Design of Human – Machine Interface and Altering of Pelvic Obliquity with RGR Trainer

Maciej Pietrusinski, Ozer Unluhisarcikli,
Constantinos Mavroidis*

Mechanical and Industrial Engineering
Northeastern University
360 Huntington Avenue, Boston, MA, 02115, USA
*Corresponding Author - mavro@coe.neu.edu

Iahn Cajigas, Paolo Bonato*

Motion Analysis Laboratory
Spaulding Rehabilitation Hospital
Boston, MA, USA
*Corresponding Author - pbonato@partners.org

Abstract— **The Robotic Gait Rehabilitation (RGR) Trainer targets secondary gait deviations in stroke survivors undergoing rehabilitation. Using an impedance control strategy and a linear electromagnetic actuator, the device generates a force field to control pelvic obliquity through a Human-Machine Interface (i.e. a lower body exoskeleton). Herein we describe the design of the RGR Trainer Human-Machine Interface (HMI) and we demonstrate the system's ability to alter the pattern of movement of the pelvis during gait in a healthy subject. Results are shown for experiments during which we induced hip-hiking – in healthy subjects. Our findings indicate that the RGR Trainer has the ability of affecting pelvic obliquity during gait. Furthermore, we provide preliminary evidence of short-term retention of the modified pelvic obliquity pattern induced by the RGR Trainer.**

Keywords - robotic rehabilitation; stroke; exoskeletons; impedance control

I. INTRODUCTION

Each year 800,000 people suffer a stroke in the United States alone [1]. Stroke is a leading cause of disability. Stroke survivors experience weakness and difficulties moving one side of the body (i.e. they are affected by hemiparesis), with a negative effect on the performance of motor activities such as walking. Walking allows individuals to perform activities of daily living [2-3]. The ability to walk is strongly correlated with quality of life [4]. Hemiparesis and abnormal synergy patterns are characteristic of gait disorders following stroke. Abnormal synergy patterns include equinus synergy, paretic synergy and reflex coactivation [5]. Comfortable walking speed is reduced in stroke survivors. Asymmetries mark post-stroke ambulation. Asymmetry of stance time during gait, a common feature following stroke, often limits walking efficiency, results in instability, and causes an aesthetically sub-optimal gait pattern. Therefore, the restoration of a normal gait pattern is an important goal of post-stroke rehabilitation.

Many rehabilitation approaches have been used to promote functional recovery in stroke survivors. Unfortunately, the rehabilitation process is labor intensive, since it often relies on a one-to-one administration of therapy, i.e. clinicians work with a single patient at a time. Robotic systems for gait retraining have been recently developed to facilitate the administration of

intensive therapy. Most of the existing systems focus on the correction of primary gait deviations, such as knee hyperextension during stance and stiff-legged gait (defined as limited knee flexion during swing). It is often assumed that secondary deviations would be no longer observed once primary deviations are corrected. Secondary gait deviations are gait abnormalities that result from compensatory movements associated with a primary gait abnormality. Secondary deviations often involve the control of the pelvis. For instance, stiff-legged gait is often associated with hip hiking and/or circumduction of the limb. Hip hiking is an exaggerated elevation of the pelvis on the affected side of the body to allow toe clearance during swing. Circumduction of the limb is marked by an exaggerated hip abduction in combination with an exaggerated rotation of the pelvis. Abnormal control of pelvic obliquity and pelvic rotation are common secondary gait deviations observed in stroke survivors. They are often present even after rehabilitation interventions addressing the primary gait deviation that they are thought to be related to (e.g. limited knee flexion during swing, namely stiff-legged gait).

