GRAVITY-ASSIST: A Series Elastic Body Weight Support System with Inertia Compensation

Hammad Munawar Volkan Patoglu

***Abstract*— We present GRAVITY-ASSIST, a series elastic active body weight support and inertia compensation system for use in robot assisted gait rehabilitation. The device consists of a single degree of freedom series elastic actuator that connects to the trunk of a patient. The series elastic system is novel in that, it can provide the desired level of dynamic unloading such that the patient experiences only a percentage of his/her weight and inertia. Inertia compensation is important, since the inertial forces can cause significant deviations from the desired unloading force, specially at low support forces and fast walking speeds. Furthermore, this feature enables the inertia of the harness and force sensing unit attached to the patient to be compensated for, making sure that the device does not interfere with the natural gait cycle. We present a functional prototype of the device, its characterization and experimental verification of the approach.**

I. INTRODUCTION

The ability to walk is one of the most important functions of the human body and walking disabilities negatively affect the quality of life [1], [2]. A disability in walking can be caused by a number of factors, while the restoration of proper gait always requires physical therapy. Gait rehabilitation therapy is known to be most effective when it is intense and repetitive, which makes robot assisted devices well-suited for this job. Consequently, robot assisted gait rehabilitation is an active area of research and new techniques are continuously being explored in order to improve efficacy of such therapies.

Body weight support (BWS) systems are an indispensable component of robot assisted gait rehabilitation. It has been clinically shown that gait rehabilitation can be made significantly more effective, when a percentage of patients' weight is supported by a BWS system, as compared to the case when patients have to bear their full weight [3]–[5]. Furthermore, BWS is essential during robot assisted therapies, to ensure safety and to prevent falls.

In this paper, we present a series elastic active weight support and inertia compensation system for use in gait rehabilitation. The system is capable of providing dynamic weight support to patients while walking. In addition, it can provide compensation for the inertial forces caused by the vertical movements of the human body. Compensation of inertial forces has been largely ignored in the literature, even though these forces can cause significant deviations in the unloading force, specially when the support force is low and walking speed is fast. Furthermore, without such compensation, the added inertia due to the device may interfere with natural gait. GRAVITY-ASSIST aims to overcome these limitations by actively compensating for inertial forces using online measurements of the vertical accelerations. Furthermore, thanks to series elastic actuation, robust force control and low output impedance can be achieved with a relatively low-cost device.

A. Body Weight Support in Gait Rehabilitation

Patients with walking disabilities are often unable to support their own weight, due to muscle weakness or paralysis. Consequently, an effective gait rehabilitation system must be capable of fully/partially supporting the weight of the patient. Such systems can help to reduce the force that patients encounter on their legs during walking.

Efficacy of gait rehabilitation therapy while supporting the weight of the patient has been explored by many groups. Experimental results indicate that gait rehabilitation is more effective when the body weight of the patient is partially supported [3]–[5]. It has been documented that a weight support of no more than 30% of the body weight results in best performance, as high levels of weight support can decrease the activity of muscles in stroke patients. Furthermore, effective training is known to require gradual reduction of the weight support as patients start supporting their own body weight [6]. Lateral balance of patients has also been shown to improve, when BWS is used in gait rehabilitation [7]. Furthermore, BWS is essentially required to ensure safety and to prevent falls during walking. This feature is especially important for overground gait rehabilitation devices (as compared to treadmill based devices) where the risk of falling is greater [8], [9].

BWS systems are found in many gait rehabilitation devices to unload the patient weight during walking and to prevent falls. With respect to weight unloading, these systems can be categorized into i) static systems, ii) passive counterweight based systems, iii) passive elastic spring based systems, and iv) active dynamic systems. Figure 1 presents a schematic representation of these categories [9].

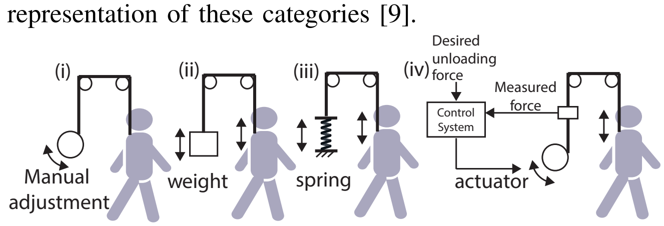

Fig. 1. Types of BWS systems (i) Static, (ii) passive counter weight based (iii) passive spring based, and (iv) active systems. (reproduced from [9])

Static systems consist of a mechanism that can be set

This work has been partially supported by TUBITAK Grant 115M698.

