

A New Active Body Weight Support System Capable of Virtually Offloading Partial Body Mass

Qi Lu, Jianxun Liang, Bing Qiao, and Ou Ma

Abstract—This paper presents a novel active body weight support (BWS) method, which is capable of virtually offloading full or partial body mass for a potential application of enhancing treadmill-based locomotion rehabilitation. The mass-offloading capability is realized by actively compensating a desired amount of body weight and inertia force using an acceleration-feedback scheme. The method has been studied by dynamics simulations and a specially designed nonhuman experiment. In the simulation study, the human and the BWS device were modeled as a multi-body dynamical system interacting dynamically with the treadmill. The ground reaction forces were recorded as the dynamic load on the person. A cam-slider-based experiment was designed and conducted to test the engineering feasibility of the mass-offloading capability. Both the simulation and experiment results demonstrated that the BWS system can compensate any desired amount of gravity force and inertia force and, therefore, has the effect of virtually reducing the mass of a person attached to the system.

Index Terms—Body mass offloading, body suspension, body weight support (BWS), dynamic load, human body modeling, locomotion training, mass-offload BWS, rehabilitation.

I. INTRODUCTION

BODY weight support (BWS) system is a key component in the treadmill-based neurorehabilitation therapy for patients with lower limbs neurological impairments. It is used to assist therapists in locomotor training of patients in both manual and robot-assisted therapies (see [1] and [2] for review and [3]–[5] for some examples). The main function of the BWS is to offload part of the weight of the patient during walking so that the burden on the patient due to his/her weight would be reduced. Furthermore, the BWS system ensures the safety and stability of the patient during walking. Research has shown that BWS-assisted locomotion training can improve the rehabilitation outcomes [6]–[8].

A BWS system usually consists of a harness, cables, pulleys, and a counterbalancing device to offload a patient's weight. The

patient walking on the treadmill is supported through the harness, which is attached to an overhead suspension cable pulley connected to the counterbalancing device. Regarding the differences of the counterbalancing devices, current BWS systems can be classified into two types: the passive BWS and the active BWS. In the former, as illustrated in Fig. 1(a), passive elements, such as counterweights [9] or springs [10], are used to balance partial weight of the patient. The latter, as depicted in Fig. 1(b), uses actively controlled electric [11], [12], pneumatic [13], or hydraulic [14] actuators to offload part of a patient's weight. A passive BWS device can only statically reduce the weight of a patient. It does not reduce the dynamic load (i.e., the inertia force) exerted on the patient's body while the patient is doing activities on the treadmill. On the contrary, it may also lead to poor offloading accuracy and dynamic performance due to the added counterweights or springs [15], [16]. On the other hand, an active BWS system can significantly improve the offloading accuracy and dynamic performance of the weight-reduction capability [11]–[13], and they are able to seamlessly adjust the amount of the offloaded weight without stopping the operation of the device. However, the existing active BWS systems are not able to reduce dynamic load either, because they were designed to compensate a predetermined amount of body weight only. Thus, the technique is referred to as the active weight-offload BWS in this paper. It should be noted that the offloaded weight in the early phase of a therapy is very large (usually 60–80% of the body weight) [11] because of the weak muscular strength of the patient in this stage. At such a large offloading level, all the existing weight-offload BWS devices can result in very large dynamic loads on the patients because they are incapable of reducing dynamic loads.

For successful establishment of a walking gait, a patient needs to adapt not only to the static load (weight) but also to the dynamic load (inertia force), which is dependent on mass, velocity, and acceleration. This is because weight and mass have independent effects on human flexors during locomotion [17]. One can easily conceive that the intuitive approach for simultaneously reducing both the weight and the inertia load is to (virtually) reduce the body mass. Obviously, if certain amount of body mass were removed, the inertia force and the weight of the patient corresponding to that amount of body mass would be simultaneously offloaded too. This goal can be realized by designing a BWS system with an acceleration-feedback capability [18]. The idea is to actively control the tension in the cable, which suspends the patient's torso using an acceleration-feedback strategy, as conceptually illustrated in Fig. 1(c). Since the proposed active BWS system can make the same dynamic effect on the patient's body as if a part of his/her body mass were removed.

Manuscript received April 20, 2011; accepted June 6, 2011. Date of publication August 4, 2011; date of current version September 12, 2012. Recommended by Technical Editor E. Richer.

Q. Lu and J. Liang are with the Department of Mechanical and Aerospace Engineering, New Mexico State University, Las Cruces, NM 88003 USA (e-mail: qilu@nmsu.edu; ljx1209@nmsu.edu).

B. Qiao is with the Department of Astronautic Control, Nanjing University of Aeronautics and Astronautics, Nanjing, 210016, China (e-mail: bqiao@nuaa.edu.cn).

O. Ma is with the NMSU Reduced-G and Biomechanics Laboratory, Department of Mechanical and Aerospace Engineering, New Mexico State University, Las Cruces, NM 88003, USA (e-mail: oma@nmsu.edu).

Color versions of one or more of the figures in this paper are available online at <http://ieeexplore.ieee.org>.

Digital Object Identifier 10.1109/TMECH.2011.2160555

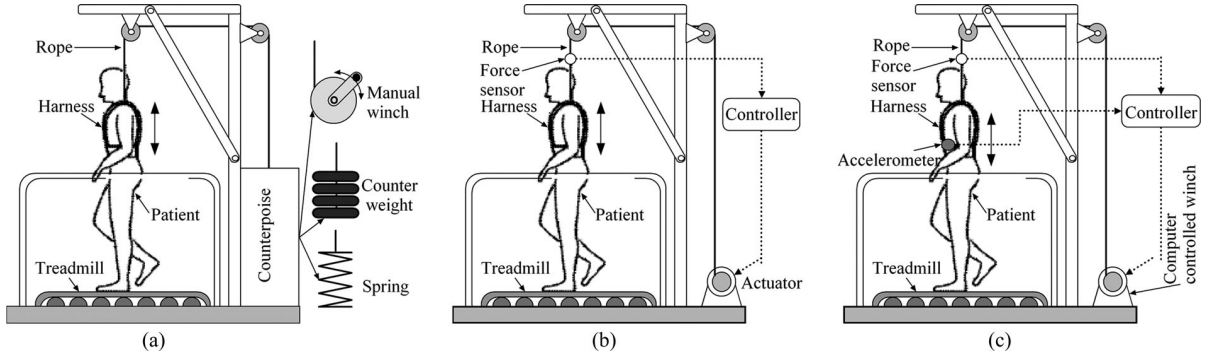


Fig. 1. BWS systems for treadmill-based locomotion training. (a) A passive BWS. (b) An existing active weight-offload BWS. (c) The proposed mass-offload BWS.

For this reason, it is referred to as the *active mass-offload BWS* in this paper.

The original idea of the mass-offload BWS was introduced in [16], which is different from that of the passive BWS and of the active weight-offload BWS. Obviously, whether or not a mass-offload BWS device can outperform weight-offload BWS devices for locomotion training can only be concluded from a clinical study. However, before a clinical study can be conducted, an engineering feasibility has to be studied first. The research work described in this paper is for such a feasibility study. The study consists of a multibody dynamics simulation and a specially designed nonhuman experiment.

