II, we introduce the improved hardware for the mobile BWS walker system. Together we introduce a pair of instrumented shoes and an IMU-equipped harness, which can measure the ground reaction force signals and transverse pelvis rotation signals in real time. We also describe the system control strategies for SBWS, VBWS, SBWS-TPRS, and VBWS-TPRS in Section III. In Section IV, we present an overground walking experiment with 10 non-disabled subjects. Finally, we discuss the experiment result and conclude our article in Sections V and VI, respectively.

II. BWS System archietcture

The mobile BWS walker system consisted of three hardware components: 1) Mobile BWS walker capable of providing cable tensions in various directions. 2) Instrumented shoe system with a high sampling rate. 3) Instrumented harness equipped with IMU. All these devices could be used simultaneously by synchronizing the time sequences within a real-time operating system.

A. Mobile BWS walker with instrumented shoe systems

The mobile BWS walker system was developed by attaching a pair of BWS units to both sides of a commercial armrest walker using aluminum profiles. The BWS units were developed using a Variable Stiffness Mechanism (VSM) [36] [37]. Each of them has one independently controlled actuator to adjust the elongation of the springs, which in turn controls the equilibrium position (virtual stiffness) of the unit. A wire was used in each unit to connect the harness with the spring system, while on the other side, it is connected to the actuator to provide the desired unloading force to the user. We designed an additional roller mechanism to guide the wire motion in this iteration. Despite the relative walking position between the user and the walker, it could provide tension in various directions.

The instrumented shoe system was developed to measure the vertical ground reaction force (v-GRF) with respect to time.

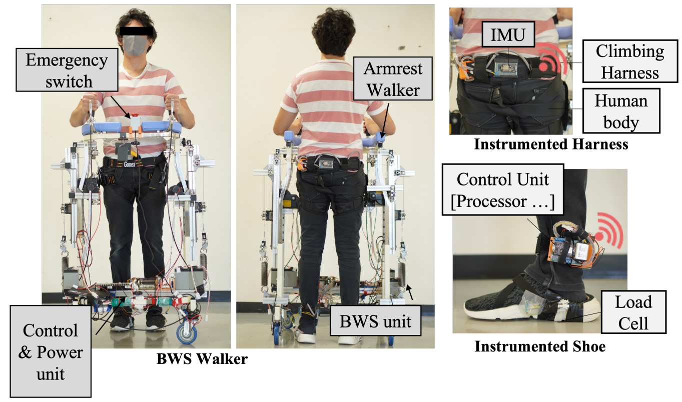


Fig. 2. Proposed BWS system structure

The developed shoe system achieved an average accuracy of around $98.83\%(\pm 0.13\%)$ of the true v-GRF value. This system can also help us detect the user's four-stage gait events by adapting a finite state machine into the control scheme so that the walker can provide variable unloading forces synchronized to the user's gait [37]. In our previous setup, we set a 10Hz sample rate, as we believe it is appropriate to detect the gait for a low walking speed (about 0.34m/s). This time, we increased the sample rate to 80Hz, so that more detailed v-GRF signals could be measured, enabling the walker to have a faster response accordingly.

B. IMU-equipped harness

We developed an IMU-equipped harness to measure the transverse pelvis rotation during overground gait training, which is a major criterion for evaluating walking performance without using external sensors like a motion capture system. Thus, the instrumented harness can be used as a complementary tool in addition to traditional optical motion capture system using infrared cameras and reflective markers [48]. This system can also be utilized for at-home gait training when professional therapists and equipment are absent [49]. The appearance of the harness system can be found in Fig.1. The harness was built by integrating an IMU (BNO055, 9-axis absolute orientation sensor), which can measure the azimuth angle for the roll, pitch, and yaw and communicate with the BWS walker over Wi-Fi under 100 Hz sample rate. The IMU sensor was positioned and fixed in the back of the harness, which was later attached to the human hip joint. Thus, it is possible to measure the orientation (yaw angle) changes, described in this article as the lateral/transverse pelvis rotation (Fig. 2). Using the dynamic motion processer from the IMU chip, the accuracy for the transverse pelvis rotation can achieve an accuracy around $1.28°(\pm 0.71°)$ [50]. The sample rate is set to 100Hz, and the approximate whole weight of the subsystem is only 0.8 Kg.

C. Lower extremity analysis using wearable devices

The abstract figure shows the whole system architecture and data flow. The function layers of the BWS walker and wearable devices allow part of the user's body weight to

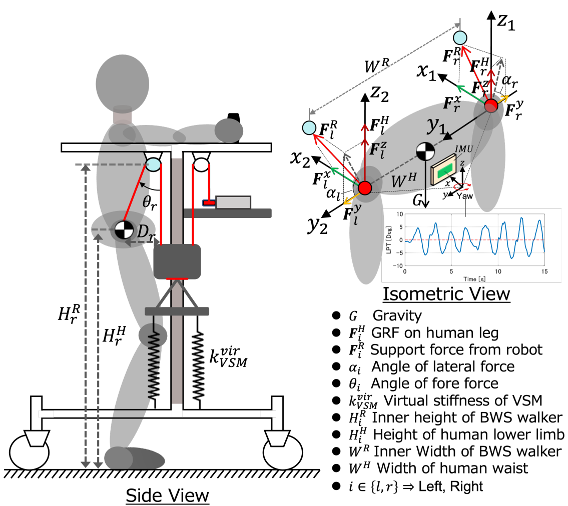


Fig. 3. Proposed BWS system structure

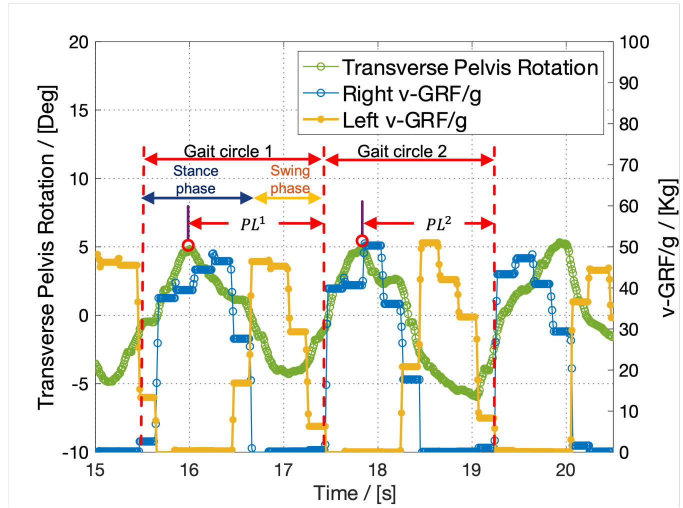


Fig. 4. Lower extremity analysis using developed wearable devices: 1. Stance phase occupied time compared to a single full gait cycle. 2. Phase Lag(PL): Timing difference between rising edges of the right foot and local maxima of the transverse pelvis rotation (when the pelvis hits the rightmost position in each gait cycle)

be alleviated while collecting the user's foot dynamics and pelvis kinematics data in real-time. A communication layer utilizes the TCP/IP protocol, to allow the data from each subsystem collected correctly and could be used within the host PC. Finally, Once the sensory information(encoder, v-GRF, and TPR) has been received within each control loop, the host PC determines the user's left/right leg's gait phases and generalizes the control signals for each motor, respectively. Note that the whole system is under the same local network with a shared IP address so that stable data communication can be achieved.