Existing systems for robotic-assisted gait training typically neglect gait deviations associated with an abnormal control of the pelvis. The Pelvic Assist Manipulator (PAM) is one of the few robotic devices that attempts to address such gait deviations [6]. However, the PAM is designed to control five degrees of freedom of the pelvis during gait. Given that abnormalities in the control of the pelvis during gait most often affect obliquity and rotation of the pelvis, one would wonder whether a simpler system that mechanically allows translations of the pelvis while controlling obliquity and rotation of the pelvis would be sufficient to achieve correction of secondary gait deviations affecting pelvic movement control. This paper summarizes the development of a system, the Robotic Gait Rehabilitation (RGR) Trainer, aimed to control the obliquity of the pelvis during gait training.

In a previous publication [7], we presented the RGR Trainer V1.0. The system was marked by a simple design aimed at achieving preliminary evidence of the validity of the proposed approach. Tests in healthy subjects showed that the RGR Trainer V1.0 could impart gait restoration forces in response to abnormal pelvic obliquity (hip hiking was

simulated by healthy subjects). In this paper, we present the RGR Trainer V1.5, with a newly-designed, more advanced Human-Machine Interface (HMI) system and an implementation of the human-machine synchronization algorithm based on the method previously proposed by Aoyagi et al [6]. We also present results of healthy subject testing, which suggest that the RGR Trainer can coax healthy subjects to alter their pelvic movement pattern during gait to exhibit a secondary gait deviation, namely hip-hiking. Just as importantly, we provide preliminary evidence of short-term retention of this newly acquired gait pattern.

II. RGR TRAINER: BRIEF DESCRIPTION

A. Mechanical Design

The RGR Trainer (Fig. 1) was designed to apply a force-field to the subject's pelvis in order to affect its obliquity via the HMI worn by the subject. The device features horizontal motion modules on the left and right sides of the subject. This design gives the subject unrestricted motion in the horizontal plane within a given range. A servo – tube linear electromagnetic actuator (model STA2508) from Copley Controls Inc. (Canton, MA, USA) is supported by one of the horizontal motion modules, allowing it to glide in the horizontal plane and apply forces in the vertical direction to control pelvic obliquity. A spherical joint is used to transfer forces from the actuator to the HMI, while a tension – compression load cell provides force feedback for control and performance evaluation purposes. The servo – tube linear actuator features integrated linear position sensing of the moving core (thrust rod), providing vertical position of the HMI on that side. On the opposite side of the body, a horizontal motion module supports a lightweight assembly with a linear potentiometer, which provides vertical position feedback of that side.

B. Device Operation

The position feedback from the servo tube actuator and the linear potentiometer on the opposite side of the body are used to estimate the angular position of the HMI worn by the subject in the frontal plane. It is this angular position that is acted upon by a force field. The patient is also given freedom as to his location in the horizontal plane, therefore minimizing the perturbations to the patient's natural gait. This is an important characteristic of the RGR Trainer. A study examining postural hip muscle activity during recovery from stroke found that horizontal force disturbances as low as 2% of the subject's body weight led to loss of balance [8].

C. Control System

The force field is realized in the physical sense by use of impedance – controlled linear actuator. The controller has been designed based on the method originally proposed by Hogan [9]. The system creates a virtual spring/mass/damper attached to the HMI, such that deviations of pelvis' obliquity from the reference trajectory result in corrective forces applied to the pelvis. Details concerning the control system can be found in our previous publication [7].

Figure 1. A subject in the RGR Trainer, wearing the new Human-Machine Interface.

D. Environment Compliance

For our purposes, we define the environment to be the collection of bodies between the rigid endpoint of the actuator and the rigid pelvis of a subject. Therefore, the environment is comprised of the HMI and the lower body's soft tissue in contact with it.

The RGR Trainer is designed to apply a force field to the subject's pelvis to affect its obliquity. The impedance controller can coax the linear actuator to display virtual stiffness and virtual damping (along with some apparent inertia). Since the environment displays a certain amount of stiffness and damping, the force field experienced by the subject's pelvis is a combination of the virtual stiffness/damping and the environment's stiffness/damping. The two are connected in series. Focusing on the stiffness, the virtual stiffness (K_c) necessary to generate a total force field stiffness (K_d) around the pelvis depends on the environment stiffness (K_e) and is computed using equation (1).

$$K_c = (K_d K_e)/(K_e - K_d) \qquad (1)$$

Therefore, in order to apply a force-field of certain strength (stiffness) to the subject's pelvis, the environment's stiffness must be known.