In the simulation study, the physical human is modeled as a multibody system with multi-degrees of freedom (DOFs), which dynamically interacts with the treadmill. The ground reaction forces are recorded during the simulation to assess the dynamic load on the patient. In order to demonstrate the advantages of the proposed active BWS in dealing with the dynamic load on the human body, all three types of BWS systems, namely, the active mass offload, active weight offload, and passive BWS, are simulated with the same human body and walking kinematics. After the simulation, a cable suspended cam-slider system is designed, which can approximate the walking gait of a human. The experimental results demonstrated that the proposed active mass-offload BWS system can simultaneously offload both the weight and the dynamic loads of the walking human and, thus, a person suspended in such a BWS system would dynamically experience the situation as if a certain amount of his/her mass were removed from a mechanical engineering perspective.

The rest of the paper is organized as follows. The theoretical foundation of the mass-offload BWS is introduced in Section II. In Section III, the computer simulation of the proposed mass-offload BWS, which includes the modeling of both the human body and the BWS mechanism, is described. The nonhuman experimental study is presented in Section IV and the conclusion and discussion are presented in Section V.

II. VIRTUALLY OFFLOADING BODY MASS

The body gait of human walking is a complicated 3-D periodic function normally with a frequency of 0.5–1.5 Hz and a magnitude of 0–0.1 m depending on the height, speed, and walking pattern of the person [19]. A body gait in 3-D space

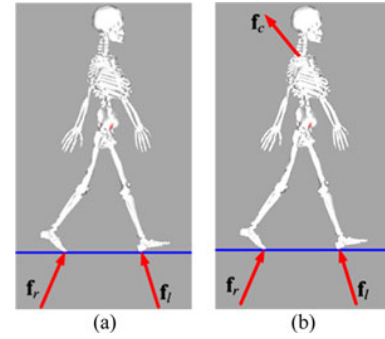


Fig. 2. Human body in walking. (a) Walking with no BWS. (b) Walking with BWS.

may be expressed as the following general time function:

$$\mathbf{p}_0 = \mathbf{p}_0(t). \quad (1)$$

A human body including arms and legs can be assumed to be an n -DOF multibody system. For a human body walking with no BWS [shown in Fig. 2(a)], the system equations of motion can be written in the following general form:

$$\mathbf{M}(\mathbf{q})\ddot{\mathbf{q}} + \boldsymbol{\tau}_n(\mathbf{q}, \dot{\mathbf{q}}) + \boldsymbol{\tau}_g(\mathbf{q}) = \boldsymbol{\tau} + \mathbf{J}_l^T \mathbf{f}_l + \mathbf{J}_r^T \mathbf{f}_r \quad (2)$$

where $\mathbf{q} \in R^n$ is a vector of the generalized coordinates defining joint displacements; $\mathbf{M}(\mathbf{q}) \in R^{n \times n}$ is the configuration-dependent (i.e., function of $\mathbf{q}$) generalized inertia matrix; $\boldsymbol{\tau}_n(\mathbf{q}, \dot{\mathbf{q}}) \in R^n$ is a vector of the velocity-related generalized nonlinear inertia torques; $\boldsymbol{\tau}_g(\mathbf{q}) \in R^n$ is a vector of the generalized gravity torques; $\boldsymbol{\tau} \in R^n$ is a vector of the generalized joint torques, which are generated by human muscles to actuate the joints; $\mathbf{f}_l \in R^3$ and $\mathbf{f}_r \in R^3$ are the ground reaction forces against the left and right feet, respectively. In this study, the ground reaction force on each foot is approximated as a concentrated force. $\mathbf{J}_l \in R^{3 \times n}$ and $\mathbf{J}_r \in R^{3 \times n}$ are the Jacobian matrices mapping all the joint rates to the Cartesian velocities of the left and right feet, respectively.

When a human walks with a BWS device [see Fig. 2(b)], an additional and unidirectional force will act on his/her body through the suspension cable. As a result, the equations of motion become

$$\mathbf{M}(\mathbf{q})\ddot{\mathbf{q}} + \boldsymbol{\tau}_n(\mathbf{q}, \dot{\mathbf{q}}) + \boldsymbol{\tau}_g(\mathbf{q}) = \boldsymbol{\tau} + \mathbf{J}_l^T \mathbf{f}_l + \mathbf{J}_r^T \mathbf{f}_r + \mathbf{J}_c^T \mathbf{f}_c \quad (3)$$

where the last term represents the projection of the Cartesian cable force $\mathbf{f}_c \in R^3$ on the joint space and $\mathbf{J}_c \in R^{3 \times n}$ is the configuration-dependent Jacobian matrix mapping the joint rates to the Cartesian velocity of the effective point of the human body on which the resultant cable force exerts. The cable force can be modeled as

$$\mathbf{f}_c = f_t \mathbf{u} \quad (4)$$

where $\mathbf{u} \in R^3$ is the unit vector along the direction of the cable and $f_t \geq 0$ is the cable tension. For an active BWS system, f_t is the force provided by the computer-controlled winch. For a counterweighted (with a mass of m_2) BWS system, f_t is simply the following:

$$f_t = m_2 g - m_2 a \quad (5)$$

where g is the gravitational acceleration and a is the acceleration of the counterweight, which equals to the vertical component of the acceleration of the human body if ignoring the elasticity of the cable and human. Based on the definition of the Jacobian matrices and the fact that all the three terms on the left-hand side of (3) are proportional to the mass of the human body m_1 , we have

$$\mathbf{J}_c^T \mathbf{f}_c = \mathbf{J}_c^T (m_2 g - m_2 a) \mathbf{u} = \alpha \tau_g - m_2 \mathbf{J}_c^T (\dot{\mathbf{J}}_c \dot{\mathbf{q}} + \mathbf{J}_c \ddot{\mathbf{q}}) \quad (6)$$

where $\alpha = m_2 / m_1$ is the offloading level (i.e., the ratio of the counterweight mass over the patient mass). Thus, the dynamics equation of the passive counterweighted BWS system becomes

$$(\mathbf{M} + m_2 \mathbf{J}_c^T \mathbf{J}_c) \ddot{\mathbf{q}} + (\boldsymbol{\tau}_n + m_2 \mathbf{J}_c^T \dot{\mathbf{J}}_c \dot{\mathbf{q}}) + (1 - \alpha) \boldsymbol{\tau}_g = \boldsymbol{\tau} + \mathbf{J}_l^T \mathbf{f}_l + \mathbf{J}_r^T \mathbf{f}_r. \quad (7)$$

Comparing (7) to the free-walking dynamics (2), we can see that, when the generalized gravity force is reduced by α percentage as expected, the resultant generalized inertia forces are also increased due to the extra mass of the added counterweight by an amount of

$$m_2 \mathbf{J}_c^T \mathbf{J}_c \ddot{\mathbf{q}} + m_2 \mathbf{J}_c^T \dot{\mathbf{J}}_c \dot{\mathbf{q}}. \quad (8)$$

In other words, a passive BWS system cannot reduce the dynamic load. Instead, it will increase the dynamic load.

