# Novel Gait Adaptation and Neuromotor Training Results Using an Active Leg Exoskeleton

Sai K. Banala, Sunil K. Agrawal, Seok Hun Kim, and John P. Scholz

***Abstract*—The gait of every adult is unique and expected to be ingrained within the neuromuscular system. The major scientific question that we ask in this paper if it is possible to alter the gait of healthy individuals using special purpose design of robots and training paradigms. This paper describes novel experimental results with an active leg exoskeleton (ALEX) and a force-field controller (FFC) developed for neuromotor training of gait and rehabilitation of patients with walking disabilities. ALEX is a motorized leg orthosis having a total of 7 DOFs with hip and knee actuated in the sagittal plane. The FFC applies forces on the foot to help the leg move on a desired trajectory. The interaction forces between the subject and the orthosis are designed to be "assist-as-needed" for safe and effective gait training. Simulations and experimental results with the FFC are presented. Experiments have been performed on six healthy subjects walking on a treadmill. It was shown that a healthy subject could be retrained in about 45 min with ALEX to walk on a treadmill with a considerably altered gait. In the coming months, this powered orthosis will be used for gait training of stroke patients.**

***Index Terms*—Force-field control, gait rehabilitation, rehabilitation robotics.**

## I. Introduction

NEUROLOGICAL damage, such as from stroke, can result in significant muscle weakness or impairment in motor control. Such patients often have substantial limitations in movement. Physical rehabilitation aims to improve walking function. It is becoming clear that for such rehabilitation to be successful, the patient must practice the task with high levels of repetition. Robotic rehabilitation has many advantages over conventional manual rehabilitation such as: 1) reduced dependence on clinical staff; 2) quantitative assessment of motion and interaction forces and torques measured using sensors; and 3) delivery of controlled repetitive training using robotics at a reasonable cost.

Currently, available lower extremity orthotic devices can be classified as either passive, where a human subject applies forces to move the leg and orthosis, or active, where actuators on the orthosis apply forces on the human leg. One of the passive devices is a gravity balancing leg orthosis (GBO), developed at the University of Delaware [1]. This orthosis can alter the level of gravity load acting at a joint by suitably altering parameters of the spring within the device. This device was tested on stroke survivors to characterize the effect of GBO on gait over a 3-month training period [2].

Passive devices cannot supply energy to the leg. Hence, they are limited in their ability to modulate the gait compared to active devices. This is the primary reason for developing active leg exoskeleton (ALEX), a motorized orthosis. One orthosis that is currently available commercially to clinics is Lokomat, which is an actively powered exoskeleton, designed originally for use with patients having spinal cord injury. A patient uses a Lokomat while walking on a treadmill [3]. Lokomot has 3 DOFs in vertical plane for each leg at ankle, knee, and hip joints, and 1 DOF for the device trunk. Mechanized gait trainer (MGT) is a single-DOF-powered machine that drives the leg to move in a prescribed gait pattern. The machine consists of a foot plate connected to a crank and rocker system. The device simulates the phases of gait, supports the subjects according to their ability, and controls the center of mass in the vertical and horizontal directions [4]. AutoAmbulator is a rehabilitation machine for the leg to assist individuals with stroke and spinal cord injuries. This machine is designed to replicate the pattern of normal gait, it has 3 DOFs for each leg [5]. HAL is a powered suit for elderly and persons with gait deficiencies [6]. This device takes EMG signals as input and produces appropriate torque to perform the task. Haptic walker [7] is a motorized machine developed for gait rehabilitation based on the principle of programmable footplates with permanent foot–machine contact. It uses mobile footplates in sagittal plane to generate desired trajectories. PAM, pelvic assist manipulator, is an active device for assisting the human pelvis motion [8]. There are a variety of active devices that target a specific disability or weakness in a particular joint of the leg [9]–[14].

A limiting feature of some of the early motorized exoskeletons developed for rehabilitation is that they move a subject through a predetermined movement pattern rather than allowing the subject to move under their own control. The failure to allow patients to actively modify their movement patterns may limit favorable changes in the nervous system required for relearning. Cai *et al.* [15] performed extensive experiments on mice and showed that forced motion along a fixed trajectory may result in "learned helplessness" [16]. This is suboptimal. Instead, "assist-as-needed" is a better approach [17]. Fixed repetitive training causes habituation of the sensory inputs and results in

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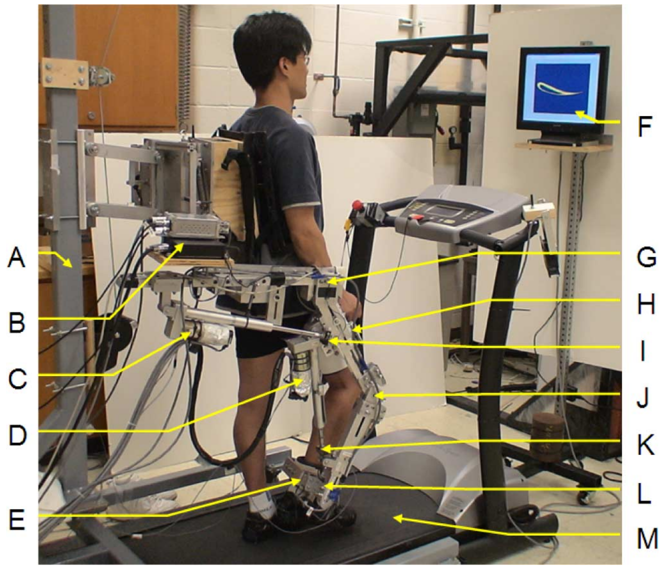



Fig. 1. Powered leg orthosis with a human subject, major components are labeled. A: walker, B: signal conditioning box, C: hip motor, D: knee motor, E: ankle brace, F: display for visual feedback, G: device hip joint, H: force–torque sensor (not visible) on device thigh, I: load-cell for hip linear actuator, J: device knee joint, K: load-cell for knee linear actuator, and L: force–torque sensor on device shank.

the patient not responding well to variations in these patterns. Hence, the interaction force between the human subject and the device plays an important role in training. For effective training, the involvement and participation of a patient in voluntary movement of the affected limbs is critical [18]–[20]. To involve a patient's participation in training, some of the patient cooperative controllers used with Lokomat are: 1) visual feedback of some measure indicating subject's performance; 2) use of impedance/admittance control; and 3) adaptive control [21]. Lower extremity powered exoskeleton (LOPES) [22] has been recently reported, which uses elastic actuators using Bowden cables. Haptic walker uses two modular robotic arms for each leg, with foot plates as end-effectors to apply forces directly on the feet of the subject.