Note that the obtained v-GRF and TPR data are shared with the PC104 host PC, running the real-time QNX OS under a 1kHz control loop. To synchronize the time sequences for both wearable devices and the main control loop, we utilized the POSIX thread model service from QNX Neutrino microkernel and utilized four threads (one for the harness, two for shoes, and the last one for the main control loop) under the same process. These threads are shared with the same memory space and clock information. Therefore, timestamps for V-GRF, TPR, and the main control loop can be shared with the same value and later collected into the same log file (Fig.4 shows the collected v-GRF and TPR data from host PC).

Since the small changes in foot dynamics can significantly influence pelvis rotation, especially when the robot alleviates the user's partial body weight. It is important to look into the lower extremity movement in more detail by engaging the transverse pelvis rotation with foot dynamics and find out how the walking pattern changed [51]. In this research, we focus on the lower extremity analysis by computing the left/right single/double support period based on only the v-GRF signal and phase lag of TPR relative to actual footsteps. Fig.3 shows the definition of phase lag. A desired phase lag should ideally stay constant, which implies that the relationship between pelvis and foot dynamics is the same, resulting in a stable step pattern. [52]

From the result of the foot dynamics study and PL, we could have a stability and consistency analysis of walking by comparing the standard deviation (SD) of PL (stable walking should have a low SD in its PL) [52].

III. SYSTEM CONTROL

A. Mathematical model for BWS walking

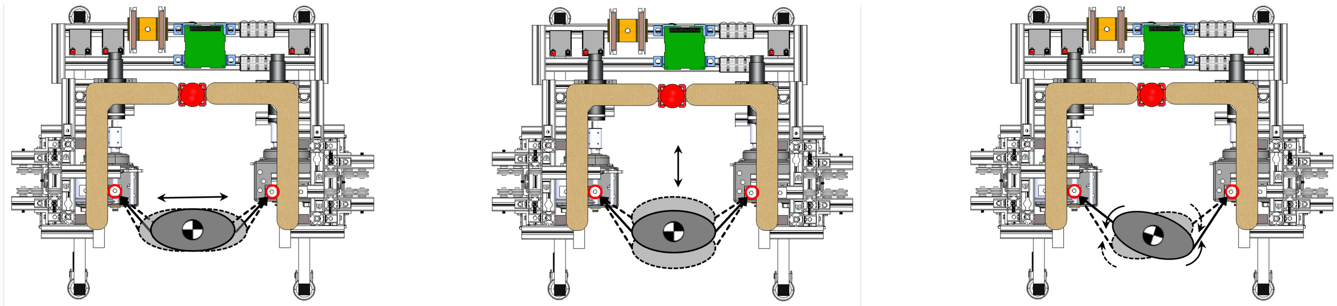
The mathematical model for interaction between the human and the walker is extended from a 2-dimensional planar system into a 3-dimensional space system. In Fig. 2, we assumed that the user walks in the horizontal center of the walker, while the center of mass is not necessary vertically below the support point. Thus, the controlled cable tension from the robot has components in $\vec{x}$, $\vec{y}$ and $\vec{z}$ directions. In this case, the motion equation of the system on the vertical direction ($\vec{z}$) could be described as below:

$$[\boldsymbol{F}_r^z + \boldsymbol{F}_l^z\ ] + [\boldsymbol{F}_r^H + \boldsymbol{F}_l^H] - \boldsymbol{G} = m\boldsymbol{a} \tag{1}$$

$$[\boldsymbol{F}_r^z + \boldsymbol{F}_l^z\ ] = \beta\boldsymbol{G} \quad \beta \in (0,1) \tag{2}$$

Here $\boldsymbol{F}_r^z$ and $\boldsymbol{F}_l^z$ stand for the controlled cable tension component in the vertical direction in the left and right side respectively, which are used to support part of the user's body weight. $\boldsymbol{F}_l^H$ and $\boldsymbol{F}_r^H$ are the ground reaction forces applied from the human left and right legs, respectively. m and $\boldsymbol{G}$ stand for the user's body weight and gravity. The body weight ratio, which means the desired percentage of BWS from the walker, is defined as β. We assume that the user walks steadily and slowly, so there is no significant motion of their center of mass in the vertical direction. In this case, we assume the $\boldsymbol{a} = 0\ \vec{z}$.

The controlled cable tension component in the forward direction ($\vec{x}$) from either left or right side helps to promote the user's gait motion by pulling each side of the user's anterior superior iliac spine forward to assist the rotation, which is defined in this article as the $\boldsymbol{F}^x$. Using the desired $\boldsymbol{F}^x$ and



(a) (b) (c)

Fig. 5. Three major pelvic movements when using the proposed robot: (a) lateral movement, (b) forward movement, (c) transverse pelvis rotation

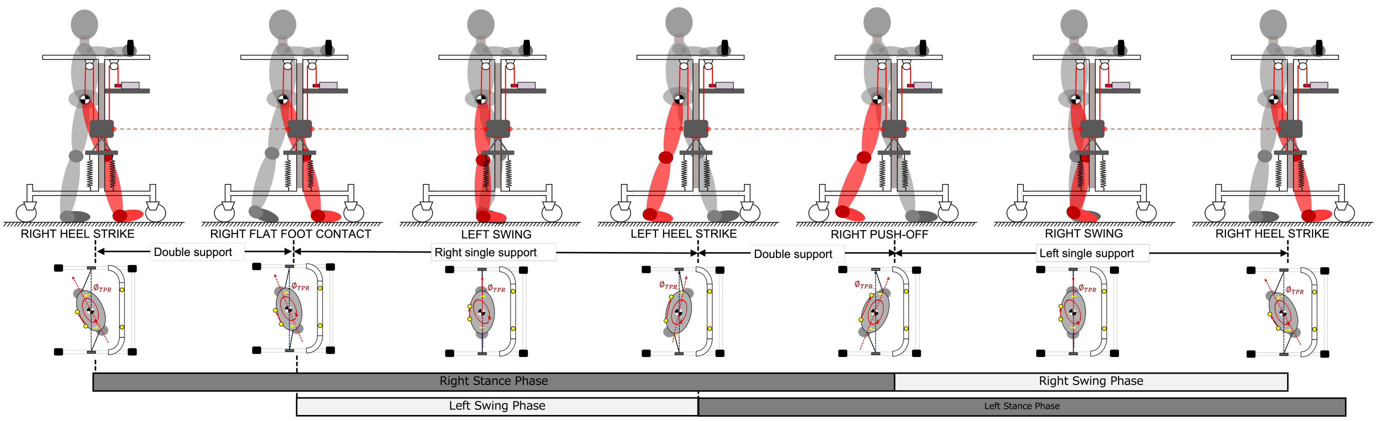


Fig. 6. SBWS Walking with both side and top view: Controlled stiffness kept constant in $\vec{z}$ direction despite gait events changes. ϕ_{TPR} is transverse pelvis rotation at the center of mass.

side angle θ, the tension component in the forward direction ($\vec{x}$) could be desired as:

$$\boldsymbol{F}^x = |\boldsymbol{F}^z| \tan(\theta)\vec{x} \tag{3}$$

$$\theta = \tan^{-1}\left[\frac{D}{H^R - H^H}\right] \tag{4}$$

Here the relative longitudinal distance D is considered as a predefined constant parameter, which is determined by using the initial side angle θ and the dimensions of both user and walker.

