Assessment of Motion of a Swing Leg and Gait Rehabilitation With a Gravity Balancing Exoskeleton

Sunil K. Agrawal, Sai K. Banala, Abbas Fattah, Vivek Sangwan, Vijaya Krishnamoorthy, John P. Scholz, and Wei-Li Hsu

***Abstract*—The gravity balancing exoskeleton, designed at University of Delaware, Newark, consists of rigid links, joints and springs, which are adjustable to the geometry and inertia of the leg of a human subject wearing it. This passive exoskeleton does not use any motors but is designed to unload the human leg joints from the gravity load over its range-of-motion. The underlying principle of gravity balancing is to make the potential energy of the combined leg–machine system invariant with configuration of the leg. Additionally, parameters of the exoskeleton can be changed to achieve a prescribed level of gravity assistance, from 0% to 100%. The goal of the results reported in this paper is to provide preliminary quantitative assessment of the changes in kinematics and kinetics of the walking gait when a human subject wears such an exoskeleton. The data on kinematics and kinetics were collected on four healthy and three stroke patients who wore this exoskeleton. These data were computed from the joint encoders and interface torque sensors mounted on the exoskeleton. This exoskeleton was also recently used for a six-week training of a chronic stroke patient, where the gravity assistance was progressively reduced from 100% to 0%. The results show a significant improvement in gait of the stroke patient in terms of range-of-motion of the hip and knee, weight bearing on the hemiparetic leg, and speed of walking. Currently, training studies are underway to assess the long-term effects of such a device on gait rehabilitation of hemiparetic stroke patients.**

***Index Terms*—Gravity balancing, passive orthosis, rehabilitation, stroke patients, walking.**

I. Introduction

In THE past decade, robotics has been used to evaluate and treat upper extremity functions in individuals with severe motor impairment. MIT-MANUS was one of the first rehabilitation robots to undergo intensive clinical studies [1]. MIME (mirror image movement enhancer) used a PUMA 260 industrial robot to move a patient's arm in three-dimensions to match the motions of contra-lateral limb [2]. A mechanical assist and measurement device, called ARM Guide, was used to study differences in reaching between healthy and hemiparetic subjects following stroke [3]. The MIT-MANUS, MIME, and ARM Guide represent early advances in robotic devices for use in rehabilitation by exploiting the built-in neural plasticity in humans [4]. The embedded sensors make robots ideal for characterizing a movement and the actuators for modulating these movements. A second group of upper extremity machines is unmotorized or passive that uses springs in the design of the upper arm [5], [6]. Recent extensions of this work include development of a passive orthosis called T-WREX for the upper extremity movement training [7]. Although new rehabilitation devices and prototypes are emerging for the upper extremity, still, orthoses for the upper extremity are typically powered and involve planar designs with limited movement training.

Several lower extremity rehabilitation machines have been developed in the last five years for gait training during walking. These lower extremity rehabilitation machines are still not common in rehabilitation clinics. Walking is more difficult to train when compared to upper body functions due to issues of posture and balance, which may result in fall and injury to the patients.

Lokomat is a motorized exoskeleton and is designed to assist patients with movement disorders while walking on a treadmill. This machine has the degrees-of-freedom necessary to accommodate flexion and extension at the hip and the knee. It has motors to drive the joints and position sensors that record the leg trajectories [8]. Mechanized gait trainer (MGT) is a single degree-of-freedom powered machine that drives the leg in a specific pattern. The machine consists of a foot plate connected to a crank and rocker system [9]. AutoAmbulator is a rehabilitation machine developed by HealthSouth to assist individuals with stroke and spinal cord injury exhibiting leg impairments. The cost of these machines is relatively high, which makes these less accessible to many patient care facilities. With these machines, safety is also an issue. Additionally, a user needs to be monitored by clinical and engineering personnel, which further increases the cost of the treatment.

A feature of early developments of these machines is that they move patients through predetermined movement patterns rather than allowing the patients to move under their own neuromotor control. They failed to provide patients the tools necessary to learn and practice appropriate movement patterns. More recent modifications of these machines are subject centric [10]. In recent years, lower extremity assist devices have been characterized as exoskeletons. HAL [11] is a powered suit for elderly which takes electromyography (EMG) signals as inputs and produces appropriate torques to perform the task. BLEEX [12] is intended as a human strength multiplier. An active limb

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exoskeleton (ALEX), developed at the University of Delaware, Newark, is a motorized exoskeleton which uses motors at the hip and knee joints and an innovative force field at the foot that helps in gait training [13].

The objective of this series of preliminary case studies is to develop and evaluate a low-cost, nonmotorized, orthoses for gait training of subjects with stroke and other neuromotor disorders. We believe that low-cost rehabilitation devices, which do not require motors or control system, can be made more accessible to subjects for treatment. We have fabricated a lower extremity unmotorized exoskeleton, called the gravity balancing orthosis (GBO), for gait training that can alter the level of gravity assistance on the thigh and shank segments of the human leg [14]. While this referenced paper [14] focuses on the mechanics, design, and initial testing with the GBO, the current paper will focus on 1) interpretation of kinematic and kinetic data obtained by the GBO and 2) our findings from a six-week training study with the GBO on a chronic stroke patient.

Gravity balancing has been used to reduce the actuator effort in machines during motion, through clever use of counter weights [15], and springs [16], [17], by making the system potential energy to be constant. Some recent sit-to-stand assist devices have been proposed using counter weights [19]. A primary limitation of design procedures using only springs to gravity balance a system is that the system loses its gravity balancing property if it changes orientation with respect to the gravity vector [20]. In order to circumvent this difficulty, the GBO design first locates the center of mass of the system using auxiliary parallelograms. The springs are then added through the center of mass and other locations such that the total potential energy of the system becomes invariant with configuration. This procedure ensures gravity balancing during the abduction motion of the hip, i.e., the leg has an out-of-plane motion with respect to the gravity vector [20]. In addition, parameters in the design are adjustable to achieve fractional balancing of the segments.

Gravity plays an important role in human movement. The torques at the joints of a leg during a typical human movement can be broken down into inertia torque, gravity torque, and muscle compliance torque. As the names suggest, the inertia torque is needed to sustain a desired joint speed and joint acceleration of the moving leg. The gravity torque is required to keep the limbs in a specific configuration under gravity. The muscular compliance torque is needed to move the limbs against passive elasticity at the joints. As one would intuitively expect, at fast gait speeds, the inertia torque dominates over the other components of the joint torque. At slow gait speeds, the gravity torque is more dominant [14]. The muscle compliance torque is higher towards the extremes of the range-of-motion of the joint. Intuitively, as one would expect, a person with a weak neuromotor control or reduced muscle strength may find it hard to swing a leg against gravity. However, the same subject may find it easier to lift and swing the leg if the gravity was reduced at the joints.