Lokomat has been used in gait training studies [23] [24]. Lokomat training on chronic stroke survivors has not yielded beneficial results. Our design philosophy differs from that used with Lokomat and is motivated from the perspective of fully engaging a subject in gait training with ALEX by providing a flexible environment where a subject has significant control on their movements.

ALEX and the gait training setup is shown in Fig. 1. The subject walks on a treadmill with the orthosis attached on the right leg. The display in front of the subject provides visual feedback of the executed gait trajectory. In addition, the subject's performance can be recorded during each training session. The information displayed to the subject can be either relative joint angles (in joint space) or the relative foot coordinates (in foot space). ALEX is attached to a walker as shown in figure, a harness on the trunk of the device compensates for the device weight. This arrangement keeps the subject stably on the treadmill during training.

A force-field controller (FFC) is used to apply desired force fields on the moving leg. The goal of this controller is to assist or resist the motion of the leg, as needed, by applying force field on the foot. The user is not restricted to a fixed repetitive trajectory. Force-field control uses the concept of force assistance, as needed [25], [26]. In other words, FFC provides close to zero impedance when the subject moves within a user-defined range of the desired gait pattern and offers higher impedance with deviations outside this range. This impedance modulation and assistance is achieved by the controller. In addition, the controller also provides a small magnitude assistance or resistance force as required to move the foot along the trajectory. Active friction compensation is used to compensate for friction in the joints, as described in Section III-A.

One of the the main advantages of ALEX over other existing motorized orthoses are the multiple DOFs that a human subject has in the orthosis that allows a more natural walking ability of the user. These DOFs are explained in detail in the following section. Additionally, the patient cooperative FFC makes a subject actively participate in the training process. As explained earlier, FFC does not just modulate the impedance based on the current position of the foot, but also provides mild assistance or resistance by applying tangential forces to the trajectory when required.

An overview of this paper is as follows. Orthosis design is presented in detail in Section II. Architecture of FFC, simulation results, and an experiment initially performed on a dummy leg for safety are presented in Section III-B. Preliminary experiments conducted using ALEX on gait retraining of six healthy subjects and their results are described in detail in Section IV.

## II. Orthosis Design

### A. Device Description

Figs. 1 and 2 show the design of ALEX. Fig. 3 shows a simple kinematic figure of the DOF in the device. The overall setup has five main components.

1) Walker, which supports the weight of the device.
2) The "trunk" of the orthosis, which is a vertical plate behind the subject's back, is connected to a walker and has 3 DOFs with respect to the walker. These DOFs are: vertical translation, due to the parallelogram mechanism [see Fig. 3, joints: R1(a,b,c,d)], horizontal translation (left–right) in frontal plane [see Fig. 3, joints: T1(a,b)], and rotation about vertical axis (see Fig. 3, joint: R2). Human trunk is secured to the trunk of the orthosis with a pelvic brace. Since shoulder straps are not used, the human trunk can rotate forward.
3) Thigh segment of the orthosis has 2 DOFs, which are orthogonal to each other, with respect to the trunk of the orthosis: One DOF for abduction–adduction motion (see Fig. 3, joint: R3) and other DOF to move the leg in a sagittal plane (see Fig. 3, joint: R4). The thigh segment is telescopic and can be adjusted to match the thigh length of a human subject.

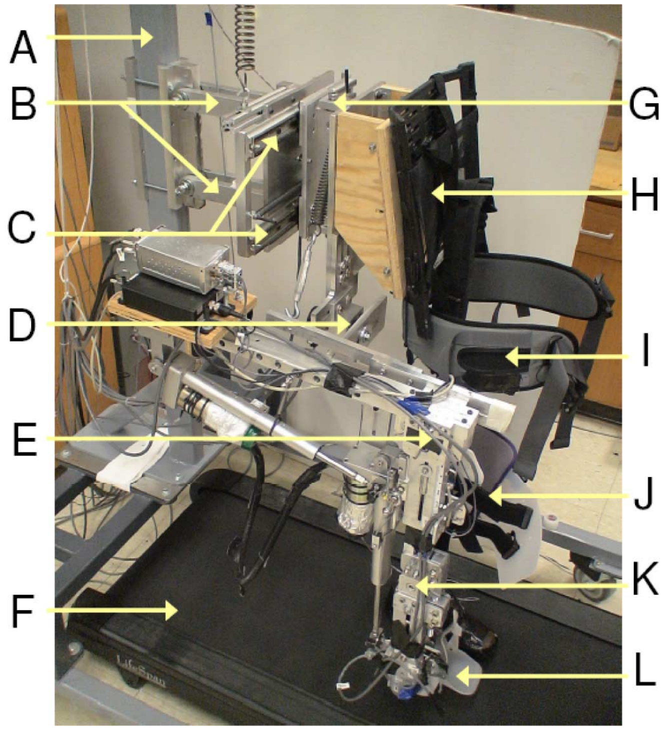



Fig. 2. ALEX in isometric view showing all the DOFs in the device. A: walker, B: parallelogram mechanism for vertical translation, C: sliders for horizontal translation in frontal plane, D: revolute joint for abduction and adduction motion of the leg, E: hip joint for thigh motion in sagittal plane, F: treadmill, G: revolute joint for rotation about vertical axis, H: back support for subject, I: pelvic straps, J: thigh brace, K: revolute knee joint, and L: shoe insert.