III. HUMAN-MACHINE INTERFACE

A. Motivation Behind Developing the New HMI

The RGR Trainer V1.0 had employed a commercially available Newport 4 Hip Orthosis (Orthomerica Inc., Orlando, FL), to transfer forces exerted by the device to the subject's pelvis. Testing with the Newport 4 Hip Orthosis suggested that transferring large forces in a repeatable manner to the pelvis in the frontal plane (to affect pelvic obliquity) requires attaching the HMI to the subject's legs. While this solution served well as a proof-of-concept, enabling significant corrective forces to be transferred to the pelvis, both comfort and adherence to the body were sub-optimal. The thigh components caused interference on the medial side of the thighs, and hip abduction/adduction was effectively blocked, which significantly altered the subjects' gait. These issues prompted us to develop the custom HMI for the RGR Trainer V1.5, which is a lower body exoskeleton.

B. Background on Lower Body Exoskeletons

There are many lower body exoskeletons that include a pelvic attachment and lower limb components coupled together. Three examples of such exoskeletons are the Lokomat [10], the LOPES [11] and the BLEEX [12]. These three devices provide the user with a certain number of DOFs in the pelvic region, both actuated and non-actuated. The Lokomat and LOPES lower body exoskeletons are supported by a stationary frame, and are not portable. The Lokomat's hip joints operate in flexion/extension only, with hip abduction/adduction (in the standard version of the system) and horizontal plane pelvis translations locked; pelvis translation in the vertical direction is allowed. The LOPES goes a step further, allowing hip flexion/extension and abduction/adduction and all three pelvis translations, but there is no explicit provision made for pelvic obliquity or pelvic rotation. The BLEEX is an autonomous exoskeleton with hip flexion/extension, abduction/adduction and external/internal rotation (non-collocated) DOFs. The BLEEX is portable and therefore the motion of its pelvic component is not restricted by an external frame. Interestingly, there is no explicit provision made for allowing the rotation of the pelvis with respect to the upper torso in the frontal plane (pelvic obliquity).

C. Overall Design of the HMI

The RGR Trainer's primary focus is the control of pelvic obliquity angle. One study with hip-hiking subjects found its magnitude to be on average about 6 degrees [13], therefore it was important to design our exoskeleton such that pelvic obliquity is not explicitly hindered by the orientation of the torso (in the frontal plane) and that application of corrective moments to the exoskeleton to affect pelvic obliquity does not directly apply moments to the upper torso. On the other hand, from our experience with the Newport 4 system we found that using just a pelvic brace/belt does not allow for reliable measurement of pelvic obliquity or transfer of forces/moments. We determined that it is necessary to employ mechanical links to the legs to transfer forces/moments to the pelvis. The incremental process employed to develop the HMI revealed
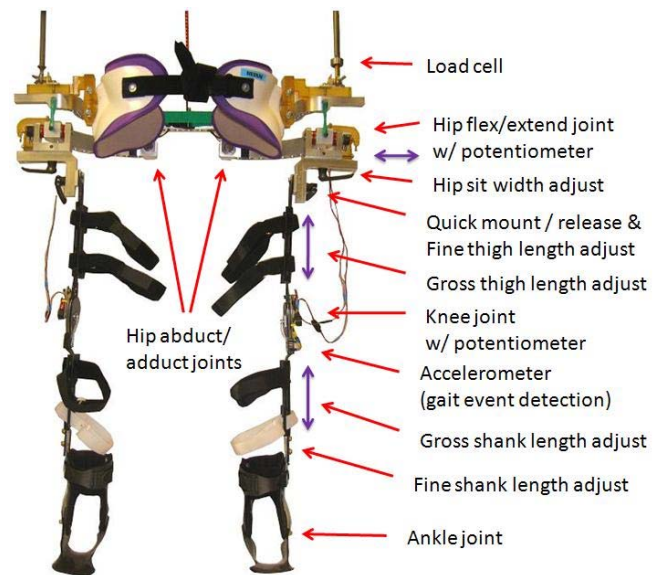