It is important to compare and contrast the training with GBO with the body weight-supported treadmill training (BWSTT), a training paradigm which is becoming popular in the literature. The fundamental behind BWSTT is that impairments after stroke limit full weight bearing on the hemiparetic leg. BWSTT is a training program that is designed to shape the patient's ability to bear weight on that limb by partial body weight support from an external equipment. This partial body weight support is gradually reduced with improvements in performance [21], [22]. BWSTT has the advantage of allowing therapists to initiate gait training at a very early stage of stroke recovery when neural plasticity can best be harnessed. This training also helps to reduce the labor-intensive effort required by standard clinical gait training.

It is important to note that during BWSTT, the gravity still acts on the moving limbs, while the weight of the upper body borne by the leg is partially reduced. In contrast, GBO training is not designed to reduce weight bearing on the hemiparetic leg. The GBO uses a clever mechanical design using springs that can partially eliminate the gravity torque at the joints of the hemiparetic leg as it swings during motion. In addition, the gravity assistance to the swinging leg can be lowered as the patient gains better control over the movement.

We believe that lower extremity exoskeletons, that can unload the human joints from gravity either partially or fully, can provide new paradigms for movement training and new insights into human movement. Here are some logical questions that relate to training with the GBO.

1) How will the kinematics and kinetics of the joints of a leg get altered if the gravity is reduced during walking?
2) What role can reduced gravity play in gait rehabilitation of stroke patients?

Our goal in this paper is to answer these questions using qualitative analysis and data collected from studies with human subjects.

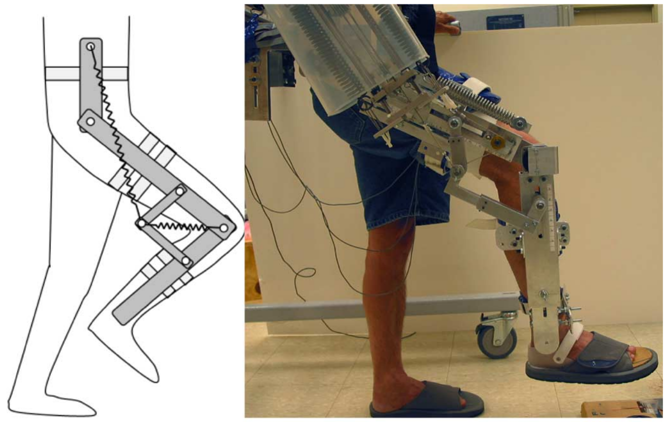
The remainder of this paper is organized as follows. Section II describes the features of the exoskeleton along with a short summary of how it is used by a patient. Section III presents the effects of the gravity balancing exoskeleton on kinematics and kinetics of a swinging leg for healthy and stroke patients. Details of the training protocol and results of a six-week training study of a stroke patient are described in Section IV. These are followed by key conclusions and discussions in Section V.

II. Features of the GBO

The passive exoskeleton, GBO, is designed to partially or fully unload the hip and knee joints from the gravity throughout the range-of-motion of the leg. The underlying principle of full gravity balancing is to make the potential energy of the combined leg–machine system invariant with configuration of the leg [14]. A schematic of the design and a close-up view of the fabricated exoskeleton is shown in Fig. 1. A photograph of this device, worn by a stroke subject on a treadmill, is shown in Fig. 2.

The features of GBO, along with a short summary of its use, are as follows.

1) A backpack, with metal backing and straps, is connected to a walker frame that has three degrees-of-freedom with respect to it—two translations (sideways and vertically up and down), and one rotation (about a vertical axis), all with respect to the walker.
2) The exoskeleton itself has three planar degrees-of-freedom to accommodate sagittal plane motion of the human leg. Its limbs are machined out of lightweight aluminum and



(a) (b)

Fig. 1. (a) Schematic of the gravity balancing exoskeleton. (b) Physical prototype of the exoskeleton from a closed view.

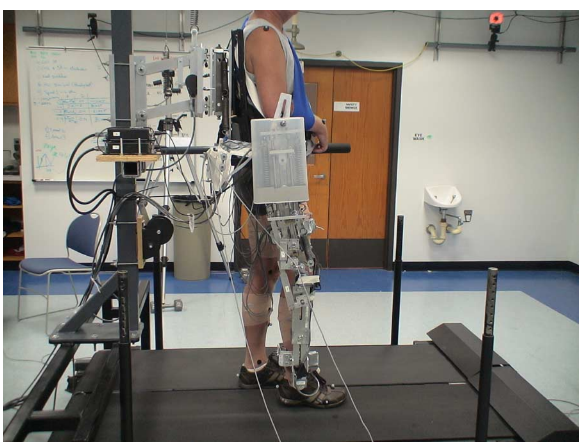


Fig. 2. View of a stroke patient wearing the exoskeleton on a treadmill instrumented with force-plates.

have straps that snugly wrap around the human limb segments. The thigh and shank segments of the machine are telescopic to accommodate variability in the leg geometry of the human subject wearing it. A parallelogram mechanism with springs is designed to accommodate variability in inertia of a subject and achieve a gravity setting, between zero and one gravity. The foot segment of this exoskeleton is a shoe insert.

3) The exoskeleton is connected to the metal backing of the backpack with an additional degree-of-freedom to accommodate hip abduction/adduction.

The GBO is used by a subject in the following way.

1) A subject straps on the backpack and the segments of the exoskeleton are strapped around the thigh and shank segments of the human leg. The subject wears the shoe along with the shoe insert, which is then connected to the upper part of the exoskeleton.
2) The telescopic links of the thigh and shank segments of the exoskeleton are adjusted to conform to the user. The geometry of the links of the gravity balancing mechanism and the spring locations are adjusted to achieve a gravity setting, between zero and one gravity.
3) The exoskeleton is fitted with seven joint encoders that record the movement of the trunk with respect to the walker, the abduction motion at the hip, and the sagittal plane joint movements of the leg. In addition, the exoskeleton has two six-axis force-torque sensors that record the interaction force and torque between the machine thigh and shank segments and the corresponding segments on the human.
4) The walker is positioned over a treadmill and the human subject walks at different settings of the treadmill.

