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Robotically Generated Force Fields for Stroke Patient Pelvic Obliquity Gait Rehabilitation

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Abstract — **The Robotic Gait Rehabilitation (RGR) Trainer, was designed and built to target secondary gait deviations in patients post - stroke. Using an impedance control strategy and a linear electromagnetic actuator, the device applies a force field to control pelvic obliquity through an orthopedic brace while the patient ambulates on treadmill. Healthy human subject testing confirmed efficacy of the method to impart significant gait restoration forces as a response to abnormal pelvic obliquity (hip hiking). This novel approach to application of force fields using endpoint impedance controlled linear actuators takes into account soft tissue compliance.**

I. Introduction

EACH year in the United States alone, over 750,000 people suffer strokes. Stroke is the leading cause of disability, and about 80% of stroke victims experience weakness or trouble moving one side of their body, and require rehabilitation [1]. Walking allows individuals to perform Activities of Daily Living (ADLs), and the ability to walk is also strongly correlated with the quality of life [2-4]. Hemiparesis in the affected side and abnormal synergy patterns are characteristic of gait disorders following stroke. Abnormal synergy patterns include equinus synergy, paretic synergy and reflex coactivation [5]. The commonly observed comfortable walking speed (CWS) among stroke survivors is 0.5 m/s which is less than half of that seen in healthy subjects (1.52 m/s) [5, 6]. Asymmetries mark post-stroke ambulation, which are caused in part by weakness in the affected limb and abnormal synergy patterns of both the affected and unaffected limbs. 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, restoration of a normal gait pattern is a frequent goal of post-stroke rehabilitation.

Many rehabilitation interventions 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 the time. Robotic systems for gait retraining have been recently developed to facilitate administration of intensive gait retraining 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). To our knowledge, only one robotic device so far attempts to address secondary gait deviations, while addressing primary gait deviations during therapy [10]. Secondary gait deviations are gait abnormalities that result from compensatory movements associated with a primary gait abnormality. Secondary deviations often involve motor control of the pelvis. For instance, stiff legged gait is associated with hip hiking or circumduction. Hip hiking is an exaggerated elevation of the pelvis on the affected side to allow toe clearance during swing, while circumduction is an exaggerated rotation of the pelvis in combination with an exaggerated hip abduction on the affected side. Abnormal control of pelvic obliquity and pelvic rotation are the most common secondary deviations observed in post-stroke subjects. Our study focuses on these gait abnormalities associated with stiff legged gait, the most common primary gait deviation in post-stroke subjects.

From the point of view of training strategies and robot control, there are two types of robotic devices for gait rehabilitation: devices which drive the body in position mode regardless of patient effort, and devices which apply force-fields to the body, therefore modulating the forces applied onto the body depending on patient effort. The latter method, employing force-fields, has been shown to be the preferred method for retraining post-stroke subjects to regain their motor functions in upper – limb exercises [7], and we believe that it is the preferred method for gait rehabilitation as well. Three examples of robotic systems which apply force fields to body components to affect gait are the Lokomat, LOPES and Pelvic Assist Manipulator (PAM).

The Lokomat (Hocoma AG, Switzerland) uses a bilateral robotic gait orthosis attached to the thighs and shanks of the subject, which consequently restricts pelvic obliquity and rotation. The device controls the patient's leg movements in

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the sagittal plane through actuation of the hip and knee joints (4 total degrees of freedom - DOF). Although designed to be position – controlled, the Lokomat now uses impedance control to apply corrective force fields to the patient's legs [8].

The LOPES, a rehabilitative robotic device developed at the University of Twente in the Netherlands expands on the Lokomat's functionality by allowing leg abduction / adduction (2 DOF), as well as pelvis translation (3 DOF) for a total of 8 DOF [9]. Pelvis rotations are constrained.