Figure 2. The Human-Machine Interface is made up of the pelvic brace, the coupling and the leg braces.

that a design with ankle-foot components used for the purpose of anchoring the exoskeleton onto the lower body would improve force transfer and significantly reduces migration.

The developed HMI is shown in Fig. 2. A pelvic brace wraps around the subject's waist, covering the greater trochanters of the femurs and bracing the iliac crests, as shown in Fig. 3. This geometry allows for closely tracking the motion of the pelvis, but the HMI is not directly affected by the motion of the upper torso. Straps are used for position adjustment, as can be seen in Fig. 4.

The pelvic brace is coupled to two leg braces and to the RGR Trainer's actuation system via four identical rotational joints. The joints are double-supported with two roller bearings and two thrust bearings each. A horizontal center piece in the back locates the two abduction/adduction joints coincidentally with the subject's hip joints in the frontal plane. Two curved arms, one on each side, are bolted to the center piece, and are used to transfer forces generated by the RGR Trainer's actuation system. Two 'abductors' rotate about the abduction – adduction joints, and transfer forces/moments to the flexion – extension joints, located in the frontal plane. Thus, the four rotational joints together produce two remote-center-of-rotation joints, which coincide with the subject's hip joints as shown in Fig. 3. The location of the abduction – adduction joints can be
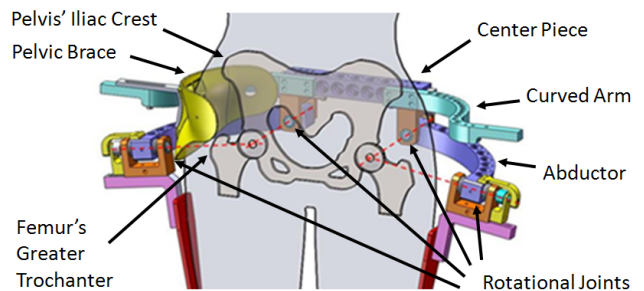

Figure 3. The four rotational joints produce two remote-center-of-rotation joints, which coincide with the subjects' hip joints.

shifted laterally (3 discrete positions) to accommodate for a range of subjects' sizes. The rotational joints were designed to support a 110 kg (244 lb) person with a safety factor of 2. The 110 kg mass corresponds to the 99th percentile US population male, as found in [14]. The structural components were optimized for light weight with FEA. They were machined from 7075 T6 aluminum stock.

The leg braces were built using DonJoy post-op TROM knee braces and Velocity ankle braces (DJO Inc., Vista, CA). Though the leg braces were not optimized for the task at hand, they have a positive effect on the HMI's ability to transfer forces to the pelvis. The ankle braces, wrapping around the feet, provide excellent force transmission to the pelvis. Though external/internal rotation of the leg is not accounted for in the HMI with a rotational joint, the flexibility of the TROM knee brace allows for a significant range of motion in that DOF.

D. Human Factors

The HMI was designed to fit a range of subjects. Several features of the system can be adjusted to fit the subject's anatomy, such as hip brace width, fore-aft and lateral position of the hip brace around the waist, vertical position of the remote centers of rotation (HMI's hip joints), thigh lateral side positions, hip joint to knee joint distance and knee joint to ankle joint distance.