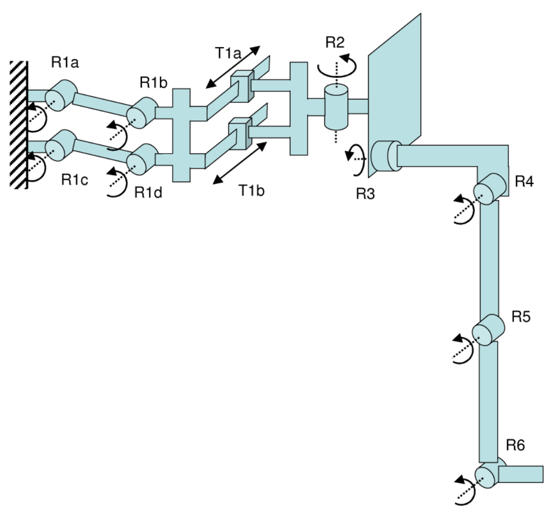



Fig. 3. Simple kinematic diagram of ALEX in isometric view showing all the DOFs in the device. R1(a,b,c,d): Revolute joints of the parallelogram mechanism for vertical translation, T1(a, b): sliders for horizontal translation in frontal plane, R2: revolute joint for rotation about vertical axis, R3: revolute joint for abduction and adduction motion of the leg, R4: hip joint for thigh motion in sagittal plane, R5: revolute knee joint, and R6: revolute ankle joint.

4) Shank segment of the orthosis, which has 1 DOF with respect to the thigh segment at knee joint (see Fig. 3, joint: R5). Shank segment, like thigh segment, is also telescopic.
5) Foot segment, which is a shoe insert, is attached to the shank of the leg with 1-DOF ankle joint for plantar/dorsi flexion (see Fig. 3, joint: R6). The foot segment also allows limited inversion–eversion motion at the ankle due to its structurally flexible design (not shown in Fig. 3).

All the previously mentioned DOFs were found to be essential for achieving a relatively natural walking motion of a subject. Fig. 2 shows a photograph of the device with the DOFs labeled. As shown in these photographs, user is attached to the device at the hip, thigh, and foot. A hip a large foam lined plastic brace is used to secure the subject, thigh is secured in a foam lined plastic brace, and the foot is secured by a shoe insert. These braces are moulded from plastic and are available in different sizes, the braces that fits a given subject are selected and attached to the device to ensure subject's comfort.

The hip joint in the sagittal plane and the knee joint are actuated using linear actuators. These motors have encoders built into them, which are used to determine the joint angles. Hip and knee joint angles can be measured with accuracy of at least 0.02°, such high accuracy is due to the encoder being mounted on the motors and not at the hip or knee joints directly. Ankle joint is not actuated but only measured using an encoder with 0.3° accuracy, which is mounted directly at the ankle of the device. An optical encoder measures the ankle joint angle. The change in the ankle joint motion is an important indicator in the rehabilitation of stroke survivors who suffer from the problem of foot drop. However, in ALEX control of hip and knee joints, this DOF is not used. The foot is considered a point mass and lumped at the ankle. Using kinematics, foot position with respect to hip joint can be computed. All other DOFs are passively held with springs. These motors can generate about 50 Nm peak torques at the device's knee and hip joints. The physical interface between the orthosis and the dummy/human leg is through two force–torque sensors, one mounted between thigh segments of the orthosis and the leg, the other mounted between shank segments of the orthosis and the leg. These force–torque sensors measure the interfacial forces and torques, these data were used in postprocessing for computing human applied hip and knee torques. The exoskeleton is designed to accommodate subjects within $\pm 10\%$ variation in thigh and shank lengths from average, and within $\pm 30\%$ variation in body mass from average (average values obtained from [27]). The hip joint range is about $\pm 40°$, where as knee joint range varies with leg lengths, it is at least 45° and can go up to 60°.

The ankle segment described earlier is used when a human subject is in the device. During initial testing of the controllers, we used a dummy leg which does not have a foot segment. Therefore, the aforementioned foot segment of the orthosis is not necessary during this testing. Heel-strike and toe-off events were recorded using flexible membrane foot switches taped under the both shoes. These events were used to determine swing and stance phases of each leg.

Since the device is motorized, it needs safety features to protect the subject using the device and the therapist, which are described in detail in the following section.

## *B. Safety*

As mentioned earlier, the device was first tested extensively with a dummy leg, which is similar to an average human leg in geometry and inertia distribution but without any actuators. One of the safety features of the device is "physical stops," placed at extreme ends of the allowed range of motion of each DOF. For the actuated hip and knee joints, the stops can withstand the maximum torque the actuators can apply. The motor force is limited to a maximum value manually set in the controller. The maximum force is also limited by the capacity of the dc motor and the power supply which, as mentioned earlier, is limited to 50 Nm peak torque at hip and knee joints. Two emergency stop switches are wired such that, when activated, can stop both the ALEX and the treadmill. One emergency stop switch is accessible to the subject, while the other is given to the personnel overseeing the therapy.

## III. Control Architecture

### A. Backdrivability of the Actuators

To meet the challenging goal of using a light weight motor and at the same time have the motor provide enough torque, we chose a linear actuator driven by an electric motor. These linear actuators cannot be back-driven, i.e., it is difficult to apply external force on the linear actuator and drive the motor. This happens because the friction and damping force in the lead screw of the motor get magnified due to the high gear ratio. A ball-screw transmission, with comparatively less friction, could not be used as none available met the speed and force transmission requirements. However, by using active friction compensation using the lead-screw-based linear actuator, the motor can be driven back easily (or "back-driven" easily). The property of "backdrivability" of actuators is important from the perspective of force control, as it increases the bandwidth of the device and makes it easier for a subject to move the leg without sizable resistance from the device.

The following friction compensation methods were tried out: 1) model-based compensation, in which we feed-forward frictional forces to the controller using a friction model obtained from experiments [28]; 2) load-cell based compensation, in which we use load cells in-line with the lead screw of the linear actuator along with a fast PI controller.

For feed-forward friction compensation, a good friction model is required. Fig. 4 shows typical frictional force data collected on the motor as a function of its linear velocity. The plot has ripples in the first and third quadrants. This behavior can be approximated by an equation $F_{\text{friction}} = K_{fs}\,\text{sign}\,(\dot{x}) + K_{fd}\dot{x}$, where $\dot{x}$ is the linear velocity of the motor, and $K_{fs}$ and $K_{fd}$ are constants determined from the data.

The friction model is only an approximation and the actual frictional force has a complicated dependency on the load applied to the motor and current configuration of the device [29], [30]. Some of the problems of model-based friction compensation were overcome by using a load cell in series with a fast PI controller with tuned parameters [31]. This strategy increases the bandwidth of the device and ensures that friction is compensated. It was found that friction feed-forward along with load-cell-based compensation gave the best operational performance.