H. Munawar and V. Patoglu are with the Faculty of Engineering and Natural Sciences at Sabancı University, Istanbul, Turkey. {hammadmunawar,vpatoglu}@sabanciuniv.edu

to unload a predefined amount of weight, when the patient is connected to the support system [10]. However, these systems do not ensure equal amount of weight support as the body center of mass (CoM) moves vertically during walking. Especially, when the body moves downwards the harness becomes tighter, restricting the movements of the pelvis. This makes static systems unsuitable for natural walking and make them less effective for gait rehabilitation. *Passive counterweight based* systems use a counterweight to dynamically unload a percentage of the patient's body weight [11]. The counterweight is set to a predefined value and moves vertically as the patients walks, maintaining a constant static unloading force. However, the movement of the counterweight also results in additional inertial forces, which can cause large fluctuations in the support forces, unless actively compensated. *Passive elastic* systems use elastic elements to provide unloading forces based on the tension in these elastic elements [12]. This is advantageous with respect to the counterweight based approach, since these systems do not induce extra inertial forces; however, passive elastic systems cannot ensure that the unloading forces stay constant, as the amount of support varies with the length of the elastic element. Consequently, all of these passive systems are not very effective for body weight support, as the patient does not feel a constant weight unloading during walking, interfering with the natural gait and potentially negatively affecting the efficacy of the therapy.

Active dynamic systems are capable of generating unloading forces dynamically [9]. In particular, these type of systems continually measure the interaction force between the patient and the BWS actuator and based on these measurements, a control system commands the actuator to move in such way that a constant amount of vertical force is felt by the patient, despite the vertical movements of the patient during walking. With a high enough control bandwidth, these systems can provide comfortable weight unloading to promote natural walking [8], [13].

A number of strategies have been employed for connecting the patient to a BWS, including utilization of an overhead system of pulleys or direct connection to the trunk, waist, hip or pelvis of the patient. The overhead pulleys is the most commonly employed method; however, this arrangement tends to produce horizontal forces when patients are not directly under the support system. These forces may pull patients towards the center and introduce balance problems. In particular, the index of lateral stability becomes low with these systems, when the supporting weight is high [14]. Even though overhead pulleys are commonly used and easy to implement, other methods of connection may be preferable to ensure better balance during gait training.

B. Inertia Compensation

When a patient is attached to a BWS system, the human body acts as an inertial load. The human weight exerts a static force on the BWS, while the vertical movements of the body during walking cause inertial forces which are proportional to its acceleration. Considering the human mass and natural walking speeds, these inertial forces can become significantly large and hinder the operation of the BWS, by causing large deviations in the interaction force from the desired level unloading. Furthermore, the added inertia of BWS may cause a decrease in the natural frequency of motion, hindering achievement of natural gait [7], [15].

A more natural unloading strategy for gait rehabilitation is to compensate, not only for the partial weight of the patient, but also for the corresponding inertial forces due the compensated mass and the BWS system, such that the patient feels as if his/her total mass has been reduced by a predetermined amount even under dynamic movements. This strategy avoids an improper ratio between the weight and inertial forces rendered to the patient and has the potential to promote natural gait. For natural motion, the percentage of inertia compensation is set to the same level as the percentage of weight unload.

While the compensation for patient/device weight and parasitic effects, such as friction and stiction, can be robustly achieved through a force controller, the inertia compensation is more challenging due to the stability issues it presents when used in a feedback control loop [7], [15]. To alleviate these stability problems, commonly employed inertia compensation approaches rely on an extra sensor to detect the instantaneous accelerations of the compensated inertia and a model-based feed-forward compensator. In particular, after presenting the coupled stability limitations of closed-loop inertia compensation approaches, [15] proposes an *emulated inertia compensation* scheme that utilizes the filtered acceleration measurements to approximately compensate for the limb and exoskeleton inertia for a lower-extremity exoskeleton. This study also provides evidence that such an inertia compensation scheme can improve the natural frequency of lower-limb swing movements.