On the other hand, consider an active weight-offload BWS system, which provides a constant cable tension as following:

$$f_t = \alpha m_1 g \quad \text{or} \quad \mathbf{J}_c^T \mathbf{f}_c = \alpha \tau_g. \quad (9)$$

Substituting (9) into (3) gives the dynamics equations of such a weight-offload BWS system

$$\mathbf{M}(\mathbf{q}) \ddot{\mathbf{q}} + \boldsymbol{\tau}_n + (1 - \alpha) \boldsymbol{\tau}_g = \boldsymbol{\tau} + \mathbf{J}_l^T \mathbf{f}_l + \mathbf{J}_r^T \mathbf{f}_r. \quad (10)$$

Obviously, the active weight-offload BWS affects only the generalized gravity forces and has nothing to do with the generalized inertia force terms. Thus, although the active weight-offload BWS system will not add an extra inertia load to the human body as a passive counterweighted BWS does, it maintains the same dynamic load on the human as if no body weight had been offloaded.

The dynamic loads can be significant at the peak acceleration of a walking gait, especially, at an early phase of a therapy treatment. These dynamic forces will put extra burden and cause

discomfort on the patient. To solve the problem, an actively controlled mass-offload BWS system is designed. The goal of the system is to virtually reduce the mass of the human body by applying active force control to the cable drive. Based on the detailed structure of the free-walking dynamics (2), it is understood that if the true mass of a free-walking subject is reduced by a factor of α ($0 \leq \alpha \leq 1$), then the resulting dynamics equations are simply scaled down as

$$(1 - \alpha)[\mathbf{M}(\mathbf{q}) \ddot{\mathbf{q}} + \boldsymbol{\tau}_n(\mathbf{q}, \dot{\mathbf{q}}) + \boldsymbol{\tau}_g(\mathbf{q})] = \boldsymbol{\tau} + \mathbf{J}_l^T \mathbf{f}_l + \mathbf{J}_r^T \mathbf{f}_r. \quad (11)$$

This means, if we design an active BWS, which can produce

$$\mathbf{J}_c^T \mathbf{f}_c = \alpha[\mathbf{M}(\mathbf{q}) \ddot{\mathbf{q}} + \boldsymbol{\tau}_n(\mathbf{q}, \dot{\mathbf{q}}) + \boldsymbol{\tau}_g(\mathbf{q})] \quad (12)$$

then the dynamics equations, i.e., (3), of a person with a BWS system will be identical to that of a reduced-mass person on free walking, i.e., (11). This shows in theory that the resulting mass-offload BWS system can make the human biomechanically (not physiologically) feel as if his/her mass were really reduced by a factor of α . Therefore, (12) is the control goal of the proposed active BWS system in joint space. Since the cable control is done in Cartesian space, (12) can be converted into Cartesian space as follows:

$$\mathbf{f}_c = \alpha[\mathbf{M}_x(\mathbf{x}) \ddot{\mathbf{x}} + \mathbf{f}_n(\mathbf{x}, \dot{\mathbf{x}}) + \mathbf{f}_g(\mathbf{x})] \quad (13)$$

where $\mathbf{x} \in R^3$ is a Cartesian-space vector representing the position of the effective point of the human body, and $\mathbf{M}_x \in R^{3 \times 3}$, $\mathbf{f}_n \in R^3$, and $\mathbf{f}_g \in R^3$ are the inertia matrix, the velocity-related inertia force, and the gravity force in Cartesian space, respectively. Equation (13) defines the control force that the mass-offload BWS controller has to provide in order to make the system behave exactly as if a mass of m_2 had been removed. The first term of (13), i.e., $\mathbf{M}_x \ddot{\mathbf{x}}$, means that an acceleration feedback is needed in the cable control system.

It should be pointed out that a cable can only pull (tension) not push (compression). This is known as the force closure problem in cable-driven robots [20]. In our case, the cable will always be in tension since the control force will always pull the cable. This is because a push force happens only if the control force shown in Fig. 2(b) is pointing downward. This means that the patient has a downward acceleration greater than the gravitational acceleration, which is physically impossible. As a result, the physical law guarantees that the control law will never intent to push the cable. Therefore, there is no need to worry about the force closure problem.

From the earlier theoretical discussion, one can conclude that an active BWS system with an acceleration feedback can make the patient biomechanically experience less body mass while doing activities with a mass-offload BWS system. Obviously, such effect cannot be achieved by a passive or active weight-offload BWS system.

Force control target (13) requires implementing a human dynamics model as well as intensive computations because of the complexity of the model. In fact, the n -DOF dynamics model is presented only for theoretical analysis and implementing computer simulations purposes. Since the cable drive is only a 1-DOF device, the force control target can be practically

simplified to

$$f_t = m_2 g + m_2 a \quad (14)$$

where m_2 is the mass to be removed. To achieve the control goal, a force controller is needed in the proposed mass-offload BWS system. On the other hand, one may wish the patient's body to follow a prescribed body gait in the neurological training process, which may be described as a motion trajectory of the mass center or a body-fixed point of the human body. The force controller, acting alone, cannot achieve this goal because the controller does not have a capability of tracking a desired gait. To solve this problem, robotic assistance (e.g., the ALEX in [21]), which implement some of the control strategies as described in [22], may be desirable. The goal of the added control loop is to guarantee that the body motion will track a desired input gait $p_0(t)$ along the vertical direction.

III. SIMULATION STUDY

In this section, a high-fidelity dynamics model is presented for a simulation-based investigation of the proposed new BWS method. The investigation is aimed only at proof of the engineering principle of the new technology as opposed to a proof of any medical efficacy.

A. Dynamics Modeling of a Human Body

It is known that a physical human body can be modeled as a multibody system [23], [24]. In such a model, the Newtonian dynamics or other approaches are employed to establish the equations of motion of the human body [23]. Since multibody dynamics models are very complicated, much effort has been devoted to developing efficient solutions to the associated inverse and forward dynamics problems [25], [26]. Recently, some commercial software, such as LifeMOD [27] and AnyBody [28], become available to significantly help the modeling and simulation of physical human activities. In our research, LifeMOD and MSC Adams were used.

LifeMOD is a plug-in toolbox for the commercially available multibody dynamics simulator MSC Adams. The human body in LifeMOD consists of 19 segments articulated by 18 joints. Each segment consists of a group of bones as they are located in human anatomy. To simplify the analysis, the segments are modeled as rigid bodies, and the mass properties of the segments are defined based on the anthropometric database and automatically set up by the software. The 18 joints are the upper neck, lower neck, thoracic and lumbar on spinal, scapular, shoulder, elbow, and wrist on left and right arms, and hip, knee, and ankle on left and right legs. Although each joint contains 3 DOFs and the model provides 54 DOFs in all, only 39 of them are involved by the default setting in the software. The other DOFs are locked either because they do not physically exist (e.g., the frontal and transverse rotation of the knee), or they are not necessary in the simulation (e.g., some of the DOFs in the neck). With this model, the motion of the human body (e.g., walking) can be simulated as long as the kinematics parameters of the motion are available.