The HMI was also designed to be donned and doffed as quickly and effortlessly as possible. The two leg braces are disconnected from the rest of the HMI and donned by the subject in a seated position. This gives the practitioner easy access to the hardware for link length and strap adjustment. Next, the subject enters the device and steps backwards into the pelvic brace, which is suspended by the actuation system in the RGR Trainer. The pelvic brace is attached to the subject, and the leg braces are fixed to the coupling's flexion/extension joints via bicycle-style quick release clamps. The whole process minimizes the subject's idle time in the machine, thereby reducing fatigue.

IV. HUMAN-MACHINE SYNCHRONIZATION

A. Introduction

Previous force field application tests in healthy subjects with the RGR Trainer V1.0 required the subjects to adjust their cadence to match the timing of the reference trajectory, which was "replayed" by the system at a constant speed. This took a considerable amount of practice and effort, and cannot be expected to be a suitable approach when training impaired subjects. Therefore, a synchronization algorithm was implemented based on the method previously introduced by Aoyagi et al. [6].

B. Algorithm Operation

The algorithm estimates the actual temporal position of the subject within his gait cycle based on the angular positions and velocities of the hip and knee joints. It relies on the fact that signals from several DOFs of lower limbs produce a combination, which is unique to that particular location in the gait cycle. The algorithm uses a synchronization reference
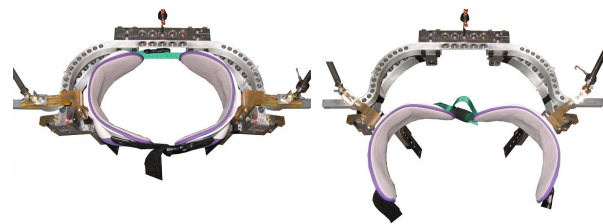

Figure 4. Top view of HMI (leg braces removed for clarity). The pelvic brace is shown in closed position (left) and open position (right). Adjusting the front and back straps shifts the fore-aft location of the HMI around the waist.

(lasting exactly 1 cycle), which is composed of eight representative time series (representative for that subject at a particular walking speed and cadence). Individual cycles used to generate the reference are found by detecting discrete gait events, like heel strike and toe off, with an accelerometer mounted on the leg brace. The algorithm finds in the reference a discrete point, which most closely matches the combination of the 8 signals at a particular instant, from the 8 DOFs. The location of this discrete point is normalized, generating a number between 0 and 0.99, which corresponds to the location in the gait cycle. This 'raw' result is conditioned via smoothing. Finally, the output is used to index through the obliquity reference trajectory such that discrete points are presented to the impedance controller at the proper time. The synchronization algorithm operates at 500 Hz and runs using LabVIEW Real-Time on a dedicated PC, with one of the CPU's 4 cores dedicated solely to this operation.

C. Performance of the Synchronization Algorithm

The synchronization algorithm was tested by employing the accelerometer-based gait events, and generating a curve that, under the assumption of constant cadence, very closely matches the true gait cycle position curve. As we can see in Fig. 5, after a positive offset is introduced to eliminate time delay (20% of gait cycle), the output of the synchronization algorithm, T_{rep}, follows very closely the true gait cycle position.
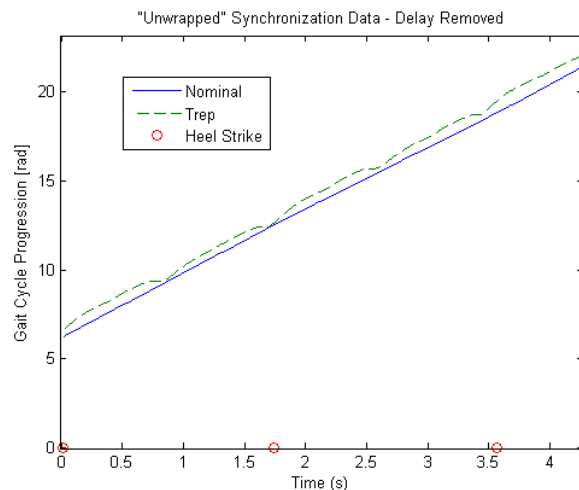

Figure 5. The output of the synchronization algorithm estimates the subjects' position in the gait cycle very well.