The Lokomat and LOPES use impedance control strategies with force feedback to target primary gait deviations, while secondary gait deviations in the pelvis are not targeted. The Pelvic Assist Manipulator (PAM) is a robotic rehabilitative device, which was designed to apply corrective forces to the pelvis and hence could be used for correction of secondary gait abnormalities. PAM uses pneumatic cylinders to apply forces in five degrees of freedom (three translations and two rotations) [10]. The PAM "drives" the movement of the pelvis. Subjects are not encouraged by the device to actively control pelvis movement, but it is rather the device that "moves" their body.

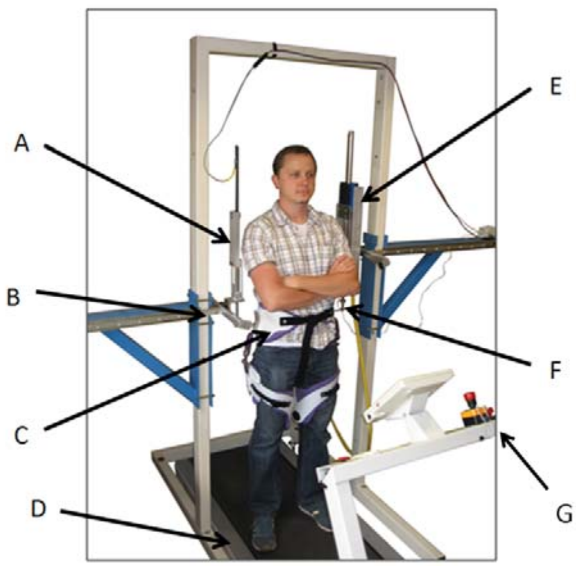

Fig. 1. RGR Trainer. A - linear potentiometer, B – linear guides, C – pelvic brace, D – treadmill, E – servo – tube linear actuator, F – load cell and spherical joint, G – emergency stop within subject's reach. A safety suspension harness to be used with patients is not shown.

In this paper the RGR Trainer v.1 is presented (see Fig. 1) to facilitate robotic gait retraining using force-fields applied to the pelvis to correct secondary gait deviations in pelvic motion. The RGR Trainer was designed to implement impedance control based human-robot interface modalities that allow post-stroke patients to interact with the device in ways that mimic the interaction with a therapist walking side-by-side with the patient and manually assisting movement of the patient. Our system addresses soft tissue compliance when applying force fields to the pelvic region. This system was designed to apply force fields to pelvic obliquity only, with the remaining DOFs left free. We believe that allowing the patients to execute their natural patterns of pelvis translation will lead to a feeling of more natural walk, and better control of their balance, which thus will lead to better results.

II. RGR Trainer: System description

A. Mechanical Design

The RGR Trainer (Fig. 1) was designed to apply force-fields to the subject's pelvis via an orthopedic pelvic brace at the pelvic obliquity level (in the frontal, or coronal plane). The RGR Trainer is composed of a frame from the Biodex Unweighing System (Shirley, NY, USA), and features

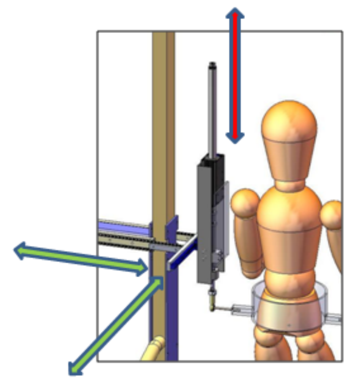

Fig. 2. Subject wearing pelvic brace with actuator attached and applying forces in the vertical direction. Slides provide free motion in horizontal plane.

horizontal motion modules on the left and right sides of the subject. These modules consist of linear motion slides, which are positioned longitudinally and transversely (with respect to the treadmill). Therefore, this design allows for largely unrestricted motion in the horizontal plane within a limited range (Fig. 2).

