An Assistive Controller for a Lower-Limb Exoskeleton for Rehabilitation after Stroke, and Preliminary Assessment Thereof

Spencer A. Murray-*IEEE Student Member*, Kevin H. Ha, *IEEE Student Member*, and Michael Goldfarb, *IEEE Member*

Abstract— **This paper describes a novel controller, intended for use in a lower-limb exoskeleton, to aid gait rehabilitation in patients with hemiparesis after stroke. The controller makes use of gravity compensation, feedforward movement assistance, and reinforcement of isometric joint torques to achieve assistance without dictating the spatiotemporal nature of joint movement. The patient is allowed to self-select walking speed and is able to make trajectory adaptations to maintain balance without interference from the controller. The governing equations and the finite state machine which comprise the system are described herein. The control architecture was implemented in a lower-limb exoskeleton and a preliminary experimental assessment was conducted in which a patient with hemiparesis resulting from stroke walked with assistance from the exoskeleton. The patient exhibited improvements in fast gait speed, step length asymmetry, and stride length in each session, as measured before and after exoskeleton training, presumably as a result of using the exoskeleton.**

I. INTRODUCTION

Each year, approximately 800,000 people in the US suffer a stroke or cerebrovascular accident (CVA), of which approximately 200,000 survivors are affected by lower-extremity hemiparesis to an extent that prevents walking without assistance six months after [1, 2]. Typical gait deficits in lower-limb-affected post-stroke individuals involve a combination of impaired muscle strength, coordination and proprioception, and often excessive muscle tone in the paretic limb. The two most immediate biomechanical effects of these impairments are instability of the paretic leg during the stance phase of gait, and insufficient foot clearance on the paretic side during the swing phase of gait. Given these biomechanical deficits, the movement objectives of post-stroke gait training primarily entail improving load acceptance on the paretic leg during stance, and improving foot clearance of the paretic leg via increased hip and knee flexion in the paretic leg during swing. These objectives have traditionally been pursued by a combination of physiotherapy (e.g., mat exercises, weight training, use of fitness equipment) and assisted overground gait training. These may be supplemented with body-weight-supported treadmill training (BWSTT) or robotically assisted treadmill training. Various methods have been proposed to control the patient-robot interaction in robotically-assisted BWSTT systems [3-7].

Emerging lower limb exoskeletons are wearable robots which permit overground walking. Thus an exoskeleton may permit the patient, rather than the treadmill belt speed, to set the pace of gait, which may be advantageous. Further, without the artificially stabilizing effect of an overhead suspension point, an exoskeleton may better promote balance recovery. Despite the efficacy of the aforementioned control methods [3-7] in robotically-assisted systems, such methods are less well-suited to walking overground in an exoskeleton. Such control methods either dictate or substantially influence the user's footpath, which may interfere with a patient's ability to quickly deviate from a prescribed path in order to adjust and maintain balance. As such, a control methodology for gait assistance for an exoskeleton should assist movement without governing the spatiotemporal nature of the footpath.

This paper describes a control approach which provides floor-referenced walking assistance without substantially affecting the user's ability to select a desired step length or time. Following a description of the control structure, the authors describe the implementation of the controller in a lower limb exoskeleton, and additionally describe some preliminary results of implementing the exoskeleton and controller on three post-stroke subjects.

II. CONTROLLER TO FACILITATE RECOVERY FOLLOWING STROKE

The general intent of the exoskeleton is to help a patient to recover the neural coordination associated with walking. The authors hypothesize that such recovery is facilitated by allowing the patient, rather than the exoskeleton, to provide movement coordination. The controller, described herein, therefore consists of the combination of three types of behaviors: gravity compensation, reinforcement of isometric joint torques, and supplementation of active joint torques, none of which act to enforce a specified trajectory. The respective components of the control approach, and the state

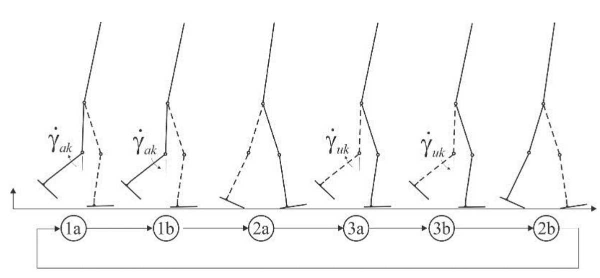

Fig. 1. Finite states corresponding to the assistive controller, where the affected leg is shown as a solid line and the unaffected leg as a dashed line. The three main states correspond to the 1) affected leg in swing, 2) double-support, and 3) unaffected leg in swing.