C. Force Control and Series Elastic Actuation

Force control is one of the crucial components of an active BWS. In the literature, stiff commercial force sensors are commonly employed in admittance control schemes. Due to the non-collocation between the force sensor and the actuator, when stiff force sensors are used, the maximum gain that the force controller can utilize is severely limited by stability constraints. Low controller gains are undesirable, since they limit the robustness of the controller and the accuracy of force control. Furthermore, with stiff force sensors, any sudden impacts from the environment are directly passed to the controller with potentially damaging effects.

Series Elastic Actuation (SEA) is a technique proposed to alleviate these issues by deliberately introducing a compliant element between the actuator and the environment. This compliant element is orders of magnitude less stiff than a force sensor; hence, SEA allows for higher controller gains to be employed for robust and accurate force control. Force is determined by measuring the deflection of the compliant element, while force control can be implemented through robust motion control of the compliant element. Mechanically, SEA possesses a low output impedance at frequencies higher than its control bandwidth, enabling these systems to absorb any sudden impacts from the environment. In addition to low output impedance, SEA features active backdriveability within its control bandwidth, which makes this technique suitable for applications requiring physical human-robot interaction. SEA can be implemented at low costs, since neither high quality drive trains/power transmission elements, nor expensive force sensors/signal conditioners are required for these systems. A potential disadvantage of SEA is its limited force control bandwidth; however, bandwidth limitation is not of high concern for rehabilitation robotics where the patient movements are relatively slow.

II. Related Work

Robot assisted devices available for gait rehabilitation feature some form of BWS system. HapticWalker [16] is an end-effector type device that utilizes a passive trunk suspension module to unload patients while walking. The BWS system keeps the absolute position of the human CoM constant by controlling the trajectory of the foot plates. MIT Skywalker [17] is another end-effector type device that features a loose chest harness to prevent patient falls and a passive spring based BWS system that uses a bicycle seat to unload patient weight. The position of the seat is pre-set by moving a linear actuator up or down using a remote control to allow for the desired level of weight support.

The gait rehabilitation system comprising of POGO and PAM [18] uses the commercial Robomedica active BWS system that connects to the patient through an overhead harness and controls the tension of the overhead cable. Lokomat [9] utilizes Lokolift active dynamic BWS system. The patient is connected to an overhead harness that connects to a spring through pulleys. The unloading amount is measured by a force sensor and a motor adjusts the spring length to compensate for the desired level of patient weight. WalkTrainer [19] is an overground gait rehabilitation device with an active BWS system that consists of a controlled preloaded spring whose length is adjusted according to force sensor measurements. This system also connects to the patient through an overhead harness. KineAssist [20] gait and balance training system supports the body weight using an active custom designed harness that connects at the pelvis of the patient The unloading force is measured using load cells embedded in the harness and desired level of unloading is implemented using a force controller. NaTUre-gaits [21] is a hybrid device that implements end-effector (foot plate) type technology with a mobile base to deliver control of foot movements and an overground walking experience. It features active BWS at its pelvis module that measures the interaction forces using force sensors and actively generates the desired amount of weight unloading by creating a virtual spring along the pelvic trajectory. All of the active BWS systems presented above rely on stiff commercial load cells for closed loop force control and neither of them features an inertial compensation scheme.

ZeroG [13] is an overground BWS system that moves on overhead rails and implements a force control strategy based on SEA. In particular, this system utilizes SEA to measure and actively control the tension in the rope that connects to the patient harness. This system does not implement an inertia compensation strategy, instead considers inertial forces due to the device as disturbances and relies on a PI force controller in an attempt to overcome the force deviations caused by these forces. ZeroG does not allow for lateral movements; thus, can cause balance problems during overground walking [14], [22]. FLOAT [22] is a multi degree of freedom version of ZeroG that consists of two overhead rails connected to the patient harness at four points so that it can allow patients to move laterally in addition to the forward/backward movements of ZeroG. Similar to ZeroG, FLOAT does not compensate for inertia of the human body while moving.

Compensation for inertial forces due to patient has been mostly ignored in BWS systems available for gait rehabilitation. However, considering that the amplitude of vertical CoM displacements is about 49.8 ± 10.3 mm at natural gait cycles [23], the inertial forces can account up to 25% of the gravitational loads.