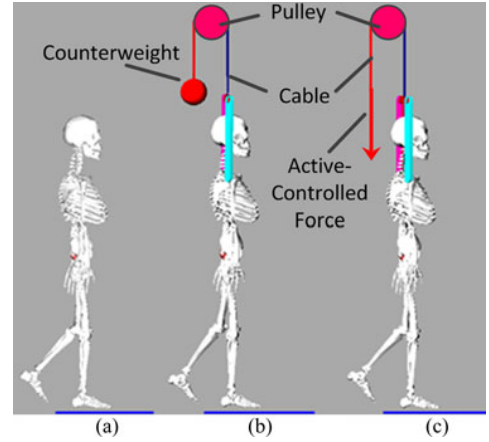


Fig. 3. Models of human walking with and without BWS. (a) No_BWS. (b) PBWS. (c) AWBWS & AMBWS.

TABLE I
PARAMETERS OF THE SIMULATIONS

Simulation Case	BWS	Body mass	Mass/weight offloading
No_BWS	N/A	$(1 - \alpha)m_1^*$	N/A
PBWS	Counterweight	m_1	$m_2 = \alpha m_1^{**}$
AWBWS	Active weight-offload	m_1	$f_t = \alpha m_1 g^{***}$
AMBWS	Active mass-offload	m_1	$f_t = \alpha m_1 (g + a)^{***}$

* $m_1 = 75 \text{ kg}$, $\alpha = 30\%$, 50% , and 70% in trials 1, 2, and 3, respectively.

** m_2 is the mass of the counterweight of the passive BWS case.

*** f_t is the actively-control force of the active BWS cases.

B. Dynamics Modeling of the BWS System

The BWS system is modeled as a cable-pulley apparatus, to one end of which the human model is connected through a massless link and two spherical joints on the shoulders of the human. The other end of the cable-pulley apparatus is either connected to a counterweight, as in a passive BWS system shown in Fig. 3(b), or applied by an actively controlled force as in an active BWS system shown in Fig. 3(c). Fig. 3(a) shows a free-walking human model without a BWS system. The physical interactions between the feet of the human and the treadmill are modeled using the contact dynamics capability provided by MSC Adams.

C. Simulation Results

In order to compare the proposed mass-offload BWS with the passive and the active weight-offload BWS, four different simulation cases are discussed here, namely, free walking (No_BWS), walking with a passive BWS (PBWS), walking with an active weight-offload BWS (AWBWS), and walking with an active mass-offload BWS (AMBWS). In this study, each case has three trials, which simulate three levels of mass or weight offloading (i.e., $\alpha = 30\%$, 50% , and 70%). The conditions of these 12 trials are summarized in Table I. It should be noted that in all the trials, the human models are driven by the same set of motion capture data extracted from the CMU human motion database. By doing so, the human models will have identical kinematics at the same time instance. In other words, the human models in different trials will perform exactly the same gait pattern as well as the same joint trajectories [i.e., the generalized

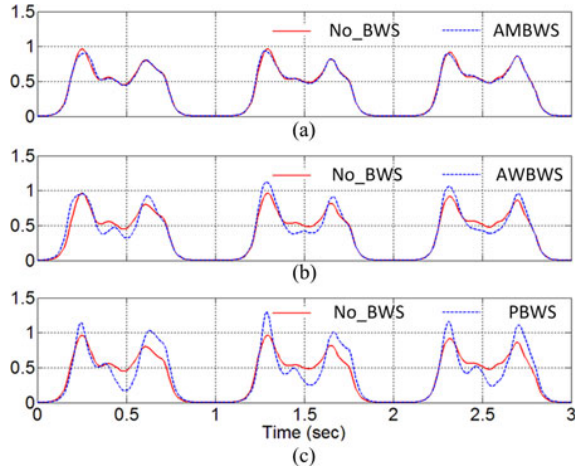


Fig. 4. Normalized ground reaction forces on the left foot ($\alpha = 30\%$).

coordinates $\mathbf{q}$ in (2)], such that the differences in the ground reaction forces and the joint torques are solely due to the different inertia properties of the entire system including both the human and the BWS device.

During the simulations, the ground reaction forces and the sagittal-plane joint torques in the ankle, knee, and hip joints are collected for the study.

1) *Ground Reaction Forces:* The normalized ground reaction forces on the left foot in each of the simulation cases with 30% offloading are plotted in Fig. 4. Fig. 5 depicts the normalized ground reaction forces acting on both feet of the human in the simulation cases of 50% offloading.

As illustrated in Fig. 4(a), the ground reaction forces in the AMBWS case have a pattern almost identical to that in the No_BWS (free walking) case. This indicates that a human walking with the 30%-offloading AMBWS device does experience the same dynamic load as a human of 70% body mass walking without the BWS device in terms of the ground reaction force. On the contrary, a larger fluctuation of dynamic ground reaction forces is observed in the PBWS case compared to that in the other two cases. This is because the extra mass of the counterweight produces an extra inertia force proportional to the acceleration and mass of the counterweight. When this extra inertia force increases to a certain degree due to a larger counterweight mass or a greater body acceleration, both feet of the human would momentarily lose contact with the ground, resulting in zero ground reaction forces, if he/she wants to keep the normal gait pattern, as illustrated in Fig. 5(d). Meanwhile, as shown in Fig. 4(b), although the AWBWS does not introduce as large dynamic loads as a PBWS does, it still has significant effects on the ground reaction forces because of the same human mass. This effect will grow more significantly if the offloading level α is further increased.

In the 50% offloading trials, it is obvious from Fig. 5(c) that the ground reaction forces in the AMBWS case are correspondingly reduced as expected, and it has the same pattern as that in the No_BWS case. However, as one can see from Fig. 5(c), the ground reaction forces in the PBWS case do not behave as what

is intuitively expected. It has not only an odd pattern but also much larger peak values when compared with that in Fig. 5(a). It is even more concerned in Fig. 5(c) that the ground reaction forces on both feet are simultaneously zero in a noticeable duration of each step cycle (e.g., in the time interval of 1.0–1.2 s, etc.). This indicates that both feet lose contact with the ground for a short period of time in each strike. If that happens, the person cannot walk properly in a normal gait. In other words, when the counterweight in a PBWS system is too heavy, the human subject may have to change his/her walking gait in order to cope with the significantly higher dynamic loads due to the mass of the counterweight. On the contrary, the simulation shows that an AMBWS device does not have such a problem. Of course, this simulation finding will have to be experimentally verified in the future.

2) *Joint Torques:* The joint torques in the hip, knee, and ankle joints in the sagittal plane are also recorded in all simulation trials. The torques for the human walking with different BWS devices are compared with those of a free walking human having less mass (i.e., the No_BWS case) at three offloading levels. The differences between joint-torque trajectories were evaluated by an index called average difference value AD, which is computed as

$$AD = \frac{1}{N} \sum_{k=1}^N |\tau_{BWS}(k) - \tau_{No_BWS}(k)| \quad (15)$$

where $\tau_{BWS}(k)$ and $\tau_{No_BWS}(k)$ are respectively the joint torque from one of the BWS cases and that from the No_BWS case at the k th simulation output data point and N is the total number of the data points. The AD value, which has the same unit as the torque, provides an overall measurement of the differences between the joint torques of the BWS case and of the No_BWS case. The AD values of the hip, knee, and ankle joints at different offloading levels are compared in Fig. 6.