Our first prototype used bronze bearings at the joints of the auxiliary parallelogram but the friction was excessive during experiments. In the next design iteration, these bronze bearings were replaced by double-row ball bearings, which reduced the friction significantly. In some experiments, a six-camera motion analysis system was used to track the movement of the leg. In some earlier testing, EMG was recorded from major muscle groups of the leg. During some of the training sessions, the kinematic trajectory of the foot and the joint angles, as recorded from the joint encoders, were provided to the subject as visual feedback superimposed on a desired foot/joint trajectory.

A. *Experimental Determination of Joint Torque*

Joint encoders manufactured by USDigital, 2500 counts per revolution, were used to record data at the hip, knee, and other degrees-of-freedom in the exoskeleton. Two ATI force-torque sensors were used to collect the interface forces/moments between the human leg and the exoskeleton during treadmill walking. A dSpace data acquisition and control system, with a 1-kHz sampling rate, was used to collect the joint motion and force data. For modeling the dynamics of the hemiparetic swinging leg with the exoskeleton, we made the following assumptions: 1) the trunk is inertially fixed; 2) motion of the leg is in the sagittal plane; 3) the thigh and shank segments of the human limb are rigidly attached to the corresponding segments of the machine; 4) the foot is a point mass at the end of the shank segment. These assumptions are reasonable because the trunk motion is quite small compared to the motion of the thigh and shank segments and the hip abduction angle is small. Please note that this model can be improved in future to include hip abduction to better model the movement of a stroke gait. A schematic for such a model for the human leg with the exoskeleton is shown in Fig. 3(a).

Fig. 3(b) shows the free body diagram of the human thigh and shank segments. Using data from interface force-torque sensors between the human and machine limbs and joint data using encoders, Newton–Euler equations can be used to compute the torque applied by the subject at the hip and the knee joints. Given the overall height and weight of a human subject, geometric and inertia parameters of the human leg are estimated using the anthropometric data [24]. The geometric and inertia parameters of the machine segments are measured. In these computations, joint angular velocity and accelerations are derived computationally using the joint angle data measured by the joint encoders. There are different methods to approximate velocity and acceleration numerically. In order to ensure robustness, we curve fit a higher degree polynomial on

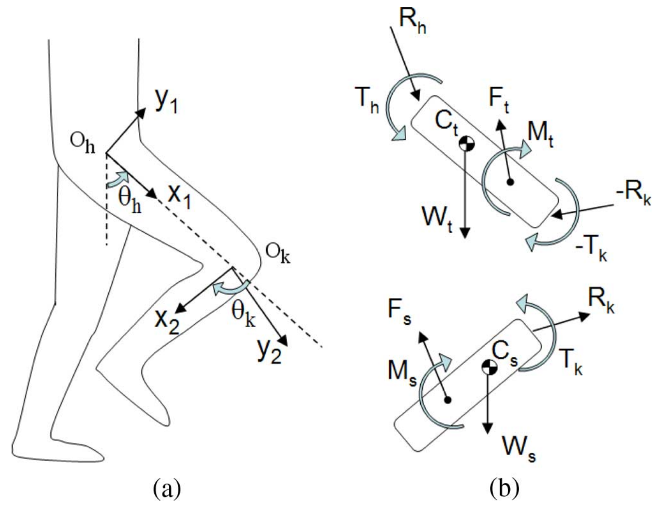


Fig. 3. (a) Two degree-of-freedom planar model of the human leg and exoskeleton. (b) Free body diagram of the human leg segments.

the joint angle data using least squares. This curve was then differentiated to obtain the joint velocity and acceleration.

The vectorial force balance equations were written for the human thigh and shank segments using Newton's laws involving the accelerations of their respective center of masses. Similarly, the rotational equations were written for the thigh and shank segments involving their rotation accelerations. This procedure results in six scalar equations involving six scalar unknowns from the quantities $\mathbf{R}_h$, $\mathbf{T}_h$, $\mathbf{R}_k$, and $\mathbf{T}_k$. The quantities $\mathbf{F}_t$, $\mathbf{F}_s$ consisting of two scalar components and M_t, M_s consisting of one scalar componet are measured using the interface force-torque sensors. More details of this procedure are available in [14].

III. Effects of the GBO on Gait Kinematics and Kinetics

For normal walking of healthy subjects, data on gait kinematics and kinetics is available in the literature [23], [24]. The qualitative behavior of joint torque during normal walking is also well documented. In this section, our goal is to understand how the joint trajectory and joint torques will change for the swinging leg as a result of reduced gravity through the exoskeleton.

The swing phase of a leg is the duration for which the foot is in the air. This duration, between the toe-off and foot strike, constitutes roughly 35% of a complete cycle of the gait. With the GBO, our goal is to analyze the swing phase of the leg in the exoskeleton. Hence, in this paper, we rescale this swing phase between 0% and 100%. In the experiments, the start and completion of the swing are recorded using foot switches mounted on the heel and toe of the subject's foot.

During normal walking, using the kinematic convention of Fig. 3, the hip swings forward from its fully extended position, roughly $-20°$, to the fully flexed position, roughly $+45°$. The knee starts out somewhat flexed at toe-off, roughly 10°, continues to flex to about $+45°$ and then straightens out close to 0° at touchdown. As one would expect, during normal walking, the beginning of the swing is characterized by flexion moment at the hip, positive torque in our convention, which helps the leg to swing forward. At the terminal swing, just before heel contact, an extension moment slows the hip flexion, negative torque, and prepares the leg for weight bearing.

Qualitatively, the knee acts as a damper throughout the motion. During the beginning phases of the swing, an extension moment, negative torque, acts on the knee to slow it down to its peak flexion value. From this peak flexed position, the gravity and the coupling inertia torque from the hip swing it forward. As the knee joint picks up speed, a flexion torque, positive torque, is required to slow down its motion before heel contact. A typical joint trajectory of the hip and the knee are shown in Fig. 4. The corresponding joint torques to sustain this motion computed using the dynamic model of a two degree-of-freedom leg, with anthropomorphic data of an average human is also shown in this figure.

In the section that follows, we provide a biomechanical analysis of the predicted effect of gravity balancing on limb motion during walking and then provide some preliminary data to show the effects of gravity balancing with the GBO on the gait pattern of healthy subjects and a few persons recovering from a chronic stroke.

We will now ask the following questions.

1) How will the joint trajectory of the leg change under reduced gravity conditions, because of the assistance provided by the exoskeleton, assuming that the human still applies a similar profile of joint torques at the hip and the knee?
2) Could this reduced gravity create an environemnt for human–motor gait training, i.e., altering the gait kinematics or applied joint torques?
3) Are these hypotheses consistent with subject data collected experimentally?