V. HEALTHY-SUBJECT TESTING

A. Overview of the Experiments

The aim of the experiments carried out in the study was to answer the following question: can the RGR Trainer generate appropriate forces, and can the HMI system transfer these forces to a subject in such a manner as to promote altering of the gait pattern?

With stroke survivors, altering the gait pattern implies reducing the degree of hip-hiking, which they exhibit. Since we were not ready to test stroke survivors and we did not want to draw conclusions from a test in which we asked healthy subjects to simulate hip-hiking, we instead designed an experiment in which we attempted to induce a hip-hiking gait pattern in healthy subjects.

B. Hip-Hiking Reference Trajectory

We used a trajectory of pelvic obliquity from a recent study by Cruz [13], in which the authors analyzed the trajectory of the pelvis during ambulation in 18 stroke survivors exhibiting hip-hiking. Using this reference trajectory, shown in Fig. 6 with a healthy subject, the RGR Trainer generates forces of proper direction (to elevate a particular side of the pelvis), while it is understood that a certain amount of position error is expected to be seen. This is due to the limited strength of the total force field acting upon the HMI (finite environment stiffness and actuator-generated force field strength), as well as the constant action of gravity.

C. Hip Hiking Experiment Implementation Details

The RGR Trainer generated forces to be applied to the subject's pelvis 'without notice'. The purpose of this modality of force generation was to induce a hip-hiking pattern. We expected the maximum magnitudes of interaction force to be the largest during the first cycle, and to diminish over several subsequent cycles as the subject adjusted their gait to comply more closely with the hip-hiking pattern. Changes in pelvic motion were achieved via switching between two reference trajectories: the subject's own obliquity trajectory and the pelvic obliquity trajectory derived from Cruz et al. [13]. A smooth switch was achieved by using a sigmoid-like curve lasting 100 msec.
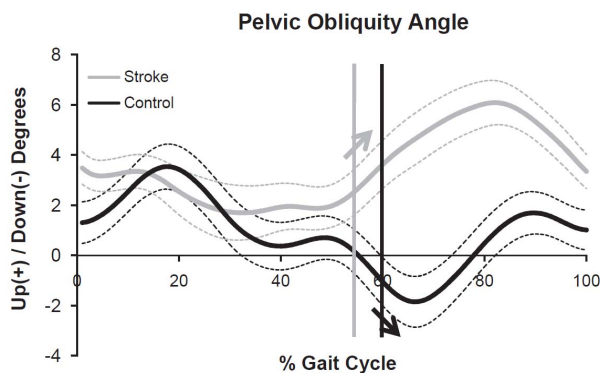

Figure 6. Pelvic obliquity (healthy and impaired subjects) [13].

D. Hip Hiking Experiment Setup

Subjects donned the leg braces and entered the RGR Trainer, where the rest of the HMI was fitted to their body. The environment stiffness was measured by controlling the actuator to apply sinusoidal force profile onto the HMI, and record the resulting displacement. This was performed at three force levels and at 1 Hz. Next, as the subjects ambulated at 1.8km/h (0.5m/s) with the RGR Trainer in back-drivable mode, the baseline pelvic obliquity and hip and knee angular position and velocity time-series, as well as the left shank – mounted accelerometer signal were recorded. The time-series were sectioned into individual gait cycles to produce the subjects' averaged baseline pelvic obliquity trajectories and synchronization trajectories. The measured environment stiffness value was used in (1) to calculate the controller proportional gain K_c necessary to attain the total desired force field stiffness K_d in the subsequent test.