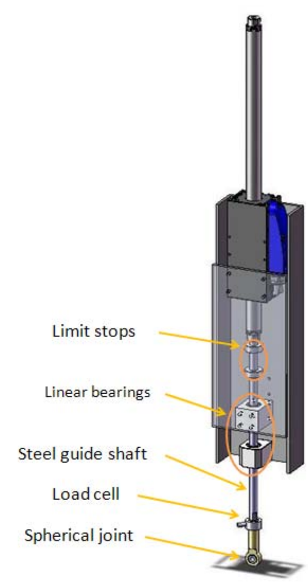

Fig. 3. Actuator assembly, with a load cell and spherical joint at end point.

A servo – tube linear electromagnetic actuator (model STA2504) from Copley Controls Inc. (Canton, MA, USA) is supported by one of the horizontal motion modules, allowing it to glide in the horizontal plane, while resisting motion in the vertical direction. The thrust rod is extended by a precision shaft, which is guided by two linear ball bearings. A spherical joint is used to transfer forces from the actuator to the pelvic brace, while a tension – compression load cell provides force feedback for control and performance evaluation purposes (Fig. 3). Hall – effect sensors provide actuator position feedback by sensing the series of permanent magnets in the thrust rod. 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 are used to calculate the angular position of the pelvic brace in the patient's frontal plane (at pelvic obliquity level). This angular position of the pelvic brace is acted upon by a force field, which is physically applied by the end point impedance controlled actuator. Owing to the device's design, the patient's position in the vertical direction is not restricted (within 14cm range of motion) or directly acted upon by the actuator (which is gravity – compensated). The patients are also given freedom as to their location in the horizontal plane, therefore minimizing the perturbations to the patients' natural gait. This is an important characteristic, since 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 [11].

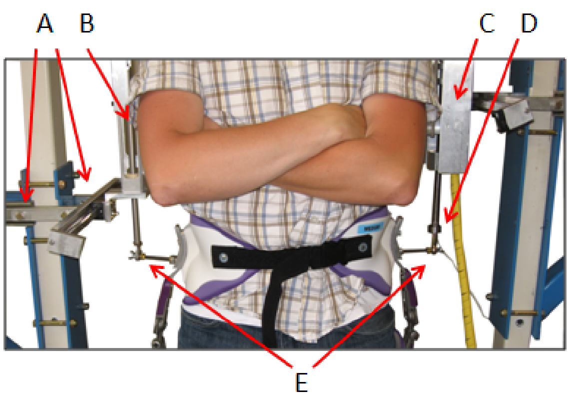

Fig. 4. Subject inside the RGR Trainer. A – linear guides, B – linear potentiometer, C – linear actuator, D – load cell, E – spherical joints.

C. Control System

The force fields are realized in the physical sense by use of an impedance controller. In general, impedance control architecture is comprised of an inner unity feedback force loop, and an outer position unity feedback loop. An end point impedance controller for the linear actuator has been designed, shown in Fig. 5, based on the design method and controller presented in [12] for a two link robotic manipulator. The derivation resulted in Equ. (1) for the force command F_{act} sent to the actuator (Fig. 5) where x is the actuator displacement. The two position loop gains K_c and B_c represent the virtual spring stiffness and virtual damping (at the actuator's endpoint), while G is the proportional force loop gain. A compression – tension load cell measures the interaction force between the brace and the actuator. The effect of the force feedback is such that the system's apparent inertia, as perceived by the subject, is reduced by the factor of G/(G+1). Due to the fact that force control is largely dependent on the environment [13], the force gain G was tuned experimentally through bench testing while the actuator's end point interacted with the subject via the pelvic brace (see Section III. B).