A. Alteration of Gait Kinematics

1) Single Degree-of-Freedom Model: Before going deeper into the details of a two degree-of-freedom pendulum model of a human leg, let us understand these ideas more intuitively by modeling the leg as a single degree-of-freedom pendulum. This one degree-of-freedom model is sketched in Fig. 5 and is representative of the forward swing of the hip. Mathematically, the equation governing this motion is given by

$$\bar{T} = \ddot{\theta} + \frac{mgl*}{I}\sin\theta \tag{1}$$

where $\bar{T} = T/I$, T is the applied joint torque, I is the moment of inertia about the joint axis, m is the mass of the pendulum, $l*$ is the distance of the center of mass from the joint axis, g is the gravitational constant, and θ is the joint angle. We prescribe a very simple motion to this pendulum as it swings forward, a rest-to-rest maneuver from $-20°$ to $40°$, within a specified duration of time $T_p/2$, where T_p is the time period of a cycle, as shown in Fig. 5(a).

Let us concentrate on the forward motion of the pendulum from $-20°$ to $40°$. During the first half of this forward swing motion, the pendulum moves with a constant positive acceleration while in the second half of this forward swing motion, the

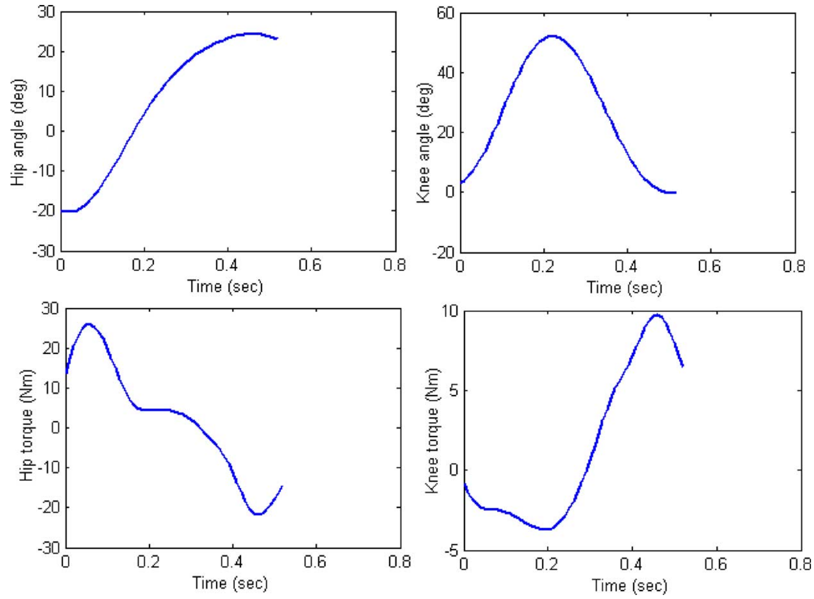


Fig. 4. Hip and knee joint trajectory during swing phase of the leg. Hip and knee joint torques computed using a dynamic model, with the given kinematic data, and anthropomorphic subject parameters. Computed joint torques match well with the torque computed experimentally, using interface force-torque sensors.

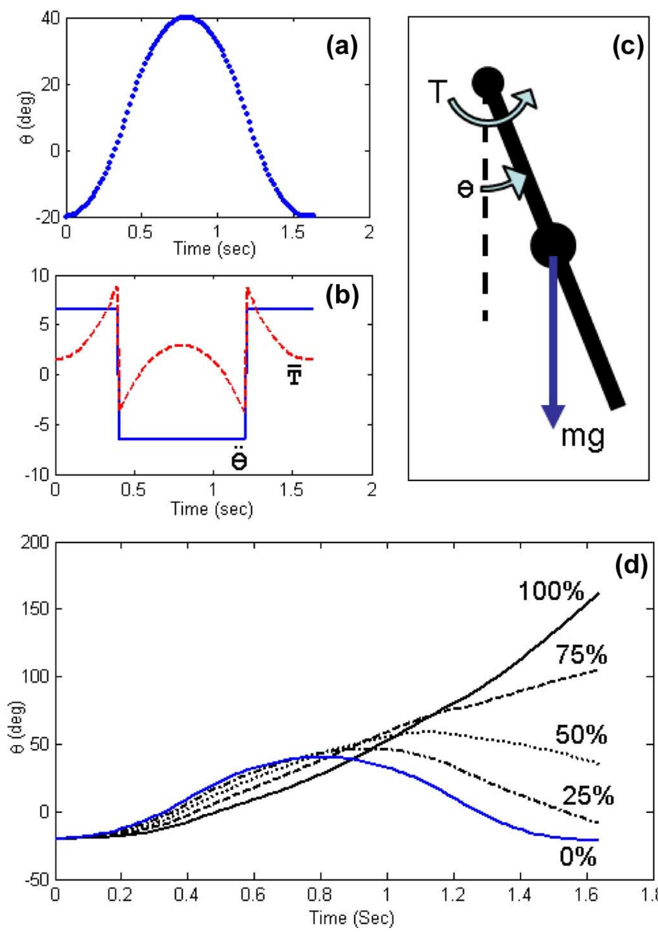


Fig. 5. Effect of reduced gravity (or gravity assistance) on the motion of a single degree-of-freedom pendulum model of a human leg. The joint torque computed from a typical movement profile is applied to the model with reduced gravity. We observe that the range-of-motion of the joint increases as the gravity assistance is increased from 0% to 100%.

pendulum has a constant negative acceleration (shown by the blue lines). As the pendulum swings from $-20°$ to $0°$, since the gravity torque is in the direction of the flexion torque, $\bar{T}$ is lower than $\ddot{\theta}$, as shown in Fig. 5(b), by the red lines. During the range $0°$ to $10°$, since the gravity torque opposes the applied torque $\bar{T}$, it is higher than $\ddot{\theta}$. During the range $10°$ to $40°$, since it is a deceleration phase and gravity is helping in the deceleration, the applied torque $\bar{T}$ is negative but higher than $\ddot{\theta}$. These plots are shown in Fig. 5(b), using blue and red lines.

We now ask the following question: What difference do we expect in the motion of this pendulum if the joint torque shown in Fig. 5(b) was applied to the pendulum from the same initial condition but under a reduced gravity condition? From the logic that we just used in the last paragraph, we can reason out that as the gravity assistance is increased from 0% to 100%, the range-of-motion of the pendulum will increase. During the range $0°$ to $10°$, there is a higher flexion torque and between $10°$ to $40°$, a lower extension torque. The motion of joint angle θ is shown in Fig. 5(d) for different settings of the gravity assistance. In these simulations, the parameters m, $l*$, and I have been chosen according to anthropormphic data of a thigh segment along with a swing time of 0.8 s. From these simulations, we clearly see that the range-of-motion of the joint increases as the gravity assistance is increased from 0% to 100%. Note that the simulaion in Fig. 5(d) for the setting of 0% is the same as the data shown in Fig. 5(a).