The last cable tension component remains in the horizontal direction, which keeps the user walking in the horizontal center of the robot. Next, we define α, which is the cable angle viewed from the coronal plane. This can be calculated by detecting the dimensions of both user's body and the walker. Note that each α is assumed to be constant during walking motion and different users with different dimensions (leg length, waist width, etc.) could have different α. Using the desired F^z, the tension component in the horizontal direction ($\vec{x}$) could be calculated as:

$$\boldsymbol{F}^y = \frac{|\boldsymbol{F}^z|}{\tan(\alpha)}\vec{y} \tag{5}$$

Finally, the desired cable tension $\boldsymbol{F}^R_{l,r}$ including force components in 3-dimensional space should be:

$$\boldsymbol{F}^R = |\boldsymbol{F}^z| \left[\tan(\theta)\vec{x} + \frac{1}{\tan(\alpha)}\vec{y} + \vec{z}\right] \tag{6}$$

Note in the above equation, when side angle $\theta_{\{l,r\}} = 0$, (6) turns into:

$$\boldsymbol{F}^R = |\boldsymbol{F}^z| \left[\frac{1}{\tan(\alpha)}\vec{y} + \vec{z}\right] \tag{7}$$

which is coincident to the formulation derived in [3] (Eq. (3)).

We should note that cable tensions $\boldsymbol{F}^R$ are connected near both sides of the pelvis, thus pelvis can move freely both in the lateral and forward direction (Fig.3(a)(b)). Transverse pelvis rotation support can be produced using the controlled cable tension component in the forward direction (Fig.3(c)).

B. Static BWS control strategy

The static BWS control strategy was realized by using an equilibrium-controlled stiffness, which aims at keeping the controlled stiffness of the robot constant, so that human could be supported with a fixed continuous BWS while walking (Fig.6). The cable tension controlled in the $\vec{z}$ and $\vec{y}$ direction help to support part of the user's body weight and keep the human walking in the center to the lateral direction. The side angle θ is assumed to be 0, which implies $\boldsymbol{F}^x = 0\ \vec{x}$.

$$\boldsymbol{F}^{BWS} = \boldsymbol{F}^z, \quad \boldsymbol{F}^{VSM} = \boldsymbol{F}^R \tag{8}$$

$$\boldsymbol{F}^{VSM} = \left|\boldsymbol{F}^{BWS}\right| \left[\frac{1}{\tan(\alpha)}\vec{y} + \vec{z}\right] \tag{9}$$

Here the desired cable tension F^R is generated using the output force F^{VSM} from the variable stiffness mechanism

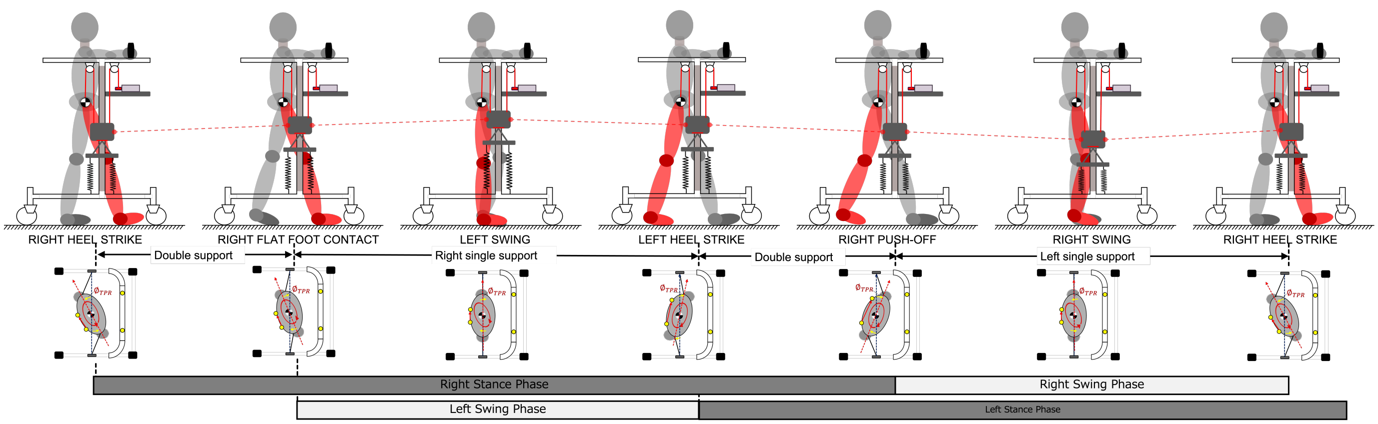


Fig. 7. VBWS walking with both side and top view: Controlled stiffness in $\vec{z}$ direction synchronized with real-time gait event. ϕ_{LPT} is transverse pelvis rotation at the center of mass

inside the BWS unit. The relationship between the desired force, stiffness and final equilibrium of the spring system was fully described in [30]. A simple proportional-plus-derivative (PD) controller was used for this control scenario.

C. Variable BWS control strategy

The variable BWS control strategy synchronizes the controlled stiffness of the walker with the user's walking motion (Fig.5). When the user's gait event changes from swing to heel strike phase, we increase the stiffness by a certain amount to maintain stability and reduce the strength necessary to keep an upright posture. When the user's gait event changes from flat contact to the push-off phase, which reduces the constraints on the leg, thus enabling a more natural gait. In this case, the desired cable tension $\boldsymbol{F}^{VSM}$ can be described below.

From double support period→ right single support (control parameter changes in the instance when gait event changes from right swing phase to right initial contact phase)

$$\boldsymbol{F}^{VSM} = \left|\boldsymbol{F}^{BWS}\right|(1+\frac{x\%}{2})\left[\frac{1}{\tan(\alpha)}\vec{y}+\vec{z}\right] \quad (10)$$

From double support period → right swing phase (control parameter changes in the instance when gait event changes from left swing phase to left initial contact phase)

$$\boldsymbol{F}^{VSM} = \left|\boldsymbol{F}^{BWS}\right|(1-\frac{x\%}{2})\left[\frac{1}{\tan(\alpha)}\vec{y}+\vec{z}\right] \quad (11)$$

Here the x% represents the net difference between the stance phase and swing phase. A bigger x% represents fewer constraints on the leg during the swing phase and less strength required to keep an upright posture during the stance phase. We should note that by rearranging (10)-(11), the total BWS level is the same as the SBWS walking scenario ((9)). The implementation of the VBWS control strategy, including the gait event classification and synchronization, is the same as we proposed previously, which can be found in [37].