Notice that the joint torques are actually the effort the patient should make to follow a prescribed gait. As one may note that, at all offloading levels, the AMBWS case has the smallest AD value in all joints. This means that, to follow a normal gait pattern, the patient with an active mass-offload BWS device just needs to provide almost the same effort as he/she needs to provide in the case of actual less mass. On the contrary, the AD values of PBWS and AWBWS cases are much higher at all offloading levels, which means that the patient has to provide more effort. The comparison shows that the active mass-offloading BWS system will benefit the patient in the sense of less joint torques or less dynamic loads, while both the passive and active weight-offload BWS systems will make the patient subject to more dynamic loads.

In the simulation of the active BWS cases, ideal acceleration signals were assumed for feedback control, and therefore, just a simple controller was applied in the closed-loop dynamics system. In reality, signals from sensors could be noisy and inaccurate, and thus, the simple controller may not produce similarly good results. Therefore, a more advanced control strategy may be needed for the actual implementation of a physical prototype of the mass-offload BWS system. A patient walking on

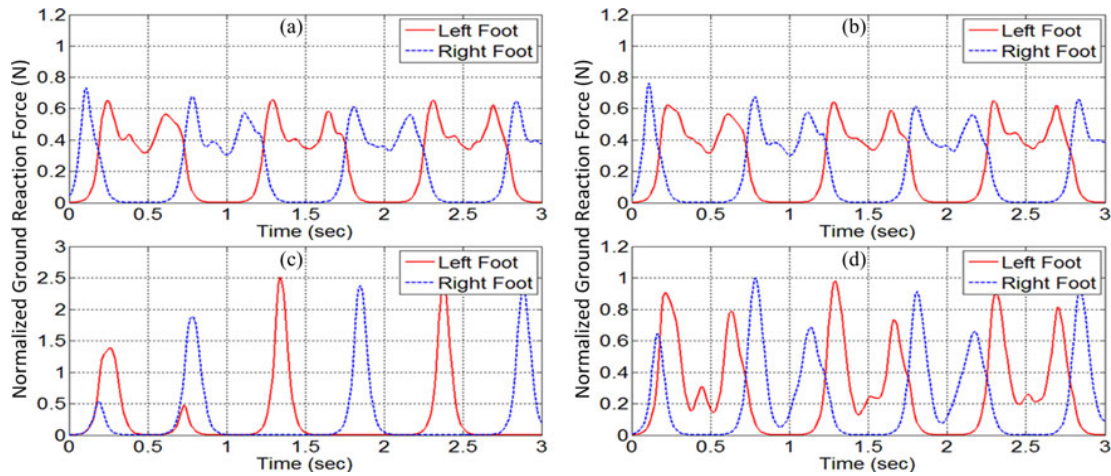


Fig. 5. Normalized ground reaction forces on both feet in (a) No_BWS, (b) AMBWS, (c) PBWS, and (d) AWBWS ($\alpha = 50\%$).

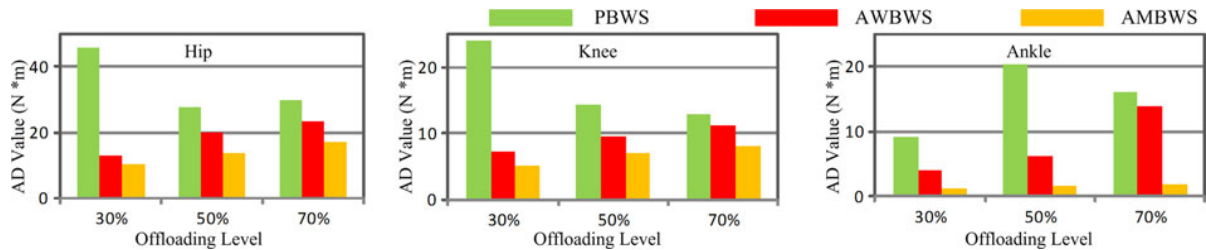


Fig. 6. Average difference values in hip, knee, and ankle joints at different offloading levels.

such a treadmill with the mass-offload BWS system will experience a feeling as if he/she did lose partial body mass from the perspective of the ground reaction force.

IV. EXPERIMENTAL VERIFICATION

Even for proof of principle, the feasibility of the proposed unique mass-offload BWS still needs to be experimentally investigated. A simplified nonhuman testbed was designed based on two basic requirements: 1) a nonhuman test of the proposed mass-offload BWS system is needed in the early development stage and 2) the testbed should behave reasonably close to its application case and can provide test data for a quantitative engineering assessment of the new technology.

A. Experimental System Design

A testbed was designed based on the goal of proving the engineering principle, namely, the mass-offload BWS system can virtually offload partial body mass. The conceptual and the physical testbed are shown in Figs. 7 and 8, respectively.

The testbed consists of two independent devices, namely, an active BWS system and an active cam-slider mechanism. As indicated in Fig. 7, the active BWS system has all the essential components of the proposed mass-offload BWS device including cables, a pulley, an electrical winch, a cable tension load cell, an encoder, an accelerometer, and the mass-offload BWS control system for driving the winch. Such an active BWS system

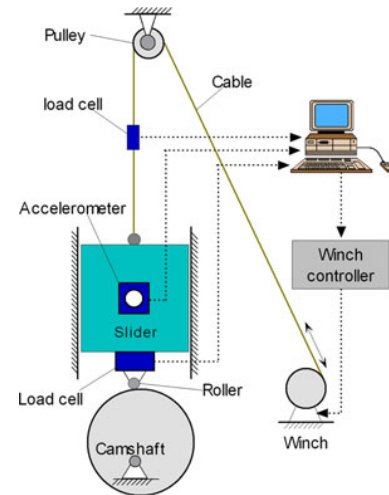


Fig. 7. Nonhuman testbed of the active mass-offload BWS.

can be easily turned into a counterweight-based passive BWS system simply by replacing the winch with a counterweight.

Also shown in Figs. 7 and 8, the active cam-slider mechanism consists of a dc motor, a cam, a weight-adjustable slider, two guiding rails, an encoder, a compression load cell, and a speed controller. Completely independent of the aforementioned active or passive BWS system, the cam-slider system is used to generate a motion pattern mimicking a human gait (in the vertical direction only) for the purpose of testing the new BWS

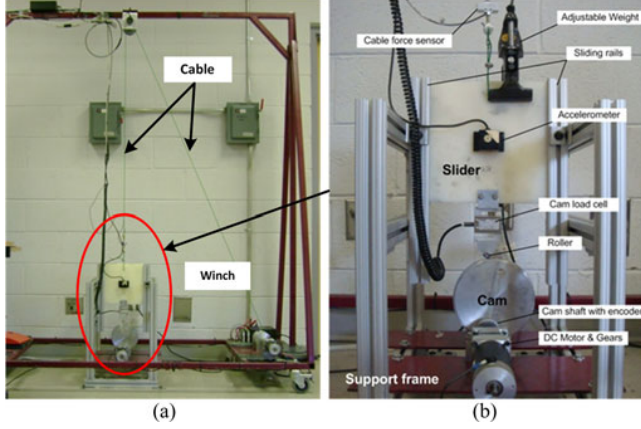


Fig. 8. Physical system of the nonhuman mass-offload BWS testbed. (a) Active winch-cable system. (b) Cam-slider based device.

technology. The sliding rails are used to mechanically constrain the slider to move vertically up and down without wobbling in the other directions. Although the sliding motion inevitably introduces friction forces on the sliding surfaces of the slider, no effort was taken in the active BWS controller to compensate the undesired friction. This was because 1) this sliding friction will not exist in a real BWS application case and 2) the friction showed up equally for all the comparison cases and, thus, would not affect the intended study. However, the incurred nonlinear friction forces in the experiment were precisely identified for the purpose of more accurate analysis of the system and the correct interpretation of the experimental data. Details of the nonlinear friction identification in our experimental work have been reported in [15].