2) Two Degree-of-Freedom Model: In order to show the effects of gravity assistance on the range-of-motion on a two degree-of-freedom model of a human leg, we used the hip and knee joint motion and torque data, shown in Fig. 4, and forward simulated the dynamic model of a swinging leg with reduced gravity to predict the joint motions of the hip and knee joints. These results for five values of g between 1 and 0, i.e., 0% gravity assistance to 100% gravity assistance, are shown in Fig. 6. In these simulations, all geometric and inertial parameters were taken from anthropomorpic data of average human. In the simulation, we added damping at the knee 2 Nms/rad, to stabilize the joint

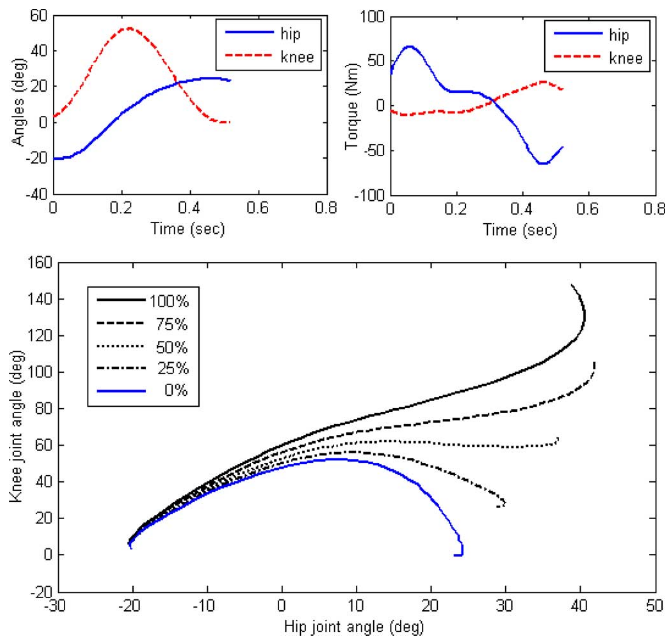


Fig. 6. Effect of reduced gravity or gravity assistance, on the motion of a swinging human leg. The joint torques computed from a typical movement of the leg is used to compute the range-of-motion of the leg under reduced gravity. We observe that the range-of-motion of the two joints increases as the gravity assistance is increased from 0% to 100%.

TABLE I
DEMOGRAPHICS, GEOMETRIC AND INERTIA PARAMETERS OF SUBJECTS

Group	Initials	Gender	Height (m)	Weight (kg)	preferred speed (m/s)
Healthy	VS	male	1.78	81.2	0.71
Healthy	VK	female	1.61	56.69	0.76
Healthy	JR	male	1.71	58.97	0.49
Healthy	PH	male	1.90	72.57	0.71
Stroke	ET	female	1.65	83.5	0.23
Stroke	FS	male	1.82	81.6	0.23
Stroke	DG	male	1.75	68	0.45

motions. As expected, the range-of-motion of the two joints increases as the gravity assistance is increased from 0% to 100%.

B. Data From Healthy and Stroke Subjects

The first data, presented here, was collected from four healthy young adults and three subjects with right hemiparesis, following a stroke. Table I provides details of demographics, geometry, and inertia parameters of these subjects involved in the experiments. Subjects gave informed consent according to procedures approved by the institutional review board of the University of Delaware. A stroke subject walked at the preferred speed, while the healthy subjects walked at 30% and 60% of their preferred speeds to make the speeds comparable to those of the stroke subjects. Five trials of walking were collected and the time duration of each trial was about 30 s. Walking task was conducted within the device with the following settings: 1) leg and device fully gravity balanced, refered to as "full-balanced" condition; 2) device only is gravity balanced, refered to as "device balanced" condition. It is important to note that all these subjects were first time users of this machine and had not received any short-term or long-term gait training prior to the data collection.

TABLE II
PEAK KNEE AND HIP FLEXION EXCURSIONS DURING SWING PHASE WHEN WALKING IN GBO WITH "DEVICE ONLY GRAVITY BALANCED" AND WITH "LEG AND DEVICE (FULL) GRAVITY BALANCED" FOR THREE STROKE SUBJECTS WALKING AT THEIR PREFERRED SPEED AND FOUR HEALTHY SUBJECTS WALKING AT 30% OF THEIR PREFERRED SPEED, COMPARABLE TO THE SPEED OF THE STROKE SUBJECTS

ID	Device-balanced Peak Knee Flexion	Full-balance Peak Knee Flexion	Percent Increase(+) Decrease(-)	Device-balanced Peak Hip Flexion	Full-balance Peak Hip Flexion	Percent Increase(+) Decrease(-)
Stroke Subjects, (100%) preferred speed						
DG	17.002 ± 1.785	34.451 ± 2.436	102.6	24.279 ± 2.044	38.161 ± 2.692	57.2
ET	26.976 ± 2.202	46.750 ± 4.437	73.3	17.748 ± 1.449	46.369 ± 3.466	161.3
FS	23.508 ± 1.890	63.337 ± 2.057	169.4	18.551 ± 1.670	60.553 ± 3.617	226.4
Healthy Subjects, 30% preferred speed						
JR	39.791 ± 2.074	45.038 ± 3.378	13.2	8.907 ± 1.078	8.841 ± 2.474	-0.7
PH	40.111 ± 3.633	47.085 ± 5.089	17.4	18.369 ± 1.625	27.476 ± 1.968	49.6
VK	58.183 ± 5.072	78.041 ± 2.636	34.1	30.261 ± 1.946	49.437 ± 3.060	63.4
VS	51.368 ± 1.609	63.576 ± 3.976	23.8	20.458 ± 2.091	32.428 ± 1.982	58.5

Fig. 7 shows the plots of the hip joint angle versus the knee joint angle during entire gait for a representative healthy and stroke subject performing walking task. It is clear from these plots from healthy and stroke subjects that for the "full balanced" condition, the range-of-motion at both hip and knee joints is larger than with "device balanced" condition. All three stroke subjects exhibited a greater than 50% increase in the peak hip and knee flexion obtained during swing phase when the device was fully gravity-balanced compared to the device-balanced condition. In all cases for the stroke subjects, the improvement was well above 50% (Table II). The healthy subjects showed similar results overall although of smaller magnitude when walking at speeds well below their preferred speed, but comparable to those of the stroke subjects. This increase in peak joint flexion during swing phase is expected according to our analysis shown in Sections III-A. Representative videos of these motions can be seen at our website[1] on the link "medical robotics."