D. Static BWS with TPRS control strategy

Both proposed strategies provide partial BWS, but we need to consider the horizontal pelvis rotation to achieve a more natural gait, which is difficult under a high level of BWS because the pelvis is constrained by the forces exerted by the BWS unit. To solve this issue, we propose the TPRS control strategy. Using this method, θ is no longer assumed to be 0, and the force component in the forward direction is considered.

$$\boldsymbol{F}^{VSM} = \left|\boldsymbol{F}^{BWS}\right|\left[\tan(\theta)\vec{x}+\frac{1}{\tan(\alpha)}\vec{y}+\vec{z}\right] \quad (12)$$

Note that $\left|\boldsymbol{F}^{BWS}\right|\tan(\theta)\vec{x}$ is assumed to always exist while the user is walking. We define $\boldsymbol{F}^{TPRS}$ as the force which supports the transverse pelvis rotation. Considering the inner force interaction between the user and the walker, $\boldsymbol{F}^{TPRS}$ could be described as:

From double support period→ right single support (control parameter changes in the instance when gait event changes from right swing phase to right initial contact phase)

$$\boldsymbol{F}^{TPRS} = \left|\boldsymbol{F}^{BWS}\right|\tan(\theta)\vec{x}+(\boldsymbol{F}^{U}+\boldsymbol{F}^{\mu}) \quad (13)$$

$$\left|\boldsymbol{F}^{BWS}\right|\tan(\theta)\vec{x} = -(\boldsymbol{F}^{U}+\boldsymbol{F}^{\mu}) \quad (14)$$

From double support period → right swing phase (control parameter changes in the instance when gait event changes from left swing phase to left initial contact phase)

$$\boldsymbol{F}^{TPRS} = \left|\boldsymbol{F}^{BWS}\right|\tan(\theta)\vec{x}+\left(\boldsymbol{F}^{U}\right), \quad \boldsymbol{F}^{\mu}=0 \quad (15)$$

Here $\boldsymbol{F}^{U}$ and $\boldsymbol{F}^{\mu}$ represent the upper limb strength and static friction on the ground, respectively. Their direction is opposite from $\vec{x}$. During the single support phase, both $\boldsymbol{F}^{U}$ and $\boldsymbol{F}^{\mu}$ act in the opposite direction of the cable tension, so the user is not pulled forward by the tension of the cable. However, static friction between the leg and ground on the swinging leg during the swing phase becomes 0. The remaining unmatched force then pulls the pelvis forward, which supports the pelvis rotation.

E. Variable BWS with TPRS control strategy

Under SBWS-TPRS walking, the forward support tension is produced as soon as the foot is taking off from the ground.

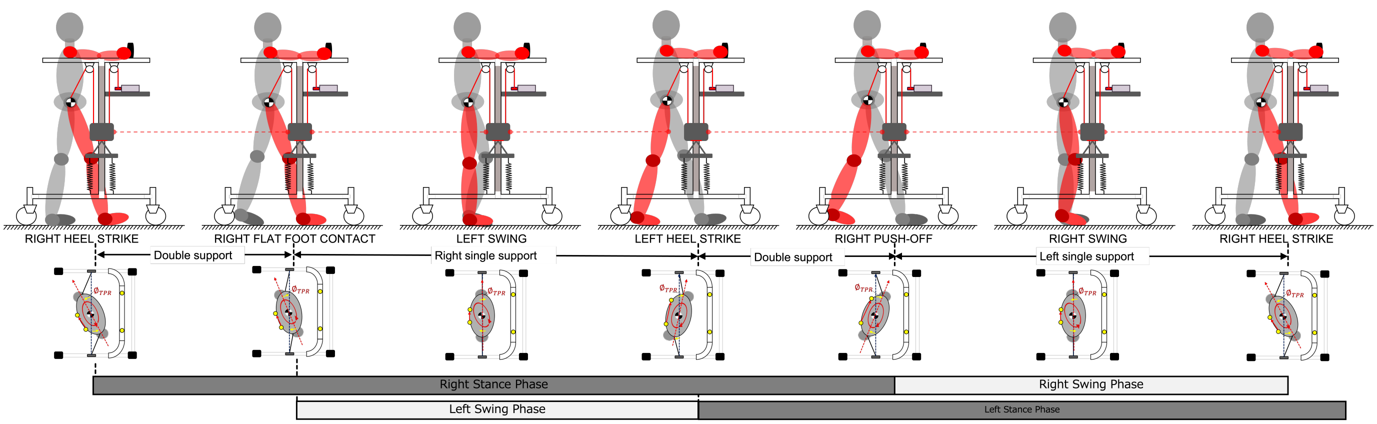


Fig. 8. SBWS-TPRS Walking with both side and top view: Controlled stiffness kept constant in all 3 directions despite gait events changes. ϕ_{TPR} is transverse pelvis rotation at the center of mass. $\theta^{i}_{l,r}$ is the initial side angle. $\theta^{t1}_{l,r}$, $\theta^{t2}_{l,r}$,$\theta^{f}_{l,r}$ stands for the change of side angle during and at the end of half gait circle, respectively.

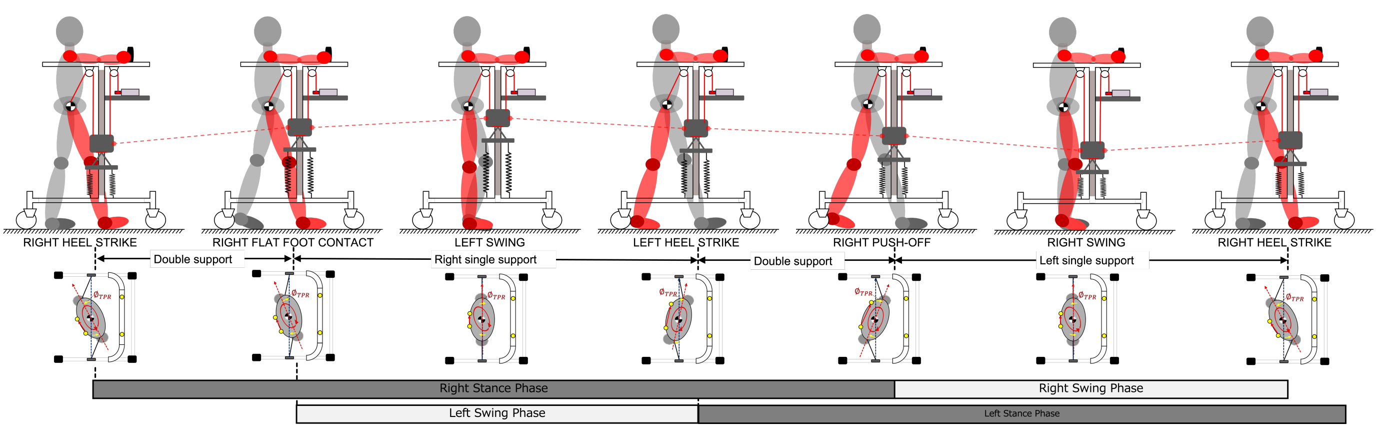


Fig. 9. VBWS-TPRS Walking with both side and top view: Controlled stiffness in all 3 directions synchronized with real-time gait event. ϕ_{TPR} is transverse pelvis rotation at the center of mass. $\theta^{i}_{l,r}$ is the initial side angle. $\theta^{t1}_{l,r}$, $\theta^{t2}_{l,r}$,$\theta^{f}_{l,r}$ stands for the side angle changes during and at the end of half gait circle, respectively.

Since $\boldsymbol{F}^{TPRS}$ increases suddenly when entering the swing phase, the SBWS-TPRS may not be the best choice as the sudden change might affect the balance of the user.