Similar to ZeroG and FLOAT, Gravity-Assist is a series elastic active BWS system that can be used to unload patients' weight during gait rehabilitation. Unlike ZeroG and FLOAT, Gravity-Assist actively compensates for the inertial forces utilizing the emulated inertia compensation scheme [15]. Furthermore, the device connects to the trunk to feature improved lateral stability during walking.

III. Gravity-Assist

Gravity-Assist is designed to satisfy the following task specifications. The active BWS system should

- dynamically compensate for inertial forces of the patient's body, in addition to unloading of patient's weight,
- not interfere with patient balance,
- ensure safety against falls,
- allow for unrestricted pelvic movements,
- provide an ergonomic and comfortable support,
- be able to lift patients from a sitting position to initiate therapies, and
- enable low cost implementations.

A. Design

To satisfy the task specifications, Gravity-Assist features a single DoF series elastic BWS system that attaches to the trunk of a patient. A connection to the trunk is preferred as it does not interfere with pelvic movements or the balance of patients. SEA ensures robust, high fidelity control of interaction forces, while simultaneously ensuring active backdriveability and a low output impedance. SEA also enables low cost implementations of the system, through use of low cost power transmission elements, robust motion controllers and position sensors. The system features appropriate stroke and output force to actively assist patients during sit-to-stand tasks. Finally, the bandwidth of the SEA is designed to be significantly higher than the 1 Hz bandwidth of natural human walking [24].

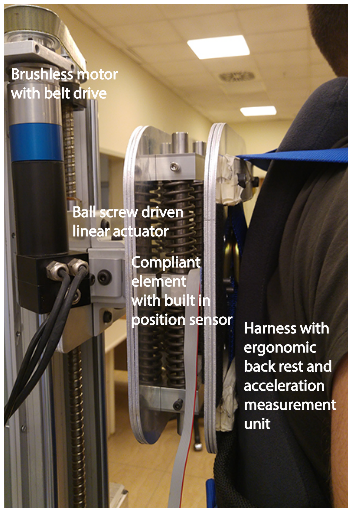

Fig. 2. GRAVITY-ASSIST attached to a volunteer

GRAVITY-ASSIST consists of three modules: series elastic element, motion controlled linear actuator, and harness and acceleration measurement unit as shown in Figure 2.

1) Linear Actuator: The linear actuator consists of a low friction ball screw mechanism, whose screw is actuated by a velocity controlled brushless DC motor connected through a belt drive. The actuator is selected such it can continuously support the weight of a patient and has a stroke of 1000 mm to enable lifting the patient from a sitting position. The velocity controller is implemented on the motor driver hardware.

2) Series Elastic Element: The series elastic element consists of linear compression springs sandwiched between two aluminum plates, one attached to the end-effector of the linear actuator and the other attached to the harness. The deflection of the springs are measured by an optical encoder that possesses 8000 count per inch resolution under quadrature decoding. Given the stiffness of the spring, the interaction force between the two plates can be estimated.

3) Harness and Acceleration Measurement Unit: The patient connects to the series elastic element through a harness that is fitted with an ergonomic back support for proper posture and comfort. Vertical force exerted by the human body is transmitted through the harness to the series elastic element, where it is measured as spring deflection. Passive movements of the harness within limited ranges are enabled (except along the vertical direction) to promote patient comfort.

The acceleration measurement unit consists of a three axes low-g MEMS accelerometer with an internal sampling frequency of 11 kHz. The accelerometer is attached to the patient using a chest strap. The acceleration measurements are low pass filtered.

B. Control

A real-time cascaded controller is implemented for SEA as shown in Figure 3. The cascaded controller consists of an inner robust velocity control loop and an outer force control loop. The inner loop of the control structure deals with imperfections, such as friction and stiction of the linear actuator, rendering the system into an ideal velocity source. The velocity controller is implemented in hardware on the motor driver with fast control rate of 20 kHz. The outer force control loop is implemented at 1 kHz for high fidelity force control. This loop imposes the desired level of support forces as determined by the therapist. The cascaded force controller [25] is augmented with emulated inertia compensation scheme as proposed in [15]. In particular, inertial forces to be compensated for are estimated based on low-pass filtered real-time acceleration measurements and the pre-determined mass of the device and the patient. Then, these inertial force estimates are provided to the force controller as a reference, in addition to the force reference used for weight unloading based a predetermined percentage ($\lambda\%$) of the patient weight set according to the therapy requirements.