C. Paradigm for Neuromotor Gait Training

A schematic of the training paradigm with the GBO is shown in Fig. 8. This training paradigm exploits the following two features of the GBO: 1) ability to change the gravity assistance; 2) ability to extend the achievable range-of-motion of the hip and knee joints of the hemiparetic leg. During the training sessions, we expect to present a template of the "hip and knee" joint angles or "foot position" in Cartesian frame which is executable within the range-of-motion of the joints/foot for the given level of gravity assistance. Using visual feedback of the joint/foot trajectory, the subject will try to match the given template. It is important to point out that a given gait template can be achieved by many choices of joint torques. Human motor learning has an

[1] http://www.mechsys4.me.udel.edu/research

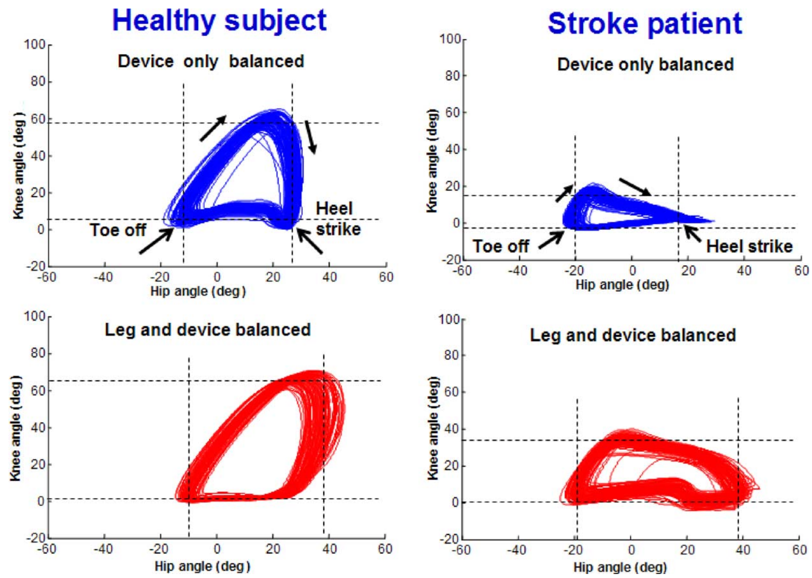


Fig. 7. Plot of knee versus hip joint angles during treadmill walking with the GBO during "device-balanced" and "full-balanced" conditions for a representative healthy subject and a stroke patient.

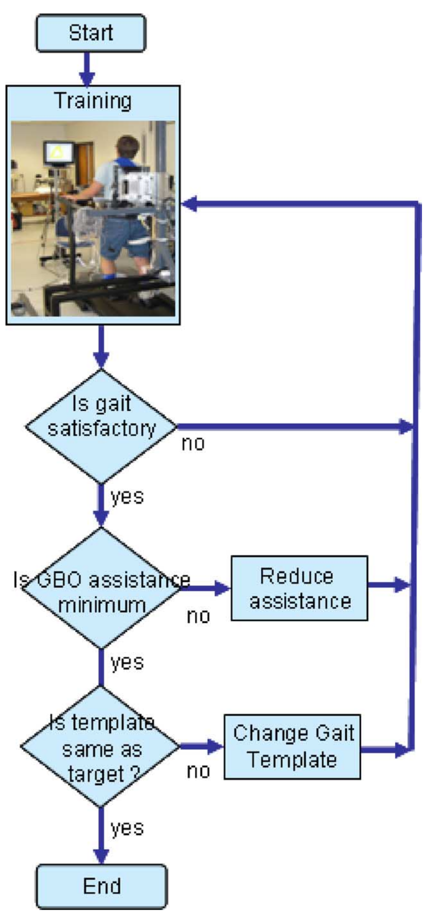


Fig. 8. A schematic of the training paradigm with the GBO, designed around its feature to extend the useful range-of-motion of the joints of the leg. A subject is presented with a gait template for a given level of gravity assistance. As a subject becomes more proficient with matching this template, the gravity assistance is decreased or the template is made more challenging.

important role to play here. According to the current ability, the subject will choose the best combination of joint torque inputs to match the given gait template. In the training paradigm, for a given level of gravity assistance, if the subject is able to match the gait template well, either the amount of assistance is lowered or the gait template is made more difficult.

IV. Training Study of a Stroke Subject With GBO

A chronic stroke survivor (three-year past stroke), 56 year old male, with right hemiparesis volunteered for more intensive training with the GBO to determine long-term training effects. Formal gait evaluations were performed during both over-ground and treadmill walking using a six-camera motion analysis system, force plates, and electromyography. An additional session was performed on a separate day to familiarize the patient with walking in the GBO. A walking evaluation was also performed in the GBO. Treadmill and over ground walking evaluations were repeated midway through the training and after the final training session. The over-ground walking evaluation was also repeated four weeks following the last training session. In all, there were a total of 15 training sessions which lasted six weeks.

In all training sessions, the patient received explicit feedback in the form of knowledge of performance while walking in the GBO. Training sessions lasted for about 2.5 h. Much of this time was rest time while the device parameters were adjusted and the patient received summary feedback. Gravity assistance began at 100% and was gradually reduced over sessions to 0%. Functional electric stimulation (FES) was confined to the ankle dorsiflexors, while the gravity assistance was provided at the hip and the knee joints. Following the six initial training sessions with 100% gravity assistance, there were nine additional training sessions with reduced level of gravity assistance, progressively going down to 0%. Each training session consisted of four blocks, 10 min each, of treadmill walking, each followed by 5 min of rest, or longer if requested. The subject began training at his preferred treadmill walking speed of 2.5 Km/h. This speed was gradually increased throughout the training as