Research supported by U.S. Department of Health and Human Services under National Institute of Health Grant 1R01HD059832-01A1.

S.A. Murray, K.H. Ha, and M. Goldfarb are with Vanderbilt University, Nashville, TN 37209 USA (email: spencer.a.murray@vanderbilt.edu, kevin.h.ha@vanderbilt.edu, michael.goldfarb@vanderbilt.edu).

machine within which they operate, are described in the following sections.

A. Control States and Notation

The exoskeleton controller is governed by a finite state machine consisting of three states, as illustrated in Fig. 1, where the affected leg is shown as a solid line, and the unaffected leg as a dashed line. Each state is comprised of two sub-states, as follows: sub-states 1a and 1b correspond to the portions of swing in which the affected knee is in a state of flexion and extension respectively; sub-states 2a and 2b correspond to double-support following heel strike of the affected leg and unaffected leg, respectively; and sub-states 3a and 3b correspond to the portions of swing in which the unaffected knee is in a state of flexion and extension, respectively. The sequence of states through which the controller would transition under normal walking conditions is illustrated in Fig. 1. The controller assumes an exoskeleton with four actuators, which provide sagittal plane torques at both the affected and unaffected hip and knee joints. The actuator torque vector corresponding to the four actuator torques can therefore be defined as:

$$\boldsymbol{\tau} = [\tau_{ak} \quad \tau_{ah} \quad \tau_{uk} \quad \tau_{uh}]^T \tag{1}$$

where τ_{ak}, τ_{ah}, τ_{uk}, and τ_{uh} are the torque commands corresponding to the affected knee, affected hip, unaffected knee, and unaffected hip joints, respectively. A torque vector corresponding to state i (as described above) may be denoted by τ_i. For cases in which the control torque changes as a function of sub state, the torque commands can be further indicated by τ_{ia} or τ_{ib}. Within each state, the control torque may consist of a combination of multiple assistive torque components which are identified by the subscript j, such that the individual control torques can be denoted by τ_{ij}. Given this notation, the control torques corresponding to the various assistive components are described below.

B. Exoskeleton Gravity Compensation

A gravity compensation component of the controller is intended to remove the gravitational burden of the exoskeleton mass from the user, and is described by the following control law:

$$\boldsymbol{\tau}_{11} = g\begin{bmatrix} m_{es}l_{ces}\cos\theta_{as} \\ m_{et}l_{cet}\cos\theta_{at} + m_{es}l_{et}\cos\theta_{at} + m_{es}l_{ces}\cos\theta_{as} \\ m_{eh}l_{ceh}\cos\theta_{hat} + (m_{et}l_{cet} + m_{es}l_{et})\cos\theta_{at} + m_{es}l_{ces}\cos\theta_{as}\cdots \\ + ((m_{eh} + m_{et} + m_{es})l_{et} + m_{et}(l_{et} - l_{cet}))\cos\theta_{ut} \\ m_{eh}l_{ceh}\cos\theta_{hat} + (m_{et}l_{cet} + m_{es}l_{et})\cos\theta_{at} + m_{es}l_{ces}\cos\theta_{as} \end{bmatrix} \tag{2}$$