E. Hip Hiking Experiment Protocol

The experiments lasted 7 minutes. For the first minute, the subjects walked freely within the RGR Trainer (in back-drivable mode). The device was synchronized to the subject's gait using the synchronization algorithm and the subject's own synchronization reference. During the 2nd minute, the force field was activated, while the subjects ambulated with their own baseline serving as the reference. Therefore, any deviation of the measured pelvic obliquity from the reference resulted in corrective forces being applied onto the subject. Starting at the 3rd minute of the experiment, the RGR Trainer switched to a hip-hiking pelvic obliquity pattern as the position reference, with the force field still active. The resulting forces provided haptic feedback to the subjects to alter their gait and adopt a hip-hiking pattern in their pelvic obliquity. This lasted for 3 minutes. Starting at the 6th minute, the position reference was switched back to the subjects' baseline pelvic obliquity, with the force field active. Finally, during the 7th minute of the experiments, the force field was de-activated, hence the subjects walked freely within the device, which was in back-drivable mode.

F. Environment Stiffness Measurement Results

Three healthy subjects participated in the experiment. The environment stiffness of each subject is listed in Table I. The stiffness values across the three subjects and force levels show much less variability than those reported in our previous paper [7] with the Newport 4 pelvic brace and thigh components, having ranged between 11 and 25 N-m/deg, with highest stiffness values occurring at lowest force level.

It was observed that a significant amount of flex occurred at the leg braces' thigh components. These components were not designed to resist the significant moments applied to the HMI.

G. Hip Hiking Experiment Results

Three levels of desired force field strength K_d were tested: 8, 6 and 4 N-m/deg. Environment stiffness of 12 N-m/deg was used with each K_d, which produced actuator impedance values ranging from 6 to 24 N-m/deg. Fig. 7 shows results from a test with subject 3, with the desired force field strength K_d set to 6

N-m/deg, which produced the actuator impedance gain of $K_c = 12$ N-m/deg. As the position reference was switched from subject's own pelvic obliquity to hip-hiking obliquity, the interaction force reached 200N, and decreased exponentially as the subject adopted a new, altered gait pattern.

TABLE I. ENVIRONMENT STIFFNESS

Force Level	Subject / Stiffness [N-m/deg]		
	1	*2*	*3*
40N	12	13	13
60N	11	12	12.5
80N	9	11	12

The results of the environment stiffness experiments, shown in Table I, suggest that the new HMI offers an improved performance compared to the previously – used method. While the maximum stiffness values recorded with the HMI are lower, those readings occurred at the lower end of the applied force levels, and when higher force levels were used, this discrepancy was not observed anymore. The previous method transferred forces to the body mostly through soft tissue, the stiffness of which may not vary linearly with the applied shear force. On the contrary, the HMI relies on the attachment to the feet, in order to transfer forces to the legs. In fact, the relatively small variability of environment stiffness figures across three different subjects suggests that a significant amount of the compliance in response to the applied forces, which produced the stiffness figures, may have occurred in the HMI itself, as opposed to the subject's body, while not restricting leg abduction/adduction, attesting to the effectiveness of the method.

H. Discussion

The hip hiking experiment was designed to demonstrate that the RGR Trainer can teach a subject a new gait pattern. The results suggest that this is in fact what happened during the tests. In the sample data shown in Fig. 7, we observe a large spike in the interaction force as the hip-hiking position reference is introduced. The exponential decay trend line fitted to the in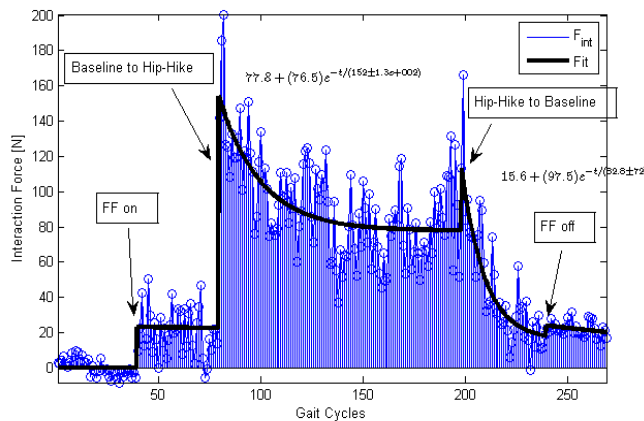teraction force data shows that the subject alters his gait pattern in order to reduce these interaction forces. The short 1 minute period to re-acclimate to baseline pelvic obliquity is not quite long enough to reach a steady-state level.