The variable BWS with TPRS may solve the above problem. This method allows the robot to decrease the cable tension in all 3 directions when the detected user's gait event changes from flat contact to push-off phase and increase the tension in all directions when the gait event changes from swing to heel-contact phase. The desired cable tension is similar to the VBWS control strategy, with only an additional term in the $\vec{x}$ direction.

From double support period→ right single support (control parameter changes in the instance when gait event changes from right swing phase to right initial contact phase)

$$\boldsymbol{F}^{VSM} = \left|\boldsymbol{F}^{BWS}\right|(1+\frac{x\%}{2})\left[\tan(\theta)\vec{x}+\frac{1}{\tan(\alpha)}\vec{y}+\vec{z}\right] \tag{16}$$

$$\boldsymbol{F}^{TPRS} = \left|\boldsymbol{F}^{BWS}\right|(1+\frac{x\%}{2})\tan(\theta)\vec{x}+(\boldsymbol{F}^{U}+\boldsymbol{F}^{\mu}) \tag{17}$$

From double support period → right swing phase (control parameter changes in the instance when gait event changes from left swing phase to left initial contact phase)

$$\boldsymbol{F}^{VSM} = \left|\boldsymbol{F}^{BWS}\right|(1-\frac{x\%}{2})\left[\tan(\theta)\vec{x}+\frac{1}{\tan(\alpha)}\vec{y}+\vec{z}\right] \tag{18}$$

$$\boldsymbol{F}^{TPRS} = \left|\boldsymbol{F}^{BWS}\right|(1-\frac{x\%}{2})\tan(\theta)\vec{x}+(\boldsymbol{F}^{U}) \tag{19}$$

We can find that the difference of $\boldsymbol{F}^{LPTS}$ between the SBWS-TPRS and VBWS-TPRS is that on the latter, the BWS and TPRS decrease simultaneously to reduce the constraint on the swinging leg while providing some support on the pelvis rotation.

IV. EXPERIMENT

We conducted an overground walking experiment to evaluate the performance of BWS walking based on the different control methods. This study mainly focused on the kinematics of TPR, stance period duration, and user experience between different BWS levels and controls as compared with 0% BWS in a group of the non-disabled population. We utilized the instrumented shoes to validate the BWS capability by measuring the changes in body weight ratio. We analyzed

TABLE II
EXPERIMENT PROTOCOL

BWS Ratio × 100		SBWS	VBWS	SBWS-TPRS	VBWS-TPRS
0%	-	-	-	-	-
5%	Stance Phase	-2.5%	-3.5%	-2.5%	-3.5%
	Net Difference	0.0%	±2.0%	0.0%	±3.0%
	Initial Side Angle	0°	0°	30°	30°
	Swing Phase	-2.5%	-1.5%	-2.5%	-1.5%
10%	Stance Phase	-5.0%	-6.0%	-5.0%	-6.0%
	Net Difference	0.0%	±2.0%	0.0%	±3.0%
	Initial Side Angle	0°	0°	30°	30°
	Swing Phase	-5.0%	-4.0%	-5.0%	-4.0%
15%	Stance Phase	-7.5%	-8.5%	-7.5%	-8.5%
	Net Difference	0%	±2.0%	0.0%	±3.0%
	Initial Side Angle	0°	0°	30°	30°
	Swing Phase	-7.5%	-6.5%	-7.5%	-6.5%

and compared the experimental results under natural walking (NW), static BWS walking (SBWS), variable BWS walking (VBWS), SBWS with transverse pelvis rotation support (SBWS-TPRS), and VBWS with transverse pelvis rotation support (VBWS-TPRS). Before the experiment, the research purpose, method, and data handling were fully explained to each participant, and we obtained their informed consent. Each subject conducted the experiment after we received their approval. 10 non-disabled subjects participated in this experiment. (Gender: male; mean age ± standard deviation, 25.5 ± 3.0; height, 169.5 ± 5.5 cm; body weight, 60.8 ± 5.8 kg). Demo video describing the actual experiment can be found in supplementary materials.

Table II shows the experimental protocol of the overground walking experiment under each control scenario. Natural walking means that the user walks with the robot with 0% BWS. Previous studies indicated that high BWS levels could easily lead to smaller maximum trunk and pelvis movement amplitudes than natural walking, and literature generally advises the use of a BWS level lower than 30% in healthy objects [53]. Therefore, in this experiment, under each BWS walking, we set a different BWS ratio β from 0.05 to 0.15 in steps of 0.05, in other words, unloading 5% to 15% of the user's body weight.

During SBWS walking, the user was supported under constant and continuous cable tension. Each BWS unit provides half of the desired weight support. For example, when we support 15% of the total body weight, each unit has a support ratio of 7.5%, for a total of 15%.

During VBWS walking, the weight support ratio is the same as we set for SBWS. However, the cable tension controlled by each BWS unit changes synchronously based on the user's gait events. The unit supporting the leg in contact with the floor (i.e., stance phase) will provide an increased support ratio (3.5%, 6.0%, or 8.5%), while the unit supporting the leg currently in the swing phase will provide a decreased ratio (1.5%, 4.0%, or 6.5%) so that the sum of both support ratios is equal to the SBWS walking scenario (5.0%, 10.0%, or 15.0%). The net difference between the stance phase and to swing phase in this experiment was set to the median according to our last study, which is 2%.

In both TPRS conditions, the initial relative walking position of the user with respect to the walker was calculated using Eq.(4) (initial side angle θ was 30°), and users were requested to keep that position using their upper limbs' force so that the side angle θ (Fig.6-7) is no longer ignorable but changes simultaneously along with the gait. This generates a tension that pulls the user's pelvis at the start of the swing phase, as explained in Section III.D. In the case of VBWS-TPRS, this support force in the forward direction is smaller at the start of the swing phase (Eq. (19)) to avoid affecting the user's balance, while the support force in the vertical direction is the same as the VBWS scenario.

All individuals in these experiments were supported by the BWS walker while walking through a S-shape trajectory (18.5 m), once per support pattern. The trail sequences were set in a pre-defined order so that users can make a clear comparison among different methods under each BWS level.

V. RESULT AND DISCUSSION

A. Evaluation metrics

Experimental data from the sensor-integrated shoe, harness, and questionnaire surveys were measured to investigate the effect of each controlled method quantitatively and qualitatively. Joint gait motion, including transverse pelvis rotation [20] and foot switching motion [54], which are considered natural to a human gait, need to be considered. Furthermore, to apply the proposed system under real-life conditions, an analysis of usability and acceptance should be considered [55]. Therefore, 4 major output measures were taken into consideration:

- Changes in BW ratio compared to NW (Section V.B)
- Kinematics of pelvis rotation including MPA-TPR, MA-TPR, and TPM (Section V.C)
- Temporary relationship between pelvis rotation and foot landing motion (Section V.D)
- Workload and user experience (Section V.E)

We compare these output results with natural walking to provide insight into assessing the naturalness of overground BWS walking, which helps us determine the possible use cases of each method (Section V.F).