For safety, in addition to emergency stops, limits on the maximum speed and minimum position of the linear actuator have been implemented. In case the patient falls or is unable to support his/her own weight, the linear actuator holds the patient and slowly moves down to a comfortable height, where a stool can be placed under the patient. The system also allows for connecting of patients at a sitting position and patients to be slowly raised to a standing position.

C. Experimental Characterization

GRAVITY-ASSIST has been experimentally characterized and the technical specifications of the prototype are summarized in Table I.

TABLE I
TECHNICAL SPECIFICATIONS OF GRAVITY-ASSIST

Parameter	Value
Stroke	1 m
Output force	600 N
Force sensing resolution	0.3 N
Velocity control bandwidth	12 Hz
Force control bandwidth	9.5 Hz (20 N – low force) 6.5 Hz (250 N – moderate force) 4.0 Hz (500 N – high force)

Figure 4 presents the small, moderate and large force control bandwidths of the device. With 4 Hz large force control bandwidth, the device is evaluated to be significantly faster than natural human gait cycle taking place at 1 Hz.

Figure 5 depicts force tracking performance of the device for a chirp force reference input, for which an RMS error of 1.34% can be reported for frequencies up to 5 Hz.

IV. EVALUATION OF INERTIA COMPENSATION

We have experimentally evaluated the effect of inertia compensation on the assistance provided. For the trials, we mounted GRAVITY-ASSIST on a treadmill. A 26 years-old healthy volunteer was connected to the device. The volunteer signed an informed consent form approved by the IRB of Sabanci University before taking part in the experiments.

The experimental protocol consisted of attaching the device to the patient at a sitting position, raising him to a standing posture and asking him to walk forward at a natural pace. During all trials, the control system was set to actively support 50% of the volunteer's 64 kg weight. Three conditions have been tested for inertia compensation:

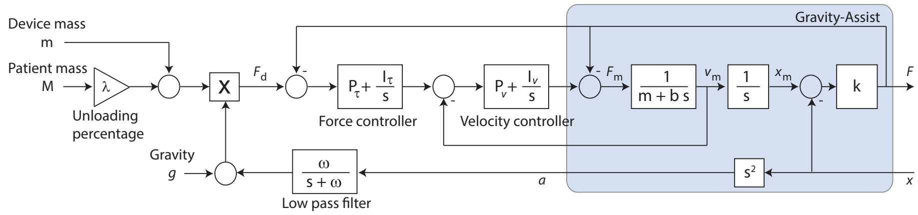

Fig. 3. Cascaded force controller with emulated inertia compensation

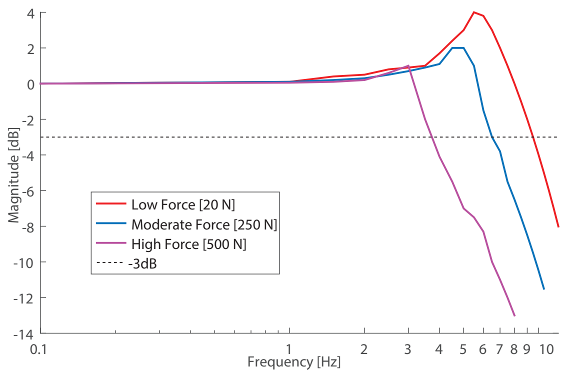

Fig. 4. Experimental characterization of the small, moderate and large force control bandwidths of GRAVITY-ASSIST

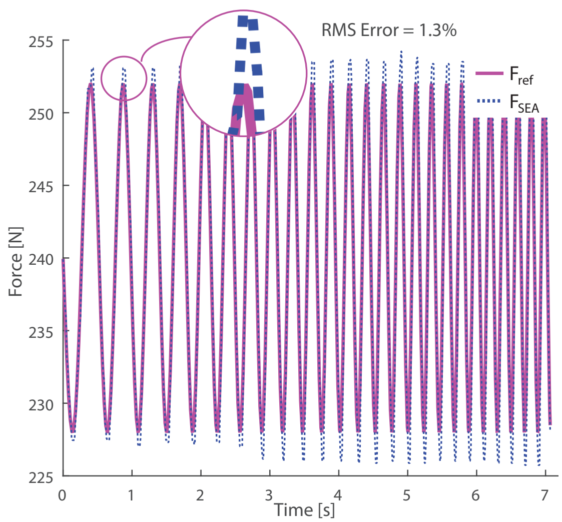

Fig. 5. Force tracking performance of GRAVITY-ASSIST for a chirp force reference input