CONCLUSION

The RGR Trainer generates a force field to control pelvic obliquity through the HMI. The system is able to alter the pattern of movement of the pelvis during gait in a healthy subject. We conducted experiments during which we induced hip-hiking in healthy subjects, and the results suggest that the RGR Trainer has the ability of affecting pelvic obliquity during gait, with preliminary evidence of short-term retention of the modified pelvic obliquity pattern. Finding the optimal force field strength to maximize retention will be the goal of future experiments.

REFERENCES

[1] V. L. Roger, *et al.*, "Heart disease and stroke statistics--2011 update: a report from the American Heart Association," *Circulation,* vol. 123, pp. e18-e209, Feb 1 2011.
[2] J. M. Guralnik, *et al.*, "Maintaining mobility in late life. I. Demographic characteristics and chronic conditions," *Am J Epidemiol,* vol. 137, pp. 845-57, Apr 15 1993.
[3] J. M. Guralnik, *et al.*, "Lower-extremity function in persons over the age of 70 years as a predictor of subsequent disability," *N Engl J Med,* vol. 332, pp. 556-61, Mar 2 1995.
[4] H. C. Chiu, *et al.*, "Physical functioning and health-related quality of life: before and after total hip replacement," *Kaohsiung J Med Sci,* vol. 16, pp. 285-92, Jun 2000.
[5] E. Knutsson and C. Richards, "Different types of disturbed motor control in gait of hemiparetic patients," *Brain,* vol. 102, pp. 405-30, Jun 1979.
[6] D. Aoyagi, *et al.*, "A robot and control algorithm that can synchronously assist in naturalistic motion during body-weight-supported gait training following neurologic injury," *IEEE Trans Neural Syst Rehabil Eng,* vol. 15, pp. 387-400, Sep 2007.
[7] M. Pietrusinski, *et al.*, "Robotically Generated Force Fields for Stroke Patient Pelvic Obliquity Gait Rehabilitation," presented at the 2010 IEEE International Conference on Robotics and Automation (ICRA2010), Anchorage AK, 2010.
[8] S. G. Kirker, *et al.*, "Changing patterns of postural hip muscle activity during recovery from stroke," *Clin Rehabil,* vol. 14, pp. 618-26, Dec 2000.
[9] N. Hogan, "Stable execution of contact tasks using impedance control," in *IEEE International Conference on Robotics and Automation*, Raleigh, NC, 1987, pp. 1047-1054.
[10] M. Bernhardt, *et al.*, "Physiological Treadmill Training with the 8-DOF Rehabilitation Robot LOKOMAT," presented at the Jahrestagung der deutschen Gesellschaft für biomedizinische Technik, Nürnberg (Germany), 2005.
[11] J. F. Veneman, *et al.*, "Design and evaluation of the LOPES exoskeleton robot for interactive gait rehabilitation," *IEEE Trans Neural Syst Rehabil Eng,* vol. 15, pp. 379-86, Sep 2007.
[12] A. B. Zoss, *et al.*, "Biomechanical design of the Berkeley lower extremity exoskeleton (BLEEX)," *IEEE/ASME Transactions on Mechatronics,* vol. 11, pp. 128-138, 2006.
[13] T. H. Cruz, *et al.*, "Biomechanical impairments and gait adaptations post-stroke: multi-factorial associations," *J Biomech,* vol. 42, pp. 1673-7, Aug 7 2009.
[14] H. Dreyfuss, *The measure of man: human factors in design*: Whitney Library of Design, 1967.

Figure 7. Result of hip-hike inducing test (subject 3, $K_d = 6$kN/m).