B. BWS feasibility

In Fig. 8, we can see the measured weight ratio while walking under each method for 5%, 10%, and 15% BWS,

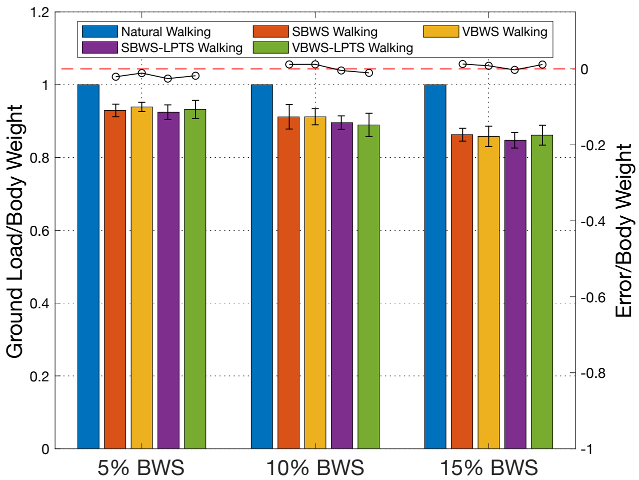


Fig. 10. Result of BWS walking. Each bar stands for the body weight ratio. Black circles above each bar stand for BWS errors relative to the user's BW. Error bar means standard deviation under each BWS method.

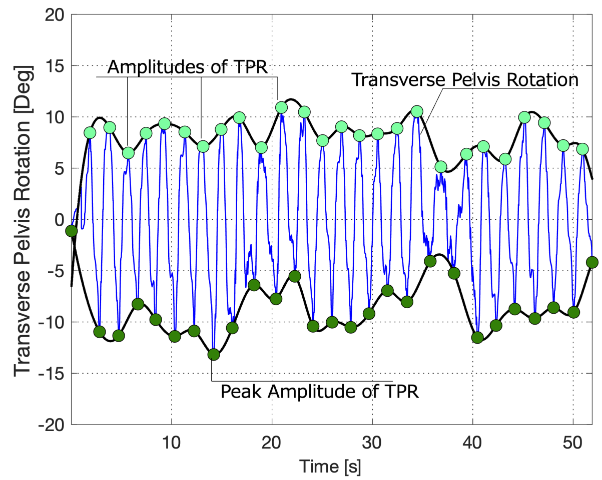


Fig. 11. Transverse Pelvis Rotation (TPR) trajectory collected using instrumented harness.

respectively. Each bar stands for the average body weight ratio, defined as the measured weight divided by the user's original body weight. We can evidence that all four control methods enable the walker to support part of the user's body weight. There was no significant difference between them. The overall average BWS error is around 1.21% ($\pm 0.22\%$) of the user's original body weight.

C. Transverse Pelvis Rotation (TPR)

In this experiment, the Mean Peak Amplitude of TPR (MPA-TPR), Mean Amplitude of TPR (MA-TPR), and Total Pelvis Movement (TPM) (Fig.9) were analyzed as the significant metrics that define the performance of the user's gait. The Mean Peak Amplitude is the average of the largest amplitude in each experimental run, the Mean Amplitude is the average of all the amplitudes in each experimental run, and the Total Pelvis Movement is the integration of the pelvis angle over time for each subject, from the start to the end of the walking motion.

Fig. 10(a) shows the mean result of MPA-TPR from 10 subjects. We found that TPR reached a peak mean amplitude under NW around $13.64°$. This value decreased around 39.65% throughout 5%$\sim$15% under SBWS walking. Under the VBWS control method, this value decreased around 35.41% overall. If we look into the MA-TPR of SBWS-TPRS and VBWS-TPRS, this value only decreased around 28.72% and 29.54%, respectively.

Next, we investigate the result of MA-TPR (Fig. 10(b)). Similar to the result of MPA-TPR, TPR reached a peak average rotation to around $6.732°$ under NW. Under SBWS and VBWS, MA-TPR decreased around the same percentage, 38.5%. However, under SBWS-TPRS and VBWS-TPRS, MA-TPR decreased to 25.1% and 31.2%, respectively.

Finally, from the result shown in Fig.10(c), pelvis motion experienced the maximum trajectory under natural walking. Under SBWS walking, the PTM get decreased accordingly (around 31.54% on average). However, we can find that the decrease was smaller under VBWS, SBWS-TPRS, and VBWS-TPRS, with values of 29.24%, 17.17%, and 18.17%, respectively.

A one-way repeated measures ANOVA using Shaffer's modified sequentially rejective Bonferroni procedure (alpha level: 0.05) shows a significant difference in the MPA-TPR ($\rho = 0.0406$) and PTM ($\rho = 0.0389$) between SBWS and SBWS-TPRS. It also shows a significant difference in the MA-TPR between SBWS, VBWS, and SBWS-TPRS (p = 0.0162). This result suggests that the SBWS-TPRS method can enable a significantly improved pelvis rotation than pure SBWS or VBWS. As a result, the user's pelvis experienced a longer moving trajectory within the same training distance. This is an important improvement since pelvic movements can affect voluntary muscle activation, improving the outcome of a gait training section [20].

From these results, we can infer that under the same desired level of BWS, the VBWS control enables users to walk more naturally under the same desired level of BWS. Especially, the pelvis rotation can get significantly larger under TPRS control, indicating that the gait performance is closer to the NW.

D. Phase Lag (PL) and single support period analysis

We looked into the left-right single support period within the gait periods by only considering the GRF signals since the single support period plays a critical role in weight-bearing and balance ability. We also evaluate the standard deviation of the relative timing relationship PL between the TPR and foot landing motion using the real-time measurement from the instrumented shoes and harness. These results can be found in Fig.12 and Fig.13.

Fig.12(d) shows the summarized single support period from all 10 participants. We found that the single support period of natural walking is around $66.12\% \pm 13.78$, under SBWS, VBWS, SBWS-TPRS, and VBWS-TPRS, the average single support period is $64.54\% \pm 15.26$, $67.73\% \pm 10.84$, $69.48\% \pm$

10.94 and $67.34\% \pm 11.70$ respectively. This result indicates that the percentage of the single support period is closest to NW under VBWS-TPRS, and VBWS, SBWS-TPRS can lead to a longer single support period. On the contrary, the SBWS method may lead to a shorter period. Moreover, the SD of PL decreases gradually under SBWS(1.844), VBWS(1.722), SBWS-TPRS (1.470), and VBWS-TPRS (1.225) control. This infers that the TPRS strategy enables users to have a more stable gait and consistent foot landing.

E. Workload (WL) and User Experience (UE)

We measured the user's workload using the NASA TLX and user experience using a 10-point scaled questionnaire (Detailed content can be found in supplementary materials). Fig. 11 summarizes the comparison of all evaluation metrics discussed earlier in this article, along with the workload and user experience compared to the NW. Note that each result of UE is obtained using the average value from our questionnaire survey, and each value is calculated using 60 samples from all the participants. Here the blue line shows the value of each evaluation metric, and the red line of the hexagon represents the value of the same metric under NW. The closer the blue lines are to the red line, the more similar the motion is to the NW.