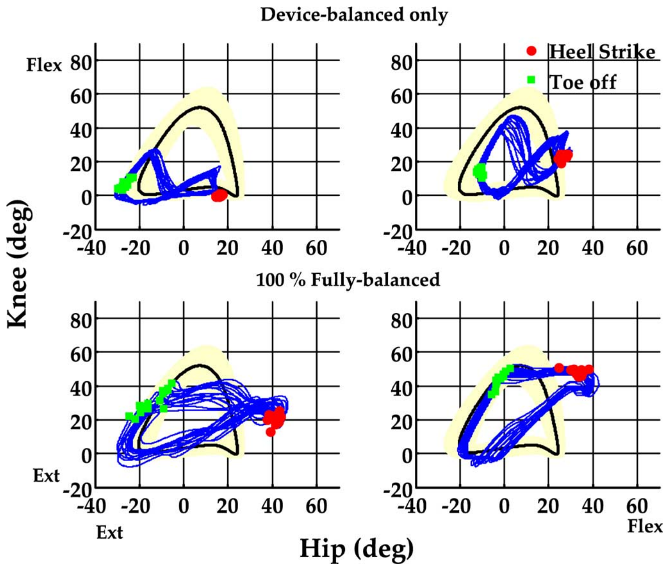


Fig. 9. Hip-Knee angle plots in GBO without gravity assistance (top) for day 1 of the training (left) and day 15 of the training (right), respectively, and 100% gravity assistance (Bottom) for day 1 of training(left) and day 3 of training (right).

he became more comfortable, the exact speed chosen by the subject. The training speed increased to 2.72 Km/h by mid-training and 3.04 Km/h by the final evaluation. In addition, the subject's preferred over ground gait speed showed a small improvement from 3.38 Km/h to 3.86 Km/h from the pretraining to posttraining evaluation.

Feedback of the joint/foot motion was provided to the subject visually in two modes: continuous or intermittent (30 s on; 30 s off). Summary knowledge of performance was also given after each 10 min training block. A video monitor was positioned immediately in front of the treadmill at the patient's eye-height to provide the online feedback. The nature of the feedback could also be changed: sagittal plane foot trajectory or angle-angle plot trajectory of the hip and knee. A template of the average trajectory of normal subjects, collected in the GBO, was shown to the patient on the video monitor. If the displayed trajectory was of the foot, it was computed from the normal subjects' average angle trajectories and scaled to the patient's segment lengths so that it was appropriate for this patient. The patient's current sagittal plane foot or hip–knee angle trajectory was also displayed. The subject was told to adjust his foot trajectory (hip–knee angles) to match the template. This subject performed best when intermittent feedback was given about the hip–knee angle, which was then given most frequently.

The top panel in Fig. 9 provides an example of the patient's knee–hip trajectories obtained from the initial walking evaluation (left) and from the last training session (session 15; right) using the GBO with device-only balancing. The patient's knee excursion during the swing phase showed impressive increases over the training sessions although he still maintained the double-loop pattern, indicating an early knee extension followed by additional knee flexion prior to heel strike. Unfortunately, improvements in knee flexion during swing were accompanied by a less desirable decrease in terminal hip extension during stance. The bottom panel in Fig. 9 illustrates changes of hip–knee coordination in the fully-balanced condition (100% assistance) between sessions 1 and 3. Note that, in

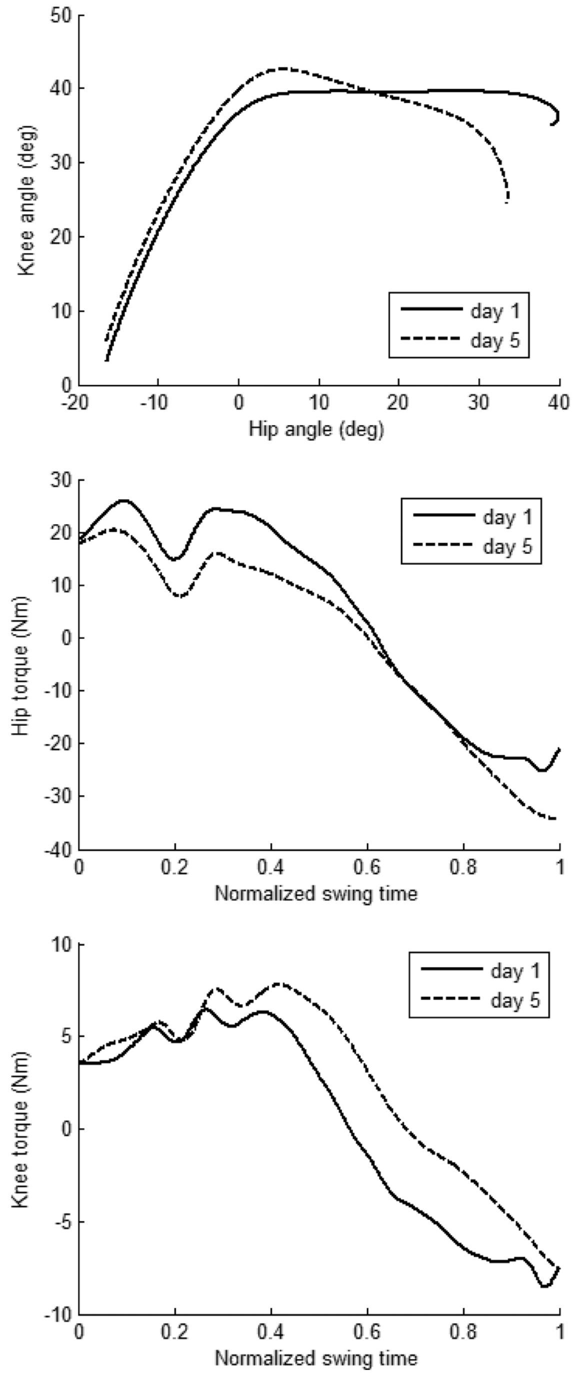


Fig. 10. Hip–knee angle/torque plots during the swing phase in the GBO for 100% gravity assistance: days 1 and 5 (top), hip torque days 1 and 5 (middle), knee torque days 1 and 5, respectively (bottom).

contrast to device-only balancing (top panel), the patient was able to perform a single loop pattern of hip–knee coordination that was more like the normal pattern when the leg was fully gravity balanced. Moreover, within three sessions of practice, the pattern became more normal looking during the first half of the cycle following toe-off.