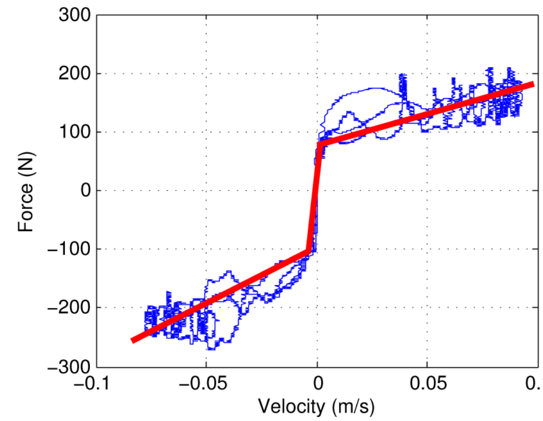



Fig. 4. Frictional force in a linear actuator as a function of its linear velocity. Thin blue line shows the experimental data and solid red line is the approximated friction model.

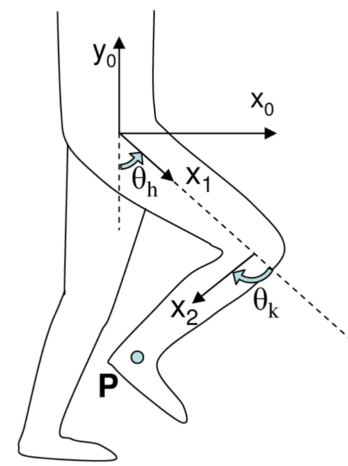



Fig. 5. Schematic showing anatomical joint angle convention.

### B. Force-Field Controller

Experiments have been conducted using the dummy leg with different controllers and based on their performance, we chose to use FFC to conduct experiments with human subjects. This controller is explained in this section.

The equations of motion for the combined device and the human leg are given shortly, note that the frictional terms are not mentioned here as they are assumed to be compensated using compensation methods outlined in Section III-A

$$\mathbf{M}(\boldsymbol{\theta})\ddot{\boldsymbol{\theta}} + \mathbf{C}(\dot{\boldsymbol{\theta}}, \boldsymbol{\theta})\dot{\boldsymbol{\theta}} + \mathbf{G}(\boldsymbol{\theta}) = \boldsymbol{\tau}_m \tag{1}$$

where $\boldsymbol{\theta} = [\theta_h\ \theta_k]^T$ shown in Fig. 5, $\mathbf{M}$ is the mass matrix, $\mathbf{C}$ is the matrix containing coriolis and centrifugal terms, $\mathbf{G}$ is the matrix due to gravity, and $\tau_m = [\tau_1\ \tau_2]^T$ is the torque applied at the hip and knee joints. The goal of the FFC is to create a force field around the foot, while providing damping to stabilize it. This force field creates a "virtual tunnel" along the desired trajectory in the sagittal plane. Fig. 6 shows the basic structure of the controller. This controller uses gravity compensation to help the human subject. This assistance can be reduced or completely removed if required. Fig. 7 shows

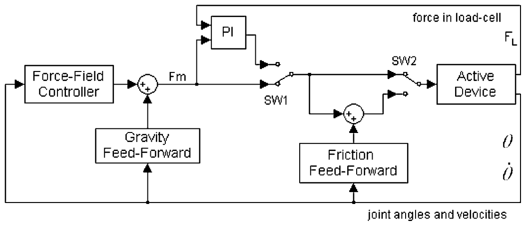



Fig. 6. FFC. $\mathbf{F_L}$ is the force measured by the load-cell. Switch SW1 turns on sensor-based friction compensation and switch SW2 turns on model-based friction compensation.

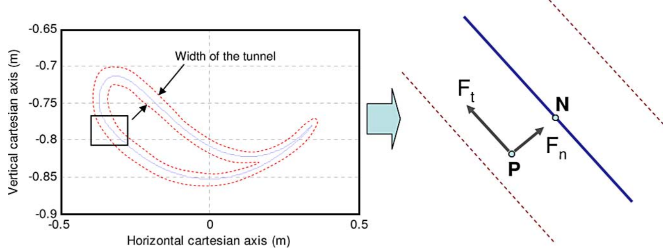
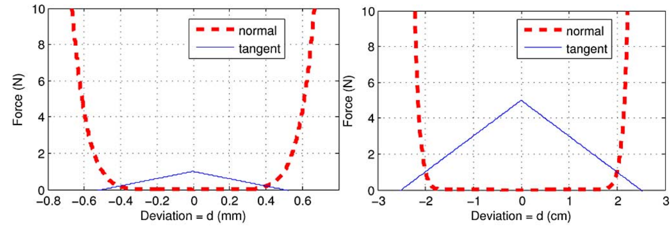



Fig. 7. Cartesian plot of the foot in the trunk reference frame, origin set at the hip joint. The solid line (in blue) is the desired trajectory of the foot and the dashed lines (in red) are the virtual walls.




(a) (b)

Fig. 8. Tangential and normal forces as a function of distance from the desired trajectory, positive tangential force points along the trajectory and positive normal force points toward the trajectory. Positive $d$ is meant to be shown as the distance in the same direction as that of the normal and negative $d$ in opposite direction. Parameters used are: (a) $K_{Ft} = 1\mathrm{N}, D_n = 0.0005\,\mathrm{m}, D_t = 0.0005\,\mathrm{m}, n = 3$, (b) $K_{Ft} = 5\,\mathrm{N}, D_n = 0.02\,\mathrm{m}, D_t = 0.025\,\mathrm{m}, n = 10$. (a) Narrow tunnel. (b) Wider tunnel.

typical shape of a virtual tunnel around a desired trajectory. It also shows the width of the virtual tunnel walls.

Since the virtual walls are used to guide the foot of the subject, the forces are applied on the foot. These forces are a combination of tangential force $\mathbf{F}_t$, normal force $\mathbf{F}_n$ to the desired trajectory, and damping force $\mathbf{F}_d$. We designed the controller such that this normal component provides proprioceptive feedback to the patient that his/her foot is deviating from the virtual tunnel and, therefore, helps keep the foot within the desired range of footpaths specified by the tunnel width. The tangential force provides the force required to move the foot along the prescribed path in forward direction, while the damping force limits the velocities. The desired foot trajectory is designed using cartesian coordinates in the reference frame attached to the hip of the subject.