During the experiment, the cam is controlled to rotate at a constant speed, such that it causes the slider to move up and down in a periodical motion pattern, which has equivalent magnitude and frequency of a normal human gait as measured at a point fixed to the human torso. This closely simulates the periodic body motion of a walking human in the vertical direction. The compression load cell is installed at the bottom of the slider to measure the contact force between the slider and the cam, which represents the ground reaction force acting on the walking subject who is supported by the BWS system. The accelerometer is installed at the center of the slider to measure the acceleration and the tension load cell is attached to the end of the cable to measure the cable tension. The acceleration data are feedback to the winch controller such that the appropriate cable tension is generated to achieve the function of mass offloading. The load cell on the slider is used only for the purpose of gathering the test data. However, the BWS system becomes passive when the winch is replaced with a set of counterweights for testing the performance of a passive BWS.

B. Dynamics of the Experimental System

From the perspective of dynamics, the BWS test system shown in Fig. 8 can be approximately modeled as depicted in Fig. 9(a), where m_1 , b_1 , and k_1 represent the mass, damping

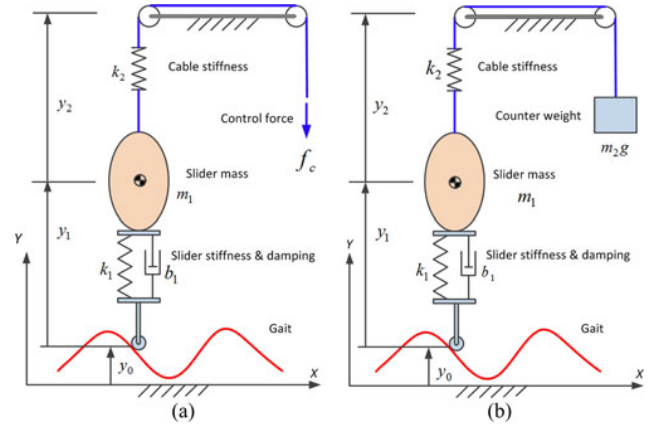


Fig. 9. Modeling of the active mass-offload and thpassive weight-offload BWS. (a) Active mass-offload BWS. (b) Passive weight-offload BWS.

coefficient, and stiffness of the slider (or the patient's body); y_1 is the vertical displacement of the mass center of the slider with respect to the contact point; y_0 stands for the vertical gait of the slider generated by the cam; and f_c is the control force produced by the active winch.

Assuming the vertical component of the reaction force exerted on the roller by the cam is f and the body acceleration in the vertical direction is a , the dynamics equation for the system can be written as

$$m_1(\ddot{y}_0 + \ddot{y}_1) + b_1\dot{y}_1 + k_1y_1 + m_1g = f + f_c. \quad (16)$$

If we apply a control force f_c to the system as

$$f_c = m_2(\ddot{y}_0 + \ddot{y}_1) + m_2g \quad (17)$$

where m_2 is the amount of mass to be virtually removed, the dynamics equation (16), thus, becomes

$$(m_1 - m_2)(\ddot{y}_0 + \ddot{y}_1) + b_1\dot{y}_1 + k_1y_1 + (m_1 - m_2)g = f. \quad (18)$$

Note that (18) is identical to the dynamics equation of the system with a reduced mass $(m_1 - m_2)$. This fact implies that a certain amount of mass m_2 is virtually removed from the slider by applying cable tension force defined in (17). On the other hand, when the slider is supported by a counterweight m_2g as shown in Fig. 9(b), which can be easily implemented by replacing the active winch with a counterweight, the dynamics equation of the system becomes

$$(m_1 + m_2)(\ddot{y}_0 + \ddot{y}_1) + b_1\dot{y}_1 + k_1y_1 + (m_1 - m_2)g = f. \quad (19)$$

One can easily find from (19) that, because an extra mass m_2 is added to the system, an extra inertia load $m_2(\ddot{y}_0 + \ddot{y}_1)$ is produced on the slider. The magnitude of this extra inertia force is coupled with the acceleration of the slider. This fact indicates that the intention to offload the gravitational load from the slider by m_2g without adding dynamic load just through a counterweight mechanism is unattainable.

Equation (17) defines the control force, which the mass-offload BWS controller must provide in order to make the system behave exactly as if an amount of mass m_2 were removed

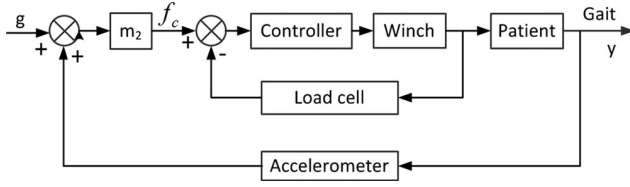


Fig. 10. Control strategy for offloading partial body mass.

TABLE II
PARAMETERS FOR ACTIVE BWS EXPERIMENTS

No	Slider mass m_1 (kg)	Mass to be offloaded m_2 (kg)	Cam velocity Hz
1	1.95	0.5	0.5
2	2.25	0.8	0.5
3	2.45	1.0	0.5

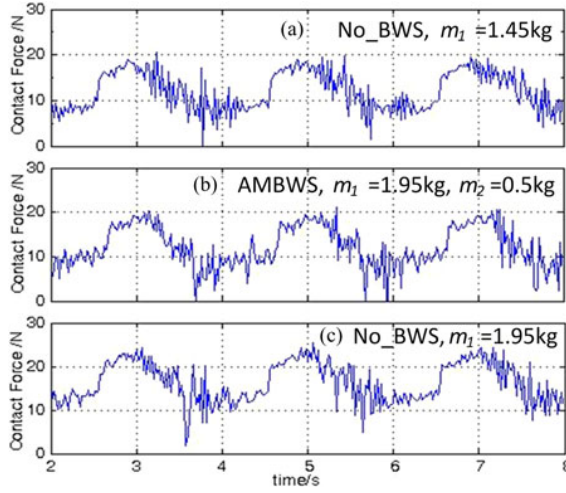


Fig. 11. Experimental results for Test 1.

from the slider. Note that the term $m_2\ddot{y}_1$ in (17) means that an acceleration feedback is needed in the cable control system. In implementation, a force-feedback loop is required in order to guarantee that the control goal (17) is achieved. An example of such a force control strategy is shown in Fig. 10. As described in Section II, no worry is needed in the control design for preventing the cable from becoming slack.