Fig. 10(a) shows the hip and knee joint data during the swing phase, averaged from the last session on days 1 and 5 (each with 100% assistance). This kinematic data was used to find the hip and knee joint torques using the interface force sensors and the dynamic model, which are shown respectively in the middle

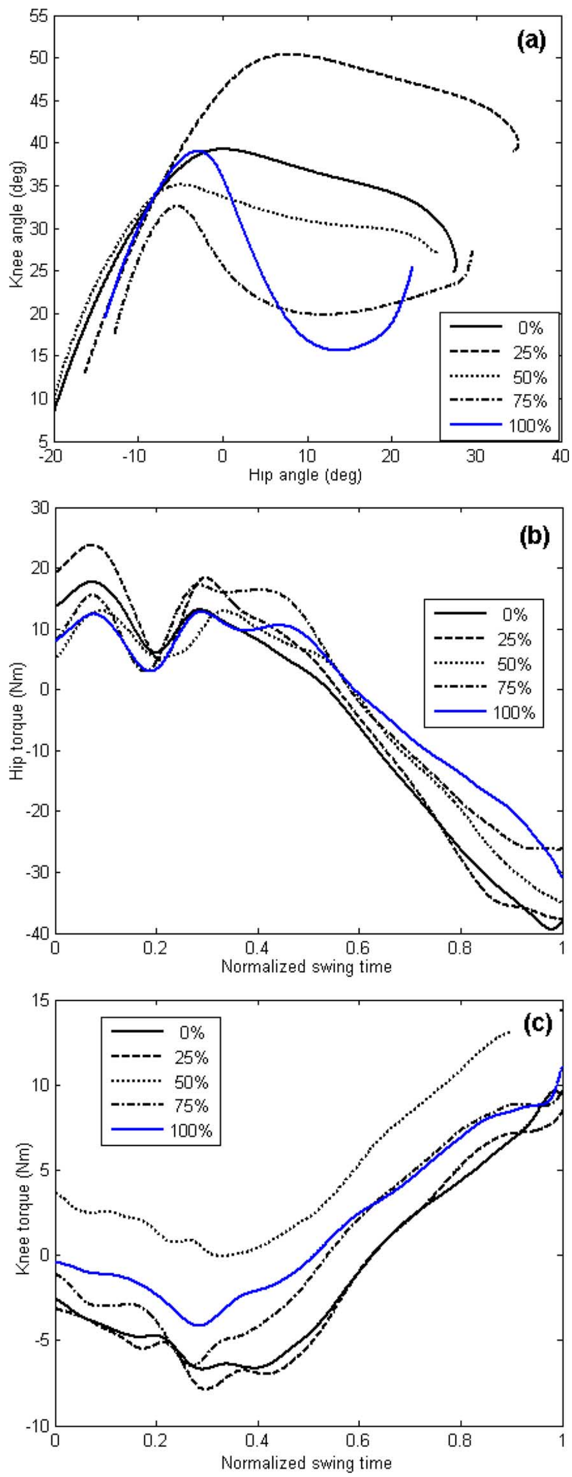


Fig. 11. Hip–knee angle/torque plots during the swing phase in GBO within the 15 day trial as the gravity assistance was progressively decreased from 100% to 0%: Gait kinematics (top), hip torque (middle), knee torque (bottom). The data shows motor learning where increased range-of-motion of the leg was achieved with decrease in the joint torque during the beginning of the swing.

and the bottom panels. It is worthwhile to point out the motor-learning aspects within the first five days of training. During all these five days, the gravity assistance was at 100%. On day 1, we see that at toe-off, the knee is flexed to about 5°. As the hip swings forward, the knee flexes quickly to 35° and stays in this flexed position until the hip is in the fully flexed position. As the therapy progresses, on day 5, the subject is able to straighten the knee towards the end of the swing phase, closer to normal walking. In addition, over time, the subject is able to attain a more natural swing pattern while still being able to decrease the applied hip joint torque during the beginning of the swing. These features are attributed to learning, which was facilitated through the GBO.

Fig. 11 shows the swing kinematic data over the 15 days of training, with varying levels of gravity assistance from (100%–0%). The middle and bottom panels show the variation of the hip and knee joint torques for the subject over the 15 days of training, as the level of gravity assistance was decreased from 100% to 0%. These data correspond to the joint kinematics data shown in top panel of this figure. As shown, the range-of-motion gradually increases with training. Additionally, with human motor learning, this increase of range-of-motion was accompanied with a decrease of applied hip torque in the initial period of the swing.

Following the 15 training sessions with the GBO, during which the subject was progressed from 100% to 0% gravity assistance, a number of improvements in performance on both clinical tests and walking evaluations were observed. Improvements of some parameters that were measured during over ground gait analyses are as follows: knee flexion excursion during swing improved from the initial evaluation to the midtraining evaluation, decreased somewhat at the first posttraining evaluation, but then increased again by the four week posttraining evaluation, as did hip flexion. In addition, the hip was slightly more extended at terminal stance during over ground walking compared to the first evaluation. The ankle was less plantar flexed at heel strike, indicating improvement in ankle dorsiflexion. The total path length of the foot increased somewhat during swing. This was primarily due to better foot clearance during swing rather than an increased right step length. These were encouraging findings that indicated continued improvement after training.

Currently, a more complete set of training studies are planned with the GBO involving new stroke patients. Due to restricted patient availability, in this study, the training from 25% balancing had to be quickly dropped down to 0%. Our group feels that training at 25% gravity balancing should have been maintained for several additional sessions. Perhaps, we will see even better training effects in future if the gravity assistance is changed less rapidly. Further improvements in the device and training protocol are certainly needed and in progress.

V. Conclusion

This paper described the details of a GBO, that has been fabricated at the University of Delaware, which can be adjusted to alter the level of gravity assistance on the joints of a swinging leg during walking. Tests have been conducted to evaluate the effectiveness of this device for gait training of stroke and other neuro-impaired subjects. The central goal of this paper was to address the following two fundamental questions.

1) How will the kinematics and kinetics of the joints of a leg get altered if the gravity is reduced during walking through the GBO?
2) What role can reduced gravity play in gait rehabilitation of stroke patients?

Answer to these questions were sought using quantitative analysis and data collected from human subjects. Our analysis of the first question concluded that gravity assistance will increase the range-of-motion of the joints of the leg. The kinematic data collected on healthy and stroke subjects, while walking on a treadmill, revealed substantial increases in the range-of-motion of the hip and the knee joints. This increase was more pronounced for the stroke subjects, more than nearly 50% at the hip and the knee. This increase in the range motion was justified using models of single and two degree-of-freedom model of a swinging leg. The answer to the second question is directly based on the increase in the range-of-motion at the joints, which was used to design a paradigm for gait training of stroke patients. In this paradigm, a gait template was displayed to a stroke patient on a computer screen and feedback of his gait was provided in real-time, as the patient tried to match his gait to this template. The level of gravity assistance was dropped as the patient learnt to match his gait with this given template. A six-week training study was conducted with a stroke patient where the gravity assistance was progressively decreased from 100% to 0%. A number of important gait improvements were observed with this patient such as increase in the knee and hip flexion. In addition, the hip was slightly more extended at terminal stance during over ground walking compared to the first evaluation. The subject was able to improve his walking speed during the six-week training, increase weight bearing on the hemiparetic leg, and was more symmetric in his walk. Currently, further training studies are planned with the GBO involving new stroke patients.

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