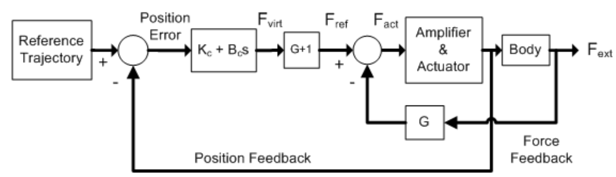

Fig. 5. Block diagram of endpoint impedance controller.

$$F_{act} = (G+1)[K_c(x_0 - x) + B_c(\dot{x}_0 - \dot{x})] - (G)F_{ext} \quad (1)$$

D. Controller Physical Implementation

The end point impedance controller was implemented in LabVIEW Real-Time (National Instruments Inc.) on a dedicated PC (target) with 6259M Data Acquisition Card (DAQ) and the user interface displayed on a Windows PC (host). The control loop operates at 10 kHz, acquiring an emulated encoder signal from the actuator's Xenus servo-amplifier (Copley Controls Inc.), which reflects the linear actuator's position. The linear potentiometer's position signal is low-pass anti-alias filtered (RC 480 Hz cutoff) and digitally filtered within LabVIEW. The obliquity angle θ of the pelvic brace is computed as shown in Fig. 6 using the two linear position signals. Numerical differentiation and a moving average algorithm provide angular velocity feedback. The tension – compression load cell signal is low-pass analog filtered, and digitally filtered within LabVIEW. The control loop's output is an analog force command, which is sent to the Xenus servo-amplifier. The Xenus servo-amplifier operates in current – mode at 15 kHz, using

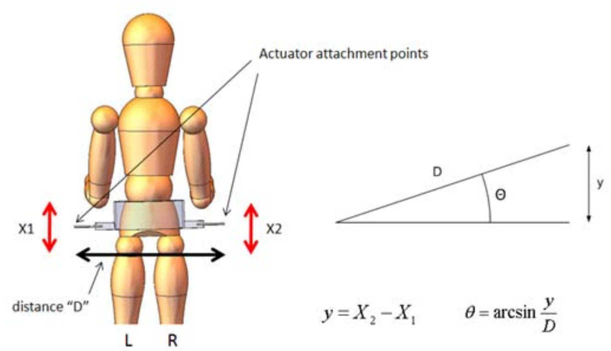

Fig. 6. Measurement of obliquity angle of the pelvic brace.

electrical current consumption information in its own control loop.

E. Safety

In order to ensure safety, an emergency switch is placed within the subject's reach (Fig. 1). This switch depowers both the actuator and the treadmill. In addition, impaired subjects will wear a safety harness in order to prevent falls.

An unsafe condition arises when the DAQ PC encounters an error – the DAQ card maintains the last known output (enable signal and force command signal). To address this potentially unsafe condition, a simple analog circuit was built. The DAQ - generated sinusoidal enable signal is first high-pass filtered, and then rectified with a Gratz bridge rectifier. Finally, a capacitor smoothes the output to the Xenus's Schmidt trigger. In case of an error, DAQ's output changes from 100Hz sinusoid to a constant voltage. This signal is blocked by the high – pass filter, disabling the Xenus amplifier.

F. Environment Compliance

The primary purpose of this system is to apply a force field to the subject's pelvis at the obliquity level. The above-described impedance controller can coax our linear actuator to display virtual stiffness and virtual damping (along with some apparent inertia). In contrast to rehabilitative devices, which apply forces to lower limbs in the transverse direction to the thigh and shank, the RGR Trainer's pelvic brace transfers forces to the pelvic area tangentially, and to the thighs in the longitudinal direction.

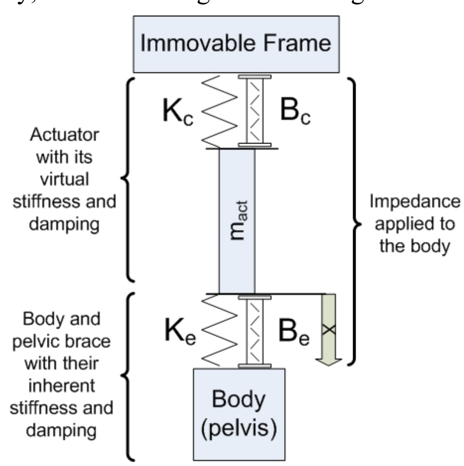

Fig. 7. Schematic of simplified dynamics between actuator and the body.