C. Experiment Parameters and Results

To demonstrate the mass offloading capability of the active BWS system, we show three comparative experiments on the aforementioned testbed. The parameters for these experiments were set to the values as listed in Table II. Note that the mass values are small comparing to human mass because of the relatively smaller size of the testbed. Nevertheless, this scale difference does not affect the outcome of the experiment.

First, we turned off the active tension control (i.e., the cable is physically disconnected from the slider). The slider with different masses of 1.45, 1.95, 2.25, and 2.45 kg were driven by

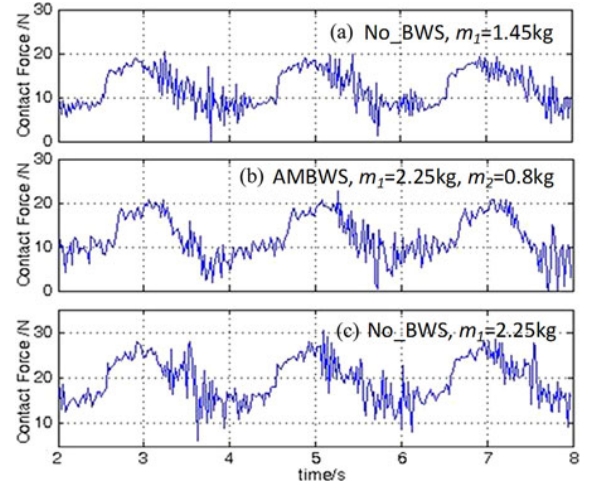


Fig. 12. Experimental results for Test 2.

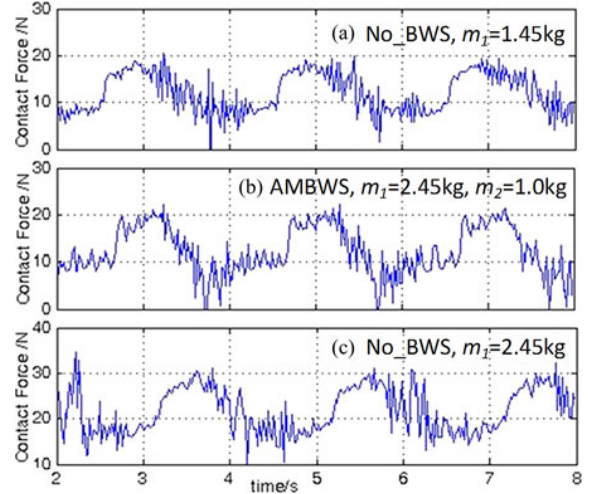


Fig. 13. Experimental results for Test 3.

the cam moving up and down with a frequency of 0.5 Hz. The contact force between the cam and the slider were measured and plotted in subplots (a) and subplots (c) of Figs. 11–13. It should be pointed out that the cam-slider contact forces for the 1.45-kg slider case are repeatedly plotted in each figures for easy reference (i.e., a No_BWS case with an actual mass of 1.45 kg).

Then, the slider was connected to the actively controlled cable to respectively reduce the mass by 0.5 kg from a 1.95-kg slider, 0.8 kg from a 2.25-kg slider, and 1.0 kg from a 2.45-kg slider, hoping to obtain nearly the same contact-force response as the 1.45-kg slider No_BWS case. The resulting contact forces are plotted in subplots (b) of Figs. 11–13. From these figures, one can easily find that subplot (b) resembles very well to subplot (a) in each of the three plots. This means that the mass-offload BWS system has effectively made the 1.95-kg, 2.25-kg, or 2.45-kg sliders to dynamically behave as if they all had the same mass of $m_1 - m_2 = 1.45$ kg.

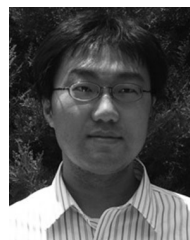
V. CONCLUSION

This paper presented a new active body weight support (BWS) method, which is a mass-offload BWS capable of virtually offloading partial body mass of an attached human. Offloading body mass can smoothly offload not only the corresponding body weight but also the related inertia force (or dynamic load) of the person during a treadmill-based activity. This unique feature is not available from the existing passive or active weight-offload BWS system and it has a potential of improving the BWS-assisted rehabilitation practice and other applications.

The mass-offload capability was studied and verified by both dynamics simulation and nonhuman experiment. In the simulation study, the mass-offload BWS system including the attached human body was modeled as a multibody dynamical system. In the experiment, a cam-slider system was designed to generate a human-like walking gait. All three types of BWS methods, namely, the passive, the weight-offload, and the proposed mass-offload BWS were studied to confirm the efficacy and advantages of the mass-offload BWS over the other BWS technologies in terms of the dynamic walking load. The study results confirmed that the weight-offload BWS methods result in larger dynamic load on the human body than the mass-offload BWS because of more inertia of the system in effect. Furthermore, a passive BWS system may also cause momentary loss of feet-ground contact when the counterweight is too large if the attached human wants to keep a normal walking gait. The proposed mass-offload BWS system does not have such a drawback because it offloads both the weight and the dynamic load through virtually removing partial body mass. Thus, it can make the attached human feel exactly as if he/she had a reduced body mass.