0% inertia compensation (simple gravity compensation case), 50% inertia compensation (to match the percentage of weight compensation) and 100% inertia compensation. After each experiment, the volunteer was also asked to report the comfort level of the assisted gait. The order of trials was randomized. The vertical acceleration of the volunteer was measured using the accelerometer attached to his chest and a cutoff frequency 3 Hz was used for low-pass filtering these measurements.

The upper row of Figure 6 presents representative plots depicting reference assistance forces and measured interaction forces between the device and the volunteer, for weight unloading with and without inertia compensation. In particular, the inertia compensation is turned off in Figure 6(a), where a constant force is rendered by the device to compensate for 50% of the volunteer's weight. In this plot, one can observe relatively large deviations of the measured interaction force from the reference due to inertial force contributions, which are as large as 15% of the assistance forces. In Figure 6(b) and (c), 50% and 100% of the inertial forces are compensated for by the controller. As expected, the difference between the reference assistance forces and the measured interaction forces decreases as higher percentage of inertial forces are compensated, since the main disturbance acting on the force control system consists of the unaccounted inertial forces.

The lower row of Figure 6 presents the interaction forces between the ground and the volunteer. In Figure 6(e), the ground interaction forces include the inertial forces due the mass of the volunteer, while in Figure 6(f) the ground interaction forces are kept close to the constant value of uncompensated weight, as almost all of the inertial forces are successfully compensated for. The 50% compensation case is presented in Figure 6(f), where the ground interaction forces closely match the calculated interaction forces for a fictitious agent with 50% of the volunteer's mass. This case is interesting, since ensuring a meaningful ratio between the weight and inertial forces may promote patients to achieve a more natural gait.

Interview with the volunteer after the trials indicates that weight support without inertial compensation results in a gait that feels unbalanced and uncomfortable, the case with full inertia compensation lacks the dynamics of the gait, while the 50% compensation feels relatively more natural.

V. CONCLUSIONS

A series elastic BWS and inertia compensation system has been presented and experimentally characterized. The inertia compensation has been performed in a feed-forward manner based on online acceleration measurements taken from the trunk of the patient, while a cascaded force-motion controller has been used for force control for SEA. Initial experiments performed on a healthy volunteer indicate that deviations from desired interaction forces can be significantly reduced when inertia compensation is utilized. Furthermore, the volunteer states that the gait feels more natural when the inertia compensation matches the weight unloading.

Future works include further evaluation of the effects of inertia compensation on kinematics/dynamics of walking and gait. Integration of GRAVITY-ASSIST to an overground trainer [26] and studies with stroke patients are also planned.

REFERENCES

[1] J. Guralnik, L. Ferrucci, E. Simonsick, M. Salive, and R. Wallace, "Lower-extremity function in persons over the age of 70 years as a predictor of subsequent disability," *New England J. of Med.*, vol. 332, pp. 556–562, 1995.