This necessitates consideration of the body's soft tissue. Since the human body's soft tissue and the pelvic brace display 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 soft tissue's stiffness/damping, as illustrated in Fig. 7. Therefore, the gain (K_c) necessary to impart a desired force field stiffness (K_d) onto the body depends on the environment stiffness (K_e) and is computed from Equ. (2). This means that in order to apply a force-field of certain strength (stiffness and/or damping) to the subject's pelvis, the stiffness and damping of the body's soft tissue and of the pelvic brace (together making up the environment) must be known (see Section III.D for experimental measurement of these parameters).

$$K_c = (K_d K_e)/(K_e - K_d) \tag{2}$$

G. Reference Trajectory

In order to facilitate robotic gait retraining targeting pelvic obliquity using our control scheme, realistic healthy – subject motion trajectories of pelvic obliquity must be provided to the controller as a reference / command input (see Fig. 5).

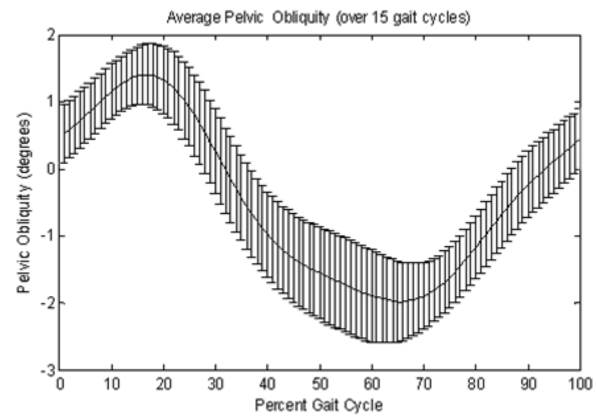

Fig. 8. Sample obliquity angle data averaged from 15 cycles at 1.4 km/h with error bars indicating +/- one standard deviation from the mean.

Lower extremity motion profiles of a 180cm tall, 82kg male were collected using the Vicon motion capture system. Using Matlab (Mathworks, Natick MA), pelvic obliquity was extracted from the 3-D trajectories of four reflective markers on the pelvis, while the markers on the legs were used to find the percentage of gait cycle (Fig. 8). This single pelvic obliquity trajectory served as a reference for the impedance controller in the subsequent healthy human subject testing. Usability and intervention testing with impaired subjects will require a range of trajectories to match individuals. Alternatively, a few reference trajectories will be made customizable to match a specific subject.

III. Testing

A. Actuator Bandwidth

The servo tube actuator's low force bandwidth was measured by commanding a sinusoidal force signal (from 0 to 15Hz) of 15N amplitude (30N peak to peak) to the Xenus servo amplifier. A loadcell measured interaction force between the thrust rod and a compression spring (k=5 kN/m) used to represent a known environmental impedance. A 25N preload was used to ensure positive compressive forces throughout the test. Matlab was used to generate a transfer function of the system. The closed loop TF was found by closing unity gain feedback around the open loop TF. The low pass force feedback filter was included in the model as well. Bode plots of the three systems are shown in Fig. 9.

B. Controller Bench – Testing

The impedance controller was first tested at the actuator's linear motion level, in order to characterize its behavior. A subject donned the Newport 4 pelvic brace and left thigh segment (Fig. 10). The actuator was attached to the pelvic brace, and with the position loop gains set to zero, a range of proportional force-loop gains was tried. The subject moved