Let $\mathbf{P}$ be the current position (see Figs. 5 and 7) of the foot in the Cartesian space in the reference frame $(\mathbf{x}_0, \mathbf{y}_0)$ attached to trunk of the subject. The origin of the reference frame is chosen at the hip joint of the subject, while the plane containing the vectors $\mathbf{x}_0$ and $\mathbf{y}_0$ is chosen to be the plane containing the thigh and shank segments of the subject. Let $\mathbf{N}$ be the nearest point to $\mathbf{P}$ on the desired trajectory, $\hat{\mathbf{n}}$ the unit normal vector from $\mathbf{P}$ to $\mathbf{N}$, and $\hat{\mathbf{t}}$ is the unit tangential vector at $\mathbf{N}$ along the desired trajectory in the forward direction. The force $\mathbf{F}$ on the foot is taken as

$$\mathbf{F} = \mathbf{F}_t + \mathbf{F}_n + \mathbf{F}_d \tag{2}$$

where $\mathbf{F}_t$ is the tangential force, $\mathbf{F}_n$ is the normal force, and $\mathbf{F}_d$ is the damping force. The tangential force $\mathbf{F}_t$ is selected as

$$\mathbf{F}_t = \begin{cases} K_{Ft}(1 - d/D_t)\hat{\mathbf{t}}, & \text{if } d/D_t < 1 \\ 0, & \text{otherwise} \end{cases} \tag{3}$$

where $K_{Ft}$ and $D_t$ are constants. $d$ is the distance between the points $\mathbf{P}$ and $\mathbf{N}$. $K_{Ft}$ is used to change the magnitude of the maximum tangential force. $\mathbf{F}_t$ is maximum along the desired footpath (where $d = 0$), its magnitude decreases linearly and becomes zero at a distance of $D_t$ from the desired footpath. Note that the tangential force decreases as the distance $d$ increases, the goal is to apply the tangential force only when the leg is close to the desired footpath. The tangential force helps the subject to move the foot along the desired footpath, the magnitude of this force can be changed during experiment by changing $K_{Ft}$. The normal force $\mathbf{F}_n$ is given by

$$\mathbf{F}_n = \left| \left( \frac{d}{D_n} \right)^n \right| \hat{\mathbf{n}} \tag{4}$$

where $D_n$ and $n$ are constants. $d = D_n$ is the distance from the desired footpath where $\mathbf{F}_n$ equals 1N. $n$ is just a scalar constant to change the steepness of the $\mathbf{F}_n$ vs $d$ curve. As $d$ increases in the previous equation $\mathbf{F}_n$ increases until it reaches a limit set in the controller. The normal force is given by Eq. (4). The higher the value of $n$, the steeper are the walls. Also, at higher values of $n$, the width of the virtual walls gets closer to $2D_n$. The damping force $\mathbf{F}_d$ on the foot, to limit velocities, is given by

$$\mathbf{F}_d = -K_d\dot{\mathbf{x}} \tag{5}$$

where $K_d$ is a constant and $\dot{\mathbf{x}}$ is the linear velocity of the foot.

Fig. 8 shows plots of tangential and normal forces as a function of distance $d$ from the desired trajectory, positive force points toward the trajectory.

The required actuator inputs [see (1)] at the leg joints that apply the previously defined force field $\mathbf{F}$ is given by

$$\tau_m = \begin{bmatrix} \tau_{m1} \\ \tau_{m2} \end{bmatrix} = \mathbf{J}^T\mathbf{F} + \mathbf{G}(\boldsymbol{\theta}) \tag{6}$$

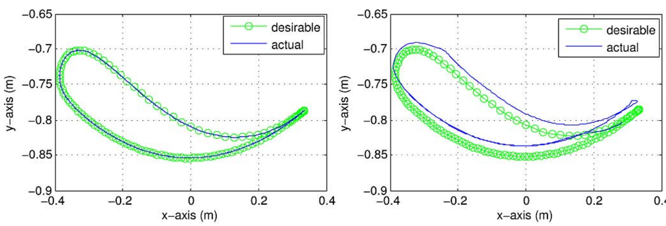
where $\mathbf{G}(\boldsymbol{\theta})$ is for gravity compensation and $\mathbf{J}$ is the Jacobian matrix. Finally, the forces in the linear actuators $\mathbf{F}_m = [F_{m1},\ F_{m2}]$ are computed using the principle of virtual work, given by

$$\mathbf{F}_{mi} = \frac{\dot{\theta}_i}{\dot{l}_i}\tau_{mi}, \qquad i = 1, 2$$

where $l_i$ is the length of $i$th linear actuator.

TABLE I
GEOMETRIC AND INERTIAL PARAMETERS OF HUMAN SUBJECTS

| Subjects | Age (yr) | Thigh Length (m) | Shank Length (m) | Body Height (m) | Body Mass (kg) |
|---|---|---|---|---|---|
| E1 | 45 | 0.405 | 0.433 | 1.78 | 70.7 |
| E2 | 27 | 0.43 | 0.42 | 1.78 | 62.6 |
| E3 | 21 | 0.388 | 0.395 | 1.70 | 61.23 |
| C1 | 34 | 0.384 | 0.427 | 1.80 | 94.8 |
| C2 | 23 | 0.43 | 0.42 | 1.78 | 81.19 |
| C3 | 24 | 0.38 | 0.42 | 1.83 | 103.42 |
| D1 | - | 0.40 | 0.40 | - | 72.57(*) |



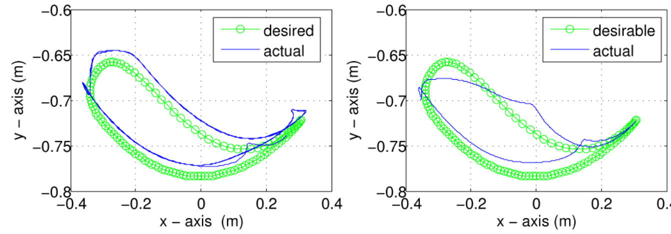

(a) (b)

Fig. 9. Simulation results of FFC showing foot trajectory in Cartesian frame attached to the trunk, origin set at the subject's hip joint. (a) Narrow tunnel. (b) Wider tunnel.

From previous equations, we can see that irrespective of where the leg is, the forces on the foot always point toward the desired trajectory. If the subject stops, the magnitude of forces will not increase with time. Also, there are no set points where the system will get stuck. In the implementation of the FFC, the nearest point $\mathbf{N}$ (see Fig. 7) is contiguously determined using the current location of the foot at point $\mathbf{P}$. This helps to have smooth change in the force without jerk.