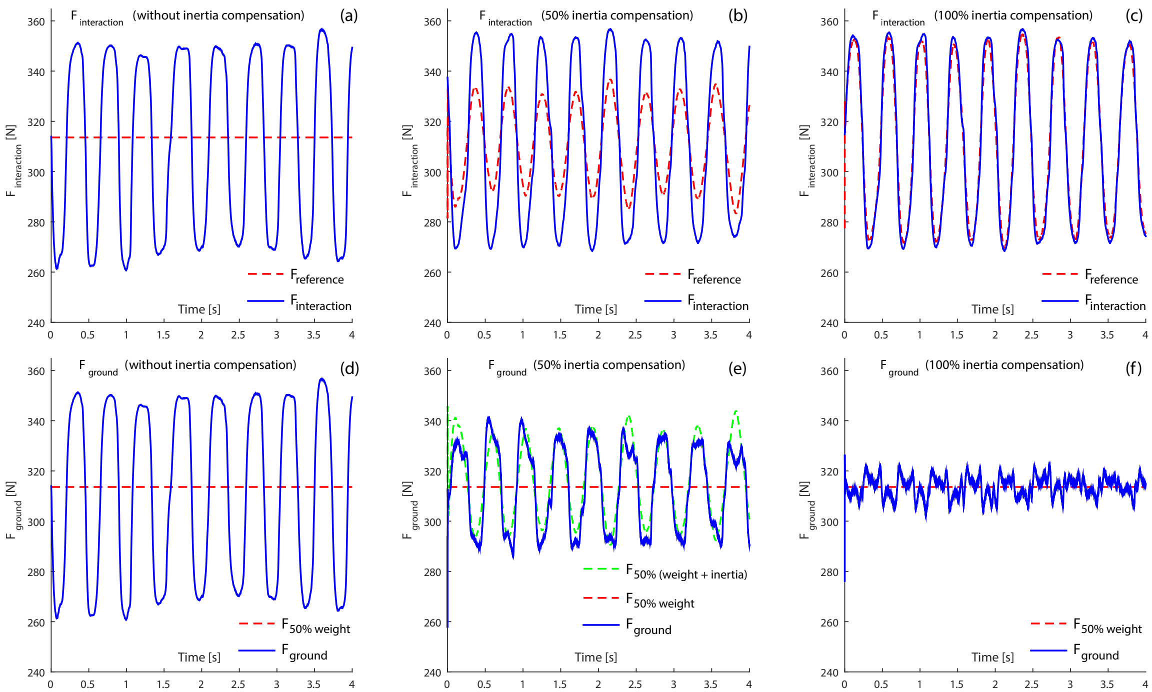

Fig. 6. Forces between volunteer and device (a) without inertia compensation, (b) with 50% inertia compensation, and (b) with 100% inertia compensation. Forces between volunteer and ground (d) without inertia compensation, (e) with 50% inertia compensation, and (f) with 100% inertia compensation.

[2] H. Chiu, J. Chern, H. Shi, S. Chen, and J. Chang, "Physical functioning and health-related quality of life: Before and after total hip replacement," *Kaohsiung J. of Med. Sci.*, vol. 16, no. 6, pp. 285–292, 2000.

[3] H. Barbeau and M. Visintin, "Optimal outcomes obtained with body-weight support combined with treadmill training in stroke subjects," *Arch. of Phy. Med. and Rehab.*, vol. 84, no. 10, pp. 1458–1465, 2003.

[4] J. Mehrholz, M. Pohl, and B. Elsner, "Treadmill training and body weight support for walking after stroke," *The Cochrane Library*, 2014.

[5] Y.-R. Mao, W. L. Lo, Q. Lin, L. Li, X. Xiao, P. Raghavan, and D.-F. Huang, "The effect of body weight support treadmill training on gait recovery, proximal lower limb motor pattern, and balance in patients with subacute stroke," *BioMed Research International*, 2015.

[6] S. Hesse, M. Konrad, and D. Uhlenbrock, "Treadmill walking with partial body weight support versus floor walking in hemiparetic subjects," *Archives of physical medicine and rehabilitation*, vol. 80, no. 4, pp. 421–427, 1999.

[7] A. Pennycott, D. Wyss, H. Vallery, and R. Riener, "Effects of added inertia and body weight support on lateral balance control during walking," in *IEEE Int. Conf. on Rehab. Robotics*, 2011, pp. 1–5.

[8] K. Gordon, D. Ferris, M. Roberton, J. Beres, and S. Harkema, "The importance of using an appropriate body weight support system in locomotor training," in *Proc. Soc. for Neurosci.*, vol. 26, 2000, p. 160.

[9] M. Frey, G. Colombo, M. Vaglio, R. Bucher, M. Jorg, and R. Riener, "A Novel Mechatronic Body Weight Support System," *IEEE Trans. on Neural Systems and Rehab. Eng.*, vol. 14, no. 3, pp. 311–321, 2006.

[10] H. Barbeau, M. Wainberg, and L. Finch, "Description and application of a system for locomotor rehabilitation," *Medical and Biological Engineering and Computing*, vol. 25, no. 3, pp. 341–344, 1987.

[11] F. Gazzani, A. Fadda, M. Torre, and V. Macellari, "Ward: a pneumatic system for body weight relief in gait rehabilitation," *IEEE Trans. on Rehab. Eng.*, vol. 8, no. 4, pp. 506–513, 2000.