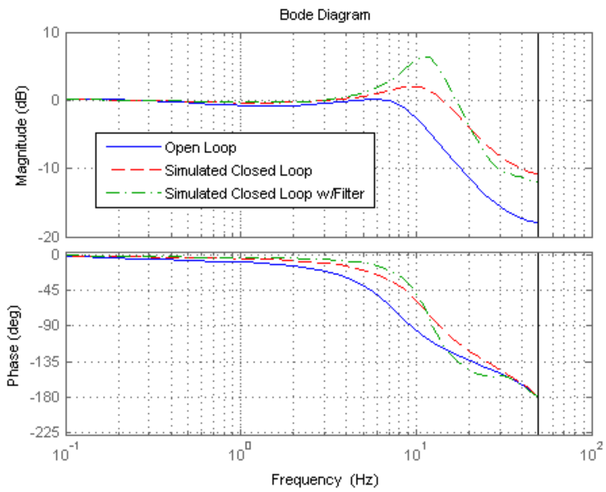

Fig. 9. System's bandwidth test results.

his pelvis in an oscillatory motion several centimeters in the vertical direction to induce a reaction from the system. A serious instability occurred at G=1.8. The highest gain which provided acceptable stable performance was found to be 1.0. This gain results in a reduction of the apparent inertia by 50% as well as a marked reduction in stiction and Coulomb friction, increasing the system's backdrivability.

Fig. 10. Bench-test setup.

Next, the end point impedance controller was characterized. A sinusoidal reference trajectory of 3cm in amplitude and 1Hz frequency was presented to the position loop. With G=1, a range of proportional gains was tried, from 1kN/m to 10kN/m. For convenience, the derivative gain was computed from Equ. (3), based on the damping ratio and the proportional gain value. The test was repeated at 3Hz and 6Hz frequencies.

$$B_c = 2\zeta\sqrt{m_{act} * K_c} \tag{3}$$

Testing revealed that while a certain amount of damping is necessary to produce smooth motion, a damping ratio above 0.5 produced undesirable vibrations at higher proportional gains. The most likely explanation was that the vibrations generated in the force loop caused oscillations in displacement, which in turn affected its derivative. A similar effect had been described in literature [8]. Therefore, limits were set on the maximum proportional gain (virtual stiffness) K_c = 10kN/m, and the damping ratio ζ = 0.5. These two values produce a derivative gain (virtual damping) B_c = 138 N-s/m.

C. Backdrivability Test

Subject 2 (Table I) entered the device and donned the Newport 4 pelvic brace along with the thigh segments. A LabVIEW program designed to apply moment – fields via obliquity – level impedance control was used to conduct the test, but with the position loop gains set to zero. Therefore the system was reduced to force control mode (with reference of 0N). The subject ambulated with his normal

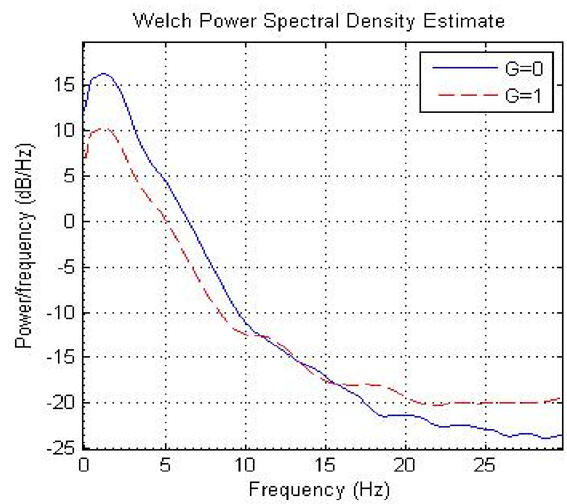

Fig. 11. Backdrivability test results (with and without negative unity gain force feedback). Approximately 50% (6dB) interaction force reduction for frequencies 0-6Hz can be seen.

gait. The force loop's proportional gain G was slowly increased from 0 to 1, and a gradual reduction in the interaction forces was observed. While the maximum interaction forces ranged between -20N and +20N without force feedback, the range of interaction forces was reduced to approx. +/- 10N with gain G=1. In order to better understand the effect of force loop gain on the system's backdrivability, the two data sets' power spectral densities were estimated. This analysis (Fig. 11) suggests that throughout the range of force signal frequencies of interest (0-6Hz), the existence of force feedback with proportional gain G=1 resulted in interaction force attenuation of about 50% (6dB).