First, we compare the results of SBWS and VBWS (Fig.11 (a)-(b)). The results indicate that under SBWS, the perceived workload is higher, and the user experience is lower than that of the NW. This is mainly because of the motion restriction on the pelvis that hinders the natural walking, which in turn requires the user to apply more torque on the pelvis, resulting in a higher workload even though part of their body weight was alleviated. On the other hand, we found that the workload was almost the same with NW, and user experience improved under VBWS. Moreover, the pelvis motion also becomes larger than SBWS, which indicates that users can walk more easily and feel more comfortable under VBWS control.

Next, we compare the above result with SBWS-TPRS and VBWS-TPRS control (Fig.11 (a)-(d)). We found the expected improvement in the MPA-TPR, MA-TPR, PTM, and PL using S(V)BWS-TPRS control methods compared with S(V)BWS control methods, indicating that a larger pelvis rotation was achieved, which can lead to a more natural gait pattern. Moreover, we found there is no significant difference between SBWS-TPRS and VBWS-TPRS method. However, under SBWS-TPRS, the perceived workload increased, and the user experience decreased compared with the other control methods. This might be caused by the suddenly increased force on the pelvis at the start of the swing phase, which in turn requires users to be more aware of their pelvis motion in order to prevent possible loss of balance. Decreasing this sudden force during the swing phase under VBWS-TPRS caused the users to have a better user experience and a reduced workload, which eventually allowed the user to use the walker more comfortably.

F. Average gait cadence and stride for walking with different BWS control methods

We analyzed the gait cadence and stride changes based on the user's foot dynamics to look into the influence of each BWS in more detail. Table III and Table IV summarized the average cadence and stride changes from 10 participants' experimental data. From the result, we confirmed that under partial BWS, SBWS could stimulate a faster stepping speed (around $32.73(\pm 5.45)$) while stride length is the smallest compared with the other three methods. This result implies a forward swing movement under SBWS is difficult due to the pelvis motion restriction.

On the other hand, the average cadence for the other three methods (VBWS: $28.60(\pm 3.98)$, SBWS-TPRS: $28.69(\pm 4.87)$, VBWS-TPRS: $27.24(\pm 4.01)$) are smaller than SBWS method, implies that a slower but stable gait can be achieved. Specifically, we confirmed that users are generally able to perform a longer stride length under VBWS $(0.65(\pm 0.18))$, SBWS-TPRS $(0.73(\pm 0.20))$ or VBWS-TPRS $(0.71(\pm 0.17))$. This implies that users may experience more time on the single/double support periods, generating more energy to complete the gait training task.

G. Possible user cases for proposed BWS control methods

Through experiments with non-disabled subjects, we confirmed that using the newly proposed control strategies, pelvis motion and stance phase duration significantly improved compared to our former SBWS and VBWS methods, although users perceive a 30% and 20% higher workload under them. Specifically, users reported a 33.9% better user experience and a 20% lower perceived workload under the VBWS-TPRS method than SBWS-TPRS, but the performance was similar under both VBWS-TPRS and SBWS-TPRS methods. Nevertheless, each method discussed until now has its pros and cons. Therefore, we need to consider which method to choose under different situations.

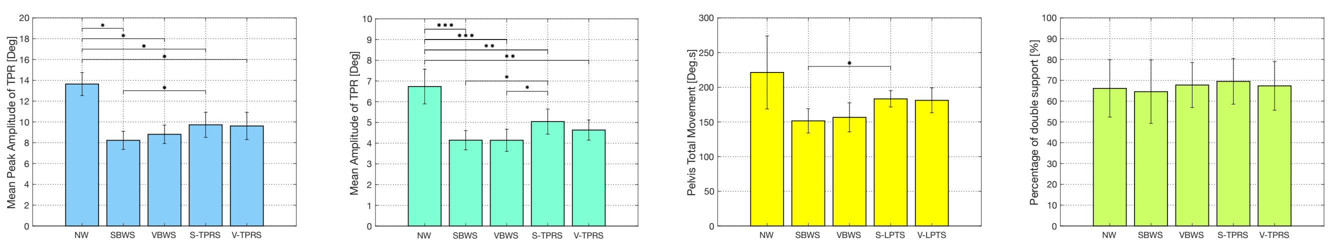
Firstly, SBWS control is the simplest approach which can easily support part of the user's body weight constantly and continuously without requiring the control of an actuator. This strategy is simple to implement and cost-effective. However, several joint motions related to a natural gait, such as transverse pelvis motion, may be restricted, which in turn can cause a higher workload and hinder the natural gait feeling. By simply modifying the position of the user relative to the walker, SBWS-TPRS control has the potential to support the pelvis rotation and increase the period of the foot contact phase significantly. Nevertheless, upper limb strength is required to compensate for the forward tension and balance the pelvis rotation during the swing phase. For users who need BWS gait training but have a limited budget in health care and social services, a robot with the SBWS control may be the right choice as it is easy to build and simple to use. Furthermore, the user can utilize their upper limb strength to support pelvis rotation (SBWS-TPRS) and simulate a more natural, stable gait pattern, even without the assistance of medical staff.

TABLE III
AVERAGE GAIT CADENCE (STEP/MIN) FOR EACH ABLE SUBJECT (AS) UNDER PROPOSED BWS METHODS

AS index	NW	SBWS	VBWS	SBWS-TPRS	VBWS-TPRS
AS1	29.77	32.48(±1.15)	30.20(±1.49)	28.92(±0.02)	26.57(±2.13)
AS2	27.35	36.56(±6.94)	31.62(±8.48)	32.40(±6.10)	28.09(±2.27)
AS3	37.20	23.67(±9.36)	22.14(±10.16)	22.53(±9.88)	20.87(±11.3)
AS4	30.54	26.77(±4.20)	26.49(±3.88)	24.32(±5.59)	25.73(±4.26)
AS5	38.91	36.01(±3.13)	33.89(±4.48)	35.70(±3.22)	35.69(±3.19)
AS6	40.16	30.80(±3.13)	30.45(±2.54)	30.40(±2.56)	28.19(±1.97)
AS7	35.13	29.55(±5.26)	29.21(±5.54)	27.14(±7.18)	25.32(±8.65)
AS8	31.18	36.01(±3.13)	25.77(±4.79)	25.24(±5.17)	25.65(±4.81)
AS9	34.34	27.10(±8.59)	23.43(±6.07)	24.10(±11.21)	24.91(±12.87)
AS10	22.68	37.39(±3.51)	32.84(±2.40)	36.14(±1.04)	31.38(±1.47)
Average	32.73(±5.45)	31.63(±4.82)	28.60(±3.98)	28.69(±4.87)	27.24(±4.01)

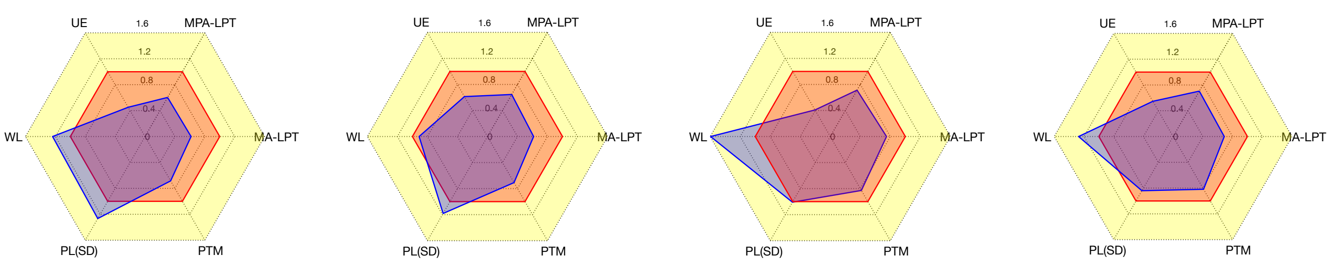
TABLE IV
AVERAGE STRIDE (M) FOR EACH ABLE SUBJECT (AS) UNDER PROPOSED BWS METHODS