### C. Simulations

Simulations were performed using the parameters shown in Fig. 8. In the simulations, healthy subject's average gait data were used as the desired foot trajectory [32]. Data of an average human being [27] was used for geometry and inertial parameters of the leg in the simulations (see Table I). The goal of these simulations was to obtain reasonable device and control parameters for use in real experiments. Fig. 9 shows the results of two simulations done with different widths of virtual walls, Fig. 9(a) uses narrow width of $D_n = 0.0005\,\mathrm{m}$ and $n = 3$, whereas in Fig. 9(b) uses $D_n = 0.02$ m and $n = 10$. With a narrow width between virtual walls, the error in the desired trajectory and the trajectory achieved is smaller when compared to the wider width in between the virtual walls. This shows that the maximum deviation of the foot from the desired trajectory can be controlled using $D_n$ and $n$ parameters. In other simulations, when $K_{Ft}$ is increased from 40 to 60 N, and all other parameters kept the same ($D_n = 0.02\,\mathrm{m}, D_t = 0.025, n = 10$), the tangential force also increases. As a result, the gait cycle period reduced from 5.0 to 3.8 s.




(a) (b)

Fig. 10. Results of FFC showing foot position in cartesian frame attached to the trunk of the subject with origin at the subject's hip joint. (a) Simulation result. (b) Experimental result.

### D. Experimental Results With Dummy Leg

For safety, the experiments with FFC were first conducted with dummy leg in the device. Here also, healthy subjects' average gait data were used to define the desired foot trajectory [32]. The dummy leg was attached to the orthosis. The orthosis and the dummy leg executed the trajectory. The parameters used were $K_{Ft} = 50\,\mathrm{N}, D_n = 0.012\,\mathrm{m}, D_t = 0.025\,\mathrm{m}, n = 3$. Dummy leg has no actuators, so it does not apply torques at its joints.

Fig. 10 shows plot of foot trajectory obtained from experiment and through simulation with the same set of parameters ($K_{Ft} = 50\,\mathrm{N}, D_n = 0.012\,\mathrm{m}, D_t = 0.025\,\mathrm{m}, n = 3$). Again, the difference between them could be attributed to the inexact mathematical model used in the simulation where a simplified friction model is used. In the figures, one can see the foot bouncing off the virtual walls. This bouncing may not appear with the human subject in the device, as the subject can look in the display (visual feedback) at his gait and correct any errors in his gait pattern. With this controller, the motion of the leg shows a limit-cycle like behavior. This limit-cycle-like behavior has been explored by other researchers to control bipeds [33].

## IV. Preliminary Experiments With Healthy Subjects

Preliminary experiments with the ALEX were performed on six healthy adult subjects. This was done to evaluate whether the device could be used to induce short-term adaptations in the walking pattern of healthy individuals as a test of concept. These subjects have given us the informed consent for performing the gait training study. In these experiments, the subjects also received visual feedback of their gait trajectory, which can also help in inducing adaptation. To determine the relative training effect of the device alone, we used two subject groups: FFC + visual guidance (VG) group and VG group. A subject in the FFC+VG group is trained with both the active assistance (FFC) and the VG. For a subject in the VG group, only visual guidance is provided. Hence, the difference in adaptation to the altered gait pattern between the FFC+VG group and VG group reflects the added influence of training with the force-field control. Data acquisition and real-time control is performed using dSpace DS1103 PPC Controller Board. Geometric and inertial parameters of human subjects are given in Table IV.

TABLE II
DETAILS OF GAIT TRAINING FOR SUBJECTS IN EXPERIMENTAL AND CONTROL GROUPS

| Session | Time (min) | Visual Feedback | Force Field Constraint | | Tunnel Width (mm) |
|---|---|---|---|---|---|
| | | | VG | FFC+VG | |
| Baseline | 5 | **off** | no | no | - |
| Pre-test | 2 | on | no | no | - |
| Training 1 | 15 | on | no | **yes** | 3 |
| Training 2 | 15 | on | no | **yes** | 6 |
| Mid-test | 2 | on | no | no | - |
| Training 3 | 15 | on | no | **yes** | 6 |
| Training 4 | 15 | on | no | **yes** | 8 |
| Post-test | 2 | on | no | no | - |
| Follow-up | 2 | **off** | no | no | - |

### A. Procedure

Three subjects out of six were assigned to the FFC+VG group, who received active assistance from ALEX. This assistance is provided by the FFC during training to produce the sagittal plane foot trajectory. As discussed in Section III-B, FFC applies normal force that simulates a virtual tunnel-like force field around the foot in sagittal plane. Equation (4) can be used to set virtual tunnel wall stiffness so that subjects encounter a nonlinear spring-like resistance when the footpath deviates from the prescribed footpath. The width between the virtual walls could be set to allowable deviation of the current footpath from the prescribed footpath. A significant spring-like resistance is encountered only when the subject reached the virtual walls due to the nature of (4). This normal force can be thought of as an impedance based controller. Tangential force is applied by the controller to assist along the desired footpath. This force was found to be necessary to help the subject to track and go around the corners of the desired trajectory well. Three subjects in the control group tried to match a prescribed template without receiving any assistance from ALEX. All subjects walked at their self-selected treadmill speed.

Table II shows the details of the training procedure for subjects in experimental and control groups. Following is the description of the steps involved.