D. Force Field Application Tests

The main task of the RGR Trainer is to apply corrective force-fields to the pelvic area in order to correct secondary gait deviation in the pelvis, namely hip-hiking. Therefore, tests were conducted with three healthy male subjects (Table

I) in order to investigate the system's ability to apply these force fields.

The total force - fields experienced applied to the subject's pelvis depend on both the robot end point impedance and environment admittance (inverse of impedance). Therefore, in order to investigate the effect of force field on pelvic obliquity, it became necessary to first measure the environment admittance. More specifically, the emphasis was placed on environment compliance (inverse of stiffness), because it is responsible for the bulk of forces applied to the body. In the future, tests investigating effects of damping on subjects displaying real secondary gait deviations could be performed.

The following test procedure was applied to each subject. The subject entered the RGR Trainer and donned the Newport 4 pelvic brace with thigh components. While the subject stood still, a LabVIEW program generated sinusoidal force commands at a frequency of 4Hz, which the Xenus amplifier executed using its internal current feedback loop when driving the servo tube actuator. The load cell provided force sensing. Three force amplitudes (peak to peak) were

TABLE I
HEALTHY MALE SUBJECTS TESTED IN RGR TRAINER

Subject	Height [cm]	Weight [kg]
1	165	80
2	165	70
3	180	90

used: 40N, 60N and 80N. The displacement resulting from the force application was used to find the environment's stiffness K_e, at the actuator level (kN/m) and at the obliquity level (N-m/deg) for each subject (Table II). This method can be improved on in the future to become a standard part of device calibration performed at the beginning of each session.

TABLE II
ENVIRONMENT STIFFNESS MEASUREMENT RESULTS
(SUBJECTS 1 – 3 RESPECTIVELY)

Force	K_e [kN/m]	K_e [N-m/deg]
40N	7, 9.7, 10	17, 23, 25
60N	5.4, 8, 8	13, 18, 18
80N	4.5, 6.5, 6	11, 16, 14

It was observed during the tests that the manner in which the brace was worn and tightened onto the subject's body had a significant impact on the stiffness value. Therefore, the force field application test was performed immediately following the environment stiffness measurement test using the stiffness measurement corresponding to that particular donning. The program was launched and the subject began ambulating on the treadmill at 1.4 km/h. Normal gait pelvic obliquity previously collected from another healthy individual (see section II.G) served as reference for all three subjects listed in Table I. Force loop proportional gain G = 1.0, damping ratio ζ = 0.3 and environment stiffness K_e as measured for each subject at 40N were used. The subject synchronized his gait with the reference trajectory and began to simulate hip-hiking. The desired force field strength value (K_d) was gradually increased to 6N-m/deg (same for all subjects), which resulted in a different impedance controller gain value K_c for each subject, since each subject exhibited unique environment stiffness value (Table II). Sample results of these tests (2 gait cycles) are presented in the figures below.