AS index	NW	SBWS	VBWS	SBWS-TPRS	VBWS-TPRS
AS1	0.62	0.38(±0.10)	0.57(±0.27)	0.53(±0.08)	0.54(±0.03)
AS2	0.58	0.61(±0.09)	0.73(±0.14)	0.82(±0.04)	0.88(±0.04)
AS3	0.80	0.81(±0.09)	0.65(±0.03)	0.72(±0.02)	0.69(±0.02)
AS4	0.69	0.72(±0.04)	0.70(±0.05)	0.80(±0.16)	0.76(±0.02)
AS5	0.69	0.65(±0.05)	0.76(±0.04)	0.63(±0.05)	0.61(±0.16)
AS6	0.46	0.48(±0.26)	0.45(±0.16)	0.68(±0.28)	0.70(±0.10)
AS7	0.80	0.90(±0.03)	0.88(±0.04)	1.12(±0.26)	0.99(±0.06)
AS8	0.74	0.71(±0.12)	0.63(±0.20)	0.77(±0.10)	0.82(±0.06)
AS9	0.84	0.76(±0.08)	0.77(±0.03)	0.82(±0.14)	0.76(±0.21)
AS10	0.50	0.33(±0.02)	0.33(±0.02)	0.38(±0.04)	0.38(±0.02)
Average	0.67(±0.13)	0.64(±0.19)	0.65(±0.18)	0.73(±0.20)	0.71((±0.17))



(a): Result of MPA-TPR (b): Result of MA-TPR (c): Result of PTM (d): Result of single-support period

Fig. 12. Walking performance according to TPR (5.0% ∼ 15% BWS, $^{*}\rho < 0.05$, $^{**}\rho < 0.01$, $^{***}\rho < 0.001$)



(a): NW versus SBWS (b): NW versus VBWS (c): NW versus SBWS-TPRS (d): NW versus VBWS-TPRS

Fig. 13. Comparison of comprehensive natural gait evaluation between natural and BWS walking

Next, we discussed a synchronously modulated BWS control strategy called VBWS to provide users with a more natural gait pattern with a lower workload and a better user experience. By synchronizing the controlled BWS with real-time gait events, users can exhibit a more natural gait including a better transverse pelvis rotation and longer stance phases. Furthermore, by adjusting their relative position to the walker, users can use the VBWS-TPRS control strategy, which provides a mild support on the pelvis directly during the swing phase. Using this method, we found that users exhibit an improved gait performance compared to pure VBWS and a gait performance comparable to that under the SBWS-TPRS control. Nevertheless, the workload is higher than VBWS control, which probably derives from keeping a fixed relative position concerning the walker and the added force in the pelvis that requires users to be attentive to prevent loss of balance.

Furthermore, experiment results with non-disabled subjects discussed in this article also show the potential to assist normal walking for stroke patients. Since only using the lower limb on the non-paretic side for a long time may gradually weaken both the upper and lower extremities [56], customized BWS on the paretic side may help them maintain an upright spine and ensure a stable gait. Additionally, instead of restricting patients to completely passive robotic gait training, the proposed system can provide patients with a certain amount of freedom for transverse pelvis movement, especially under the VBWS control. On the other hand, patients with normal upper limb extremities may benefit from SBWS-TPRS or VBWS-TPRS controls as a cost-effective method to support pelvis rotation without further concerns of reduced upper limb muscle activation.

The developed BWS system and control strategies successfully demonstrated the ability to stimulate natural walking with increased lower extremity activity without using additional actuators. Such a support strategy can potentially be a complementary method for gait training. Nevertheless, long-term clinical investigations with disabled subjects are still necessary in order to draw a solid conclusion. The findings of the present study warrant the further investigation of such strategies to help patients recover their walking ability.

VI. CONCLUSION

In this article, we have presented a mobile BWS walker to support overground gait training. We integrated the system with a pair of instrumented shoes and an IMU-equipped harness to measure the vertical GRF and TPR signals in real-time. We implemented four BWS controls to either support the body weight only or protect/assist the pelvis rotation simultaneously while only using 2 actuators.

The overall experimental result of this study shows that proposed BWS controls have the potential to become a complementary method in gait rehabilitation. Specifically, the SBWS strategy provides users with a simple and convenient way to relieve part of their weight, and it is easy to be implemented with any walking device. Users who need prolonged and consistent BWS training but with limited training space and budget in health care services may benefit from this strategy. Users may also benefit from the SBWS-TPRS strategy by utilizing their upper limb strength to support their transverse pelvis rotation and simulate a relatively natural and stable gait pattern without using additional actuators or devices. Furthermore, by wearing the instrumented shoes and harnesses, users can easily access their gait training performance data using real-time vertical GRF and pelvis kinematics feedback signals, which provides useful information for skilled therapists to monitor and track their patient's recovery status. Finally, these feedback signals can also be utilized to control the robot's motion in real-time using VBWS or VBWS-TPRS method, which helps users to generate a more natural gait including significantly improved pelvis rotation, pro-longed single support phase, and improved stride length with a lower workload and better user experience compared with SBWS or SBWS-TPRS method.

This study presents new insights related to the use of BWS on overground gait training while helping users to maintain a natural gait. However, some limitations should be considered. Firstly, this study only examined the pelvis rotation in the transverse plane, which might not account for all the pelvis changes in gait patterns. Further investigation with a full analysis of gait pattern changes should be included. Second, the findings in this study suffer from a small number of non-disabled subjects, although the experimental results showed positive improvement in pelvis rotation during gait training. Confirmatory clinical trial targeting hospital or rehabilitation center is planned in the future with more subjects. Lastly, upper limb strength is needed under TPRS controls, which may cause a higher workload, especially being supported under a higher level of BWS.

Future work with the proposed hardware system will focus on producing the forward cable tension automatically without requiring force from the user's upper limb and friction with the ground. We expect to overcome this limitation by designing a passive trunk surrounding system mounted on the center of the walker that passively keeps the distance between the user and robot while providing freedom of movement in the rotational direction. We believe this design can release the workload on the upper limbs, and the user can feel a sense of security because the robot moves and responds synchronously according to the user's gait. We would also like to design a variable BWS controller customized to each individual's needs and gait performance, such that our system can adapt to users with different gait cadences and step lengths. Future studies should carry out long-term investigations into the effects of proposed methods on different populations with gait impairments.

VII. APPENDIX

The supplementary materials including questionnaires and demo video are available from Google Drive: https://bit.ly/3FRvKDc. Before the experiment, our research purpose, method and data handling were fully explained to participants, and we obtained their informed consent. All the research has taken place with the approval of the Ethics Committee on Research Involving Human Subjects of the Graduate School of Engineering, Tohoku University, Japan.

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