[12] G. Chen, D. Schwandt, H. Van der Loos, J. Anderson, D. Ferris, F. Zajac, S. Kautz, C. Burgar, C. Patten, R. Neptune *et al.*, "Compliance-adjustable, force-sensing harness support for studying treadmill training in neurologically impaired subjects," in *Annual Gait and Clinical Movement Analysis Meeting*, 2001, pp. 25–28.

[13] J. Hidler, D. Brennan, I. Black, D. Nichols, K. Brady, and T. Nef, "Zerog: Overground gait and balance training system," *J. of Rehab. Research & Development*, vol. 48, no. 4, p. 287, 2011.

[14] A. Pennycott, H. Vallery, D. Wyss, M. Spindler, A. Dewarrat, and R. Riener, "A novel body weight support system extension: Initial concept and simulation study," in *IEEE Int. Conf. on Rehab. Robotics*, 2013, pp. 1–4.

[15] G. Aguirre-Ollinger, J. E. Colgate, M. A. Peshkin, and A. Goswami, "Design of an active one-degree-of-freedom lower-limb exoskeleton with inertia compensation," *Int. Journal of Robotics Research*, vol. 30, no. 4, pp. 486–499, 2011.

[16] S. Hussein, H. Schmidt, S. Hesse, and J. Kruger, "Effect of different training modes on ground reaction forces during robot assisted floor walking and stair climbing," in *IEEE Int. Conf. on Rehab. Robotics*, 2009, pp. 845–850.

[17] T. G. Susko, "MIT Skywalker : A novel robot for gait rehabilitation of stroke and cerebral palsy patients," Ph.D. dissertation, Massachusetts Institute of Technology, 2015.

[18] D. Aoyagi, W. Ichinose, S. Harkema, D. Reinkensmeyer, and J. Bobrow, "A robot and control algorithm that can synchronously assist in naturalistic motion during body-weight-supported gait training following neurologic injury," *IEEE Trans. on Neural Systems and Rehab. Eng.*, vol. 15, no. 3, pp. 387–400, Sept 2007.

[19] Y. Stauffer, Y. Allemand, M. Bouri, J. Fournier, R. Clavel, P. Metrailler, and R. Brodard, "The WalkTrainer A New Generation of Walking Reeducation Device Combining Orthoses and Muscle Stimulation," *IEEE Tran. on Neural Systems and Rehab. Eng.*, vol. 17, no. 1, pp. 38–45, 2009.

[20] J. Burgess, G. Weibel, and D. Brown, "Overground walking speed changes when subjected to body weight support conditions for non-impaired and post stroke individuals," *J. of NeuroEngineering and Rehab.*, vol. 7, no. 1, 2010.

[21] T. P. Luu, H. Lim, X. Qu, and K. Low, "Pelvic motion assistance of nature gaits with adaptive body weight support," in *Asian Control Conf.*, 2011, pp. 950–955.

[22] H. Vallery, P. Lutz, J. Von Zitzewitz, G. Rauter, M. Fritschi, C. Everarts, R. Ronsse, A. Curt, and M. Bolliger, "Multidirectional transparent support for overground gait training," in *IEEE Int. Conf. for Rehab. Robotics*, 2013, pp. 1–7.

[23] M. Orendurff, A. Segal, G. Klute, J. Berge, E. Rohr, and K. N.J., "The effect of walking speed on center of mass displacement," *J Rehabil Research and Development*, vol. 41, no. 6A, pp. 829–34, 2004.

[24] G. Colombo, M. Joerg, R. Schreier, and V. Dietz, "Treadmill training of paraplegic patients using a robotic orthosis," *J. of Rehab. Research and Development*, vol. 37, no. 6, pp. 693–700, 2000.

[25] A. Erdogan, B. Celebi, A. C. Satici, and V. Patoglu, "AssistOn-Ankle: A reconfigurable ankle exoskeleton with series-elastic actuation," *Autonomous Robotics*, 2016.

[26] H. Munawar, M. Yalcin, and V. Patoglu, "AssistOn-Gait: An overground gait trainer with an active pelvis-hip exoskeleton," in *IEEE Int. Conf. for Rehab. Robotics*, 2015.