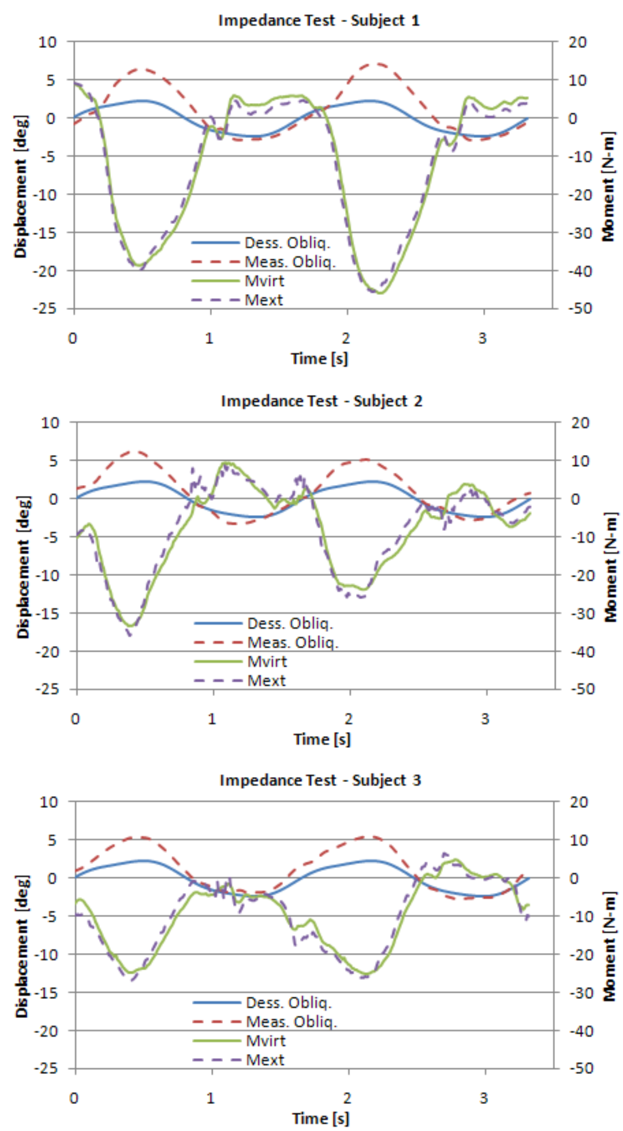

Fig. 12. Impedance controller test with the total force field strength set to 6 N-m/deg. The PD controller specified corrective moments (Mvirt) in response to the position error (Des.Obliq. – Meas. Obliq.). The inner force loop of the impedance controller executed this command rather accurately (Mext).

The “measured obliquity” curves do not represent the obliquity of the pelvis itself, but that of the attachment points of the pelvic brace, since the pelvic brace and the body’s soft tissue underwent considerable deformation under the exerted forces. Through about half of each cycle, a large discrepancy between the two curves exists, which corresponds to the subjects simulating hip-hiking. In response to the resulting position error, the position loop’s gains K_c and B_c produced what is referred to here as “virtual moment” M_{virt}. We can see that the peak of this corrective virtual moment coincides with the location of the maximum position error (Des.Obliq. – Meas. Obliq.), while diminishing down to near zero when the position error is close to non-existent. This had been demonstrated by bench tests performed earlier, while this test has shown how well the actuator system can display the commanded “virtual moment” under realistic conditions. The actual moment applied onto the body (M_{ext}) was computed from the interaction force measured by the load cell and the moment arm (37 cm for all subjects). We can see that measured moment M_{ext} follows rather closely the commanded virtual moment M_{virt}.

IV. Conclusions

The development of a robotic device, called Robotic Gait Rehabilitation (RGR) Trainer v.1, that generates force-fields to facilitate treadmill gait retraining in patients with abnormal gait patterns associated with exaggerated and uncoordinated (with other body segments) movements of the pelvis was presented in this paper. The end point impedance controller, which was implemented in the RGR Trainer, coupled with the highly backdrivable linear actuator, force feedback and minimal force transmission, produced appropriate dynamic behavior at the pelvis. Healthy subject testing has shown that the device imparts low resistance onto the subject during “free walking.” At the same time, the system accurately applies the prescribed force – field onto the pelvic area in response to hip-hiking.

Following this successful implementation of end point impedance controlled actuator to affect hip-hiking (pelvic obliquity), a new device – the RGR Trainer v.2 – is being designed with functionality expanded to pelvic rotation as well as partial body weight support. Following successful development of the complete RGR Trainer v.2, human studies will be performed with patients post-stroke, in order to assess the efficacy of rehabilitation of secondary gait deviations in the pelvis by means of force fields applied to it.

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