1) *Baseline:* First, the subjects were asked to walk in the device for 5 min to get familiar with it. At this time, force field and visual feedback were not provided. At the end of this block of time, subject's "baseline" gait data were recorded. A prescribed target foot trajectory was then devised by geometrically scaling down each subject's baseline footpath, particularly in the vertical dimension, enough to make replication of the prescribed path challenging to the subject. This was done by choosing the target foot trajectory (template) such that it had 25% less hip flexion and 15% less knee flexion when compared to the subject's baseline trajectory.
2) *Training 1 to 4:* Subjects in both groups received four 15-min blocks of training during which the prescribed template and the subject's current footpath were displayed on a video monitor in front of the subject. Visual feedback was continuous for all training blocks. All four training blocks were identical for the subjects in VG group, as force field was not provided. For the subjects in FFC+VG group, both visual guidance and force field were provided. For them, both the stiffness of the virtual walls and assisting force were kept constant ($n = 1, K_t = 50\,\mathrm{N}, K_d = 30\,\mathrm{N \cdot s/m}, D_t = 1\,\mathrm{m}$). What changed was the width of the virtual walls, i.e., the degree of constraint on the footpath. The constraint was maximal for the first block of training trials (Training 1) as the width between the virtual walls was the narrowest ($D_n = 0.003\,\mathrm{m}$). The footpath was less constrained, i.e., allowed for greater deviation of the footpath from the prescribed path before resistance was encountered for blocks 2 and 3 ($D_n = 0.006\,\mathrm{m}$). The constraint of the controller was further reduced during the fourth block (Training 4) of trials ($D_n = 0.008\,\mathrm{m}$). The rationale for the reduction in the constraint, by increasing the width between the virtual walls, is the following: when the constraint is maximum, the subject is more strongly directed back to the desired footpath. Another way of thinking about this is that the range of allowable footpaths is limited. This provides the subject with the necessary proprioceptive feedback with which the subject "feels" the trajectory. When the constraints are relaxed, the subject has greater latitude to deviate from the desired trajectory, but if training has been effective, he should still remain close to the desired trajectory without the stronger constraint. This active participation is what is required to help retrain the muscles of the patient.
3) *Pre, mid, and posttests:* Evaluation of the subjects' walking performance on the treadmill was obtained with the FFC turned off (with friction compensation still on) at three time points: 1) prior to the initial block of training (pretest); 2) prior to block 3 of training (midtest); and 3) after block 4 of training (posttest). All of these data collections were performed with the template and current foot trajectory displayed to the subject.
4) *Follow-up (Retention):* Following the posttest, visual feedback was turned off (FFC is also off) and the subject's performance was recorded to test for evidence of footpath adaptation or retention. The subjects were asked to walk as normally as possible, with no reference to the adapted foot path.

## V. TRAINING RESULTS

The goal of these experiments was to see if short-term adaptations in the gait pattern of healthy subjects could be induced by training using ALEX with force-field control. Force-field control helps the subject to voluntarily participate in the training process. This is achieved by using maximum constraints in the early part of the training (Training 1), where the narrowest width between the virtual walls is used and gradually relaxing the constraints, i.e., gradually increasing the width between the virtual walls. As the width of the virtual tunnel is increased, the subject gets less help from the device to track the desired

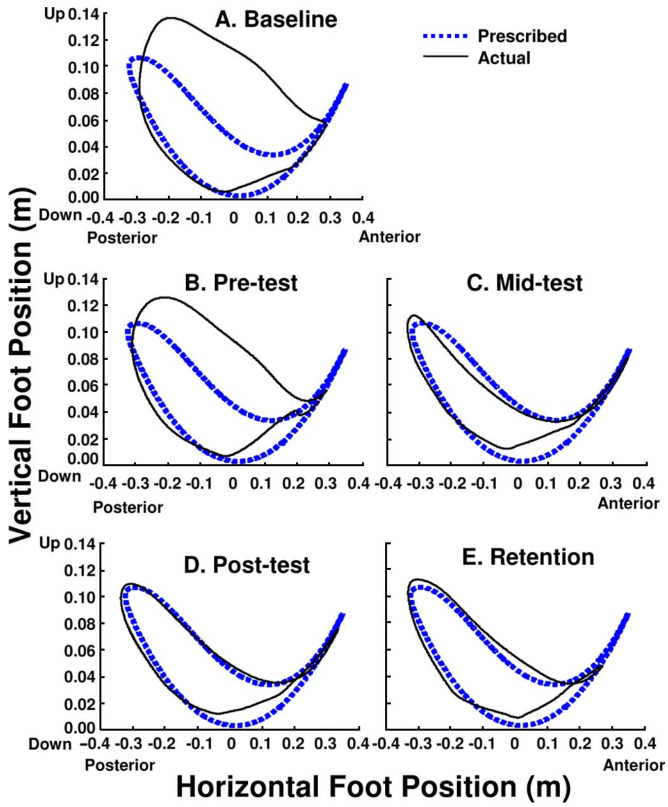



Fig. 11. Plots of sagittal plane footpath of a subject in FFC+VG group from four evaluation points after training with the ALEX. Although it is shown in the plot for reference, the prescribed trajectory was not shown to the subject during the retention test (bottom, right panel). Please note that in these plots, the $y = 0$ of the axes is on the treadmill instead of the hip joint.

foot trajectory. By the end of the training, if the subject could track the desired footpath well, this provides evidence for adaptive neuromuscular changes that allow his foot to move along a new footpath without voluntary intervention (recall that subjects were asked to walk as normally as possible without reference to the adapted gait pattern). This is what we want a patient to do. The aim of the rehabilitation is to retrain patient's leg muscles to achieve a new or improved gait pattern. Although in this pilot study, healthy subjects were used to test the concept, our ultimate goal is to apply ALEX to the training of stroke survivors in the hope of improving their gait patterns toward a pattern that will help them become community ambulators.

Results of a representative experimental subject's performance in FFC+VG group during the evaluation sessions are illustrated in Fig. 11, where the thin solid line is the subject's average footpath and the heavy dashed line is the prescribed foot path. As shown in the figure, the actual footpath deviated substantially from the required trajectory at the initial evaluation but was shaped to approximate the required trajectory by the midtraining evaluation.

To quantitatively determine the amount of adaptation, a measure called "footpath deviation area" is used, which is the geometric area included between the swing phases of given foot trajectory and prescribed trajectory. This amount of area is the deviation of the given trajectory from prescribed trajectory in

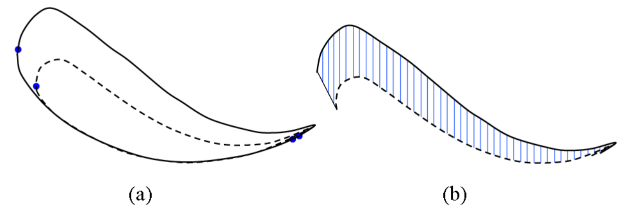



Fig. 12. Computed area between swing phases of given test gait and the template.

the template. Fig. 12(a) shows foot trajectory in one of the tests as solid black line, with the prescribed trajectory as dashed black line. The blue filled circles show the foot events, toe-off on left and heel-strike on right side, for both the trajectories. Now stance portion of both the trajectories are dropped and toe-off points of both the swing phases are connected by a straight line; similarly heel-strike points are also connected, this gives a closed polygon. To find the area of this polygon, which can potentially self-intersect, a MATLAB program was written that uses an algorithm called "plane-sweep" method [34]. In this method, the polygon is divided into multiple vertical strips, as shown in Fig. 12(b), and the area of individual strip is computed and added together to get the area of the polygon. The area of each strip is always positive and the direction of polygon vertices is not considered.