REFERENCES

- [1] M. Wessels, C. Lucas, I. Eriks, and S. de Groot, "Body weight-supported gait training for restoration of walking in people with an incomplete spinal cord injury: A systematic review," *J. Rehabil. Med.*, vol. 42, no. 6, pp. 513–519, 2010.
- [2] A. L. Hicks and K. A. M. Ginis, "Treadmill training after spinal cord injury: It's not just about the walking," *J. Rehabil. Res. Dev.*, vol. 45, no. 2, pp. 241–248, 2008.
- [3] B. H. Dobkin, "An overview of treadmill locomotor training with partial body weight support: A neurophysiologically sound approach whose time has come for randomized clinical trials," *Neurorehabil. Neural Repair*, vol. 13, no. 3, pp. 157–165, 1999.
- [4] K. P. Westlake and C. Patten, "Pilot study of Lokomat versus manual-assisted treadmill training for locomotor recovery post-stroke," *J. Neuroeng. Rehabil.*, vol. 6, 2009, doi: 10.1186/1743-0003-6-18.
- [5] T. Lam, K. Pahl, A. Krassioukov, and J. J. Eng, "Using robot-applied resistance to augment body-weight-supported treadmill training in an individual with incomplete spinal cord injury," *Phys. Therapy*, vol. 91, no. 1, pp. 143–151, 2011.
- [6] H. Barbeau and M. Visintin, "Optimal outcomes obtained with body-weight support combined with treadmill training in stroke subjects," *Arch. Phys. Med. Rehabil.*, vol. 84, no. 10, pp. 1458–1465, 2003.
- [7] A. C. Lo and E. W. Triche, "Improving gait in multiple sclerosis using robot-assisted, body weight supported treadmill training," *Neurorehabil. Neural Repair*, vol. 22, no. 6, pp. 661–671, 2008.
- [8] J. Hilder, D. Nichols, M. Pelliccio, K. Brady, D. D. Campbell, J. H. Kahn, and T. G. Hornby, "Multicenter randomized clinical trial evaluating the effectiveness of the lokomat in subacute stroke," *Neurorehabil. Neural Repair*, vol. 23, no. 1, pp. 5–13, 2009.
- [9] Y. Shibata, S. Imai, T. Nobutomo, T. Miyoshi, and S. Yamamoto, "Development of body weight support gait training system using antagonistic bi-articular muscle model," in *Proc. 32nd Annu. Int. Conf. IEEE Eng. Med. Biol. Soc.*, Buenos Aires, Argentina, Aug. 31–Sep. 4, 2010, pp. 4468–4471.
- [10] J. Yoon, B. Novandy, C. H. Yoon, and K. J. Park, "A 6-DOF gait rehabilitation robot with upper and lower limb connections that allows walking velocity updates on various terrains," *IEEE/ASME Trans. Mechatronics*, vol. 15, no. 2, pp. 201–215, Apr. 2010.
- [11] M. Frey, G. Colombo, M. Vaglio, R. Bucher, M. Jorg, and R. Riener, "A novel mechatronic body weight support system," *IEEE Trans. Neural Syst. Rehabil.*, vol. 14, no. 3, pp. 311–321, Sep. 2006.
- [12] M. Glauser, Z. L. Lin, and P. E. Allaire, "Modeling and control of a partial body weight support system: An output regulation approach," *IEEE Trans. Control Syst. Technol.*, vol. 18, no. 2, pp. 480–490, Mar. 2010.
- [13] F. Gazzani, A. Fadda, M. Torre, and V. Macellari, "WARD: A pneumatic system for body weight relief in gait rehabilitation," *IEEE Trans. Rehabil. Eng.*, vol. 8, no. 4, pp. 506–513, Dec. 2000.
- [14] K. Gordon, B. Svendsen, S. J. Harkema, and S. El-Alami, "Closed-loop force controlled body weight support system," U.S. Patent 7 381 163, Jun. 3, 2008.
- [15] J. Liang, A. Hernandez, O. Ma, B. Qiao, and R. Paz, "Nonhuman test of an active body support system for improving locomotion training," in *Proc. IEEE Int. Conf. Adv. Intell. Mechatronics*, Singapore, Jul. 14–17, 2009, pp. 1312–1317.
- [16] O. Ma, X. Diao, L. Martinez, and T. Sarkodie-Gyan, "Dynamically removing partial body mass using acceleration feedback for neural training," in *Proc. IEEE 10th Int. Conf. Rehabil. Robot.*, Noordwijk, the Netherlands, Jun. 13–15, 2007, pp. 1102–1107.
- [17] C. P. McGowan, R. R. Neptune, and R. Kram, "Independent effects of weight and mass on plantar flexor activity during walking: implications for their contributions to body support and forward propulsion," *J. Appl. Physiol.*, vol. 105, no. 2, pp. 486–494, 2008.
- [18] N. A. Tanner and G. Niemeyer, "High-frequency acceleration feedback in wave variable telerobotics," *IEEE/ASME Trans. Mechatronics*, vol. 11, no. 2, pp. 119–127, Apr. 2006.
- [19] M. W. Whittle, *Gait Analysis: An Introduction*. St. Louis, MO: Heidi Harrison Elsevier, 2007.
- [20] X. Diao and O. Ma, "Force-closure analysis of 6-DOF cable manipulators with seven or more cables," *Robotica*, vol. 27, pp. 209–215, 2009.
- [21] S. K. Banala, S. K. Agrawal, S. H. Kim, and J. P. Scholz, "Novel gait adaptation and neuromotor training results using an active leg exoskeleton," *IEEE/ASME Trans. Mechatronics*, vol. 15, no. 2, pp. 216–225, Apr. 2010.
- [22] L. Marchal-Crespo and D. J. Reinkensmeyer, "Review of control strategies for robotic movement training after neurologic injury," *J. Neuroeng. Rehabil.*, vol. 6, 2009.
- [23] R. N. Marshall, R. K. Jensen, and G. A. Wood, "A general newtonian simulation of an N-segment open-chain model," *J. Biomech.*, vol. 18, no. 5, pp. 359–367, 1985.
- [24] K. Hase and N. Yamazaki, "Computer simulation study of human locomotion with a three-dimensional entire-body neuro-musculo-skeletal model—Part I: Acquisition of normal walking," *JSME Int. J. C, Mech. Syst.*, vol. 45, no. 4, pp. 1040–1050, 2002.
- [25] M. Stelzer and O. von Stryk, "Efficient forward dynamics simulation and optimization of human body dynamics," *Zamm-Z. Angew. Math. Mech.*, vol. 86, no. 10, pp. 828–840, 2006.
- [26] T. Spagele, A. Kistner, and A. Gollhofer, "Modelling, simulation and optimisation of a human vertical jump," *J. Biomech.*, vol. 32, no. 5, pp. 521–530, 1999.
- [27] LifeModeler. LifeMod. [Online]. Available: www.lifemodeler.com
- [28] AnyBody_Technology. *The AnyBody Modeling System*. [Online]. Available: <http://www.anybodytech.com>



Qi Lu received the B.S. and the M.S. degrees in mechanical engineering from Shanghai University, Shanghai, China, in 2004 and 2007, respectively. Currently, he is working toward the Ph.D. degree in the Department of Mechanical and Aerospace Engineering, New Mexico State University, Las Cruces.

His current research interests include multibody dynamics, robotics, human body dynamics, rehabilitation, etc.



Jianxun Liang received the B.E. and the M.Sc. degrees in textile materials and design from Donghua University, Shanghai, China, in 1999 and 2002, respectively. Currently, he is working toward the Ph.D. degree in the Department of Mechanical and Aerospace Engineering, New Mexico State University, Las Cruces.

His research interests include impact-contact dynamics modeling, simulation, and experimental verification. He is also working on the design and control of mechanical systems.



Bing Qiao received the M.Eng. and the Ph.D. degrees in mechatronic engineering from Nanjing University of Aeronautics and Astronautics (NUAA), Nanjing, China, in 1994 and 1999, respectively.

From 1994 to 2002, he was a Lecturer and the Director of the CAD Centre at the College of Aerospace Engineering, NUAA. During 2002–2006, he directed the Department of Civil Aviation Management, College of Civil Aviation, NUAA. He is currently an Associate Professor in the Department of Astronautic Control, College of Astronautics, NUAA. His current

research interests include intelligent robotics, active force control, and visual servo control.



Ou Ma received the M.Eng and the Ph.D. degrees in 1987 and 1991, respectively, from McGill University, Montreal, QC, Canada.

He is currently a Professor in the Department of Mechanical and Aerospace Engineering, New Mexico State University (NMSU), Las Cruces, and also directs the Reduced-Gravity and Biomechanics Laboratory of the University. His research interests include multibody dynamics and control for the applications of space and rehabilitation robotics. Prior to joining NMSU in 2002, he had been employed by

MDA Space Missions for nearly 11 years, working as a Project Engineer and R&D Technical Leader for developing modeling, simulation, and ground-based experimental verification technologies for the space station robots Space Station Remote Manipulator System (SSRMS) and Special Purpose Dexterous Manipulator (SPDM) as well as other robotic systems for satellite on-orbit servicing missions.