If the difference between footpath deviation of baseline and foot path deviation of follow-up is large, it means the follow-up trajectory is close to the prescribed path. Hence, the larger the difference in area, the better is the gait adaptation.

Summary results of the difference in area between the baseline swing phase footpath and other test periods (i.e. area between the curves) are illustrated in Fig. 13 for both VG and FFC+VG groups. Note that the baseline footpath is farthest from the template, while the later tests are closer to the template. Thus, positive values indicate that the difference from the template is less in the other test trials compared to the baseline test. A negative value indicates that the footpath for a given test trial is farther from the required template than it was at baseline. Note that the FFC+VG group showed greater adaptability in footpath compared to the VG group, although there was large intersubject variability and statistical analysis could not be conducted due to the small number of subjects in this pilot experiment. The VG group showed some initial adaptation followed by a decrease in the ability to match the template thereafter, with poorer performance at the retention test than at baseline ("after-effect"). In contrast, subjects the in the FFC+VG group showed a consistent increase in adaptation through the midtest that was maintained even after removal of visual feedback, as seen at the retention test in the Fig. 13. *This result is striking given that walking patterns are highly ingrained through many years of practice*. Of course, interpretation of this result requires caution given that treadmill walking differs significantly from over ground locomotion and is not well-practiced. Nonetheless, the results are encouraging. Our next step is to obtain a larger set of data on healthy subjects using this paradigm and then to

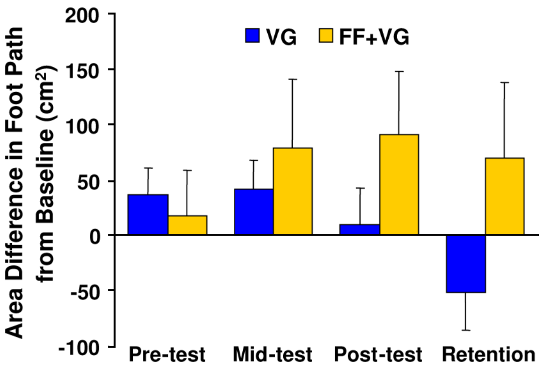



Fig. 13. Average across-subject area differences in footpath from baseline between visual guidance only (VG; $n$=3) and visual guidance plus force-field constraint (FFC+VG; $n$=3) groups. The greater the positive value on the $y$-axis, the better the adaptation. Note that the trained pattern in the FFC+VG group was retained at the retention test (no visual feedback), but not in the VG group. The error bar stands for the standard error of the mean.

evaluate short-term and longer term training of patients with neurological disorders such as stroke.

## VI. Discussion

This paper presents a motorized orthosis named ALEX for rehabilitation of motor impaired patients. This device uses the scheme of force-field control. One of the novelty of the hardware is that it has several DOFs that allows a more natural walking pattern on the treadmill. The orthosis has a spring loaded overhead harness to carry the weight of the device only. The patient's body weight is not supported; therefore, the current design is useful for only those patients who have enough strength in their legs for stance phase support of their body. Hemiparetic stroke survivors who have enough strength for stance phase support in their affected leg are the ideal candidates for rehabilitation with this device. The force-field control scheme, as described earlier, helps the patients by assisting only when needed, and resisting the undesirable motion of the leg. This is done by using maximum constraints on the leg in the initial part of the rehabilitation. These maximum constraints allow only small foot deviation, effectively helping the patients when their deviations would tend to be large. As the patient is trained and is able to match the desired trajectory well enough, the constraints are gradually relaxed and the patient have to voluntarily try to "remember" the prescribed path. This method of training currently is undergoing preliminary testing for rehabilitation of stroke patients. The effectiveness of this gait training procedure was evaluated preliminarily in the current study by conducting experiments on healthy subjects. Since healthy subjects have enough strength in their muscles to move their legs normally, the training procedure was modified slightly. Even for healthy subjects, the time of 10 min given for each training block, was sufficient to get some adaptation as seen in the results. From the results, we see that gait training of 40 min was sufficient for neuromuscular retaining required to achieve adaptation to an altered gait pattern. Especially for the subject shown in Fig. 11, the fact that the trained pattern was maintained reasonably well in this subject following training, after the feedback was turned off completely, is remarkable. This provides encouragement that the same procedures may help selective stroke survivors improve their gait pattern, with the goal of helping them become more functional community ambulators. For stroke patients, a healthy subject's foot path will be used as the prescribed footpath for gait training.

## VII. Conclusion and Future Work

ALEX was built and tested. It has linear actuators at hip and knee joints. It is also instrumented with force–torque sensors and encoders. To obtain unconstrained natural walking in the orthosis, 4 DOFs were allowed for the human trunk. FFC was developed to provide assistance to the patient. To make the linear actuators backdrivable, these controllers were used with: 1) model-based, and/or 2) load-cell-based friction compensation. This backdrivability of the actuators helps the device in making it responsive to the human applied forces by offering reduced impedance. Simulation and experimental results using FFC were presented. Experiments with a dummy leg were done initially and then with six healthy subjects. Among the six subjects, a group of three subjects receiving force-field constraints as well as visual guidance showed considerable adaptation of their normal gait pattern toward the prescribed gait pattern when compared to the group receiving visual guidance alone. Results show that from pretest to posttest the force-field group showed reduction in the foot position deviation from the prescribed path by 60%, whereas in the visual guidance group, the reduction was much smaller and actually increased at the retention test. In follow-up, the deviation in foot position for FFC+VG group was about 20% less than for the VG group. This shows that there was more adaptation of the walking pattern in the FFC+VG group than in the VG group. Currently, experiments with several healthy subjects are in progress to determine the statistical significance of training process. Simultaneously, preliminary experiments with stroke patients are also in progress to study the rehabilitation effects of the device.

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