$$\boldsymbol{\tau}_{21} = g\begin{bmatrix} \frac{1}{2}m_{eh}l_{et}\cos\theta_{at} + m_{et}(l_{et} - l_{cet})\cos\theta_{at} + \frac{1}{2}m_{eh}l_{ceh}\cos\theta_{hat} \\ \frac{1}{2}m_{eh}l_{ceh}\cos\theta_{hat} \\ \frac{1}{2}m_{eh}l_{et}\cos\theta_{ut} + m_{et}(l_{et} - l_{cet})\cos\theta_{ut} + \frac{1}{2}m_{eh}l_{ceh}\cos\theta_{hat} \\ \frac{1}{2}m_{eh}l_{ceh}\cos\theta_{hat} \end{bmatrix} \tag{3}$$

$$\boldsymbol{\tau}_{31} = g\begin{bmatrix} m_{eh}l_{ceh}\cos\theta_{hat} + (m_{et}l_{cet} + m_{es}l_{et})\cos\theta_{ut} + m_{es}l_{ces}\cos\theta_{us}\cdots \\ + ((m_{eh} + m_{et} + m_{es})l_{et} + m_{et}(l_{et} - l_{cet}))\cos\theta_{at} \\ m_{eh}l_{ceh}\cos\theta_{hat} + (m_{et}l_{cet} + m_{es}l_{et})\cos\theta_{ut} + m_{es}l_{ces}\cos\theta_{us} \\ m_{es}l_{ces}\cos\theta_{us} \\ (m_{et}l_{cet} + m_{es}l_{et})\cos\theta_{ut} + m_{es}l_{ces}\cos\theta_{us} \end{bmatrix} \tag{4}$$

where θ_{as} and θ_{at} are the angles with respect to the vertical of the affected shank and thigh segments, respectively, and θ_{us} and θ_{ut} are the angles with respect to the vertical of the unaffected shank and thigh segments, respectively; m_{eh}, m_{et}, and m_{es} are the respective masses of the exoskeleton hip, thigh and shank segments; l_{ceh}, l_{cet}, and l_{ces} are the respective distances of the center of mass of the hip, thigh and shank segments of the exoskeleton from the hip, hip, and knee joints, respectively; l_{et} is the length of the exoskeleton thigh segment; and g is the magnitude of the gravitational acceleration.

C. Partial Compensation of Swing Leg Weight

Hemiparetic patients frequently exhibit reduced muscle strength in the affected limb, which can impair the ability to achieve healthy joint excursions. In order to provide movement assistance without dictating joint trajectories, one of the components of the exoskeleton controller is a partial limb-weight compensation of the affected leg during the swing phase of gait. The weight of the limb assists movement when movement is in the direction of gravity, such that active compensation could potentially increase the energetic output required of the user in these instances. As such, partial limb weight compensation is only exerted by the controller when the torque works against the energy gradient (i.e. when the exoskeleton joint is generating power). As such, the partial limb-weight compensation controller is described by:

$$\boldsymbol{\tau}_{12} = r[\tau_{ah} \quad \tau_{ah} \quad \tau_{uk} \quad \tau_{uh}]^T \tag{5}$$

$$\tau_{ak} = \begin{cases} m_s l_{cs}\cos\theta_{as} & \text{if } (m_s l_{cs}\cos\theta_{as})\dot{\gamma}_{ak} > 0 \\ 0 & \text{otherwise} \end{cases} \tag{6}$$

$$\tau_{ah} = \begin{cases} (m_t l_{ct} + m_s l_t)\cos\theta_{at}\cdots & \text{if } ((m_t l_{ct} + m_s l_t)\cos\theta_{at}\cdots \\ \quad + m_s l_{cs}\cos\theta_{as} & \quad + m_s l_{cs}\cos\theta_{as})\dot{\gamma}_{ah} > 0 \\ 0 & \text{otherwise} \end{cases} \tag{7}$$

$$\boldsymbol{\tau}_{22} = \boldsymbol{\tau}_{32} = [0 \quad 0 \quad 0 \quad 0]^T \tag{8}$$

where γ_{ak} and γ_{ah} are the joint angles of the affected knee and hip joints, respectively; m_t, and m_s are the respective masses of the user's thigh and shank segments; l_t is the length of the thigh segment (note that this is the same value as l_{et}); l_{ct}, and l_{cs} are the respective distances of the center of mass of the user's thigh and shank segments from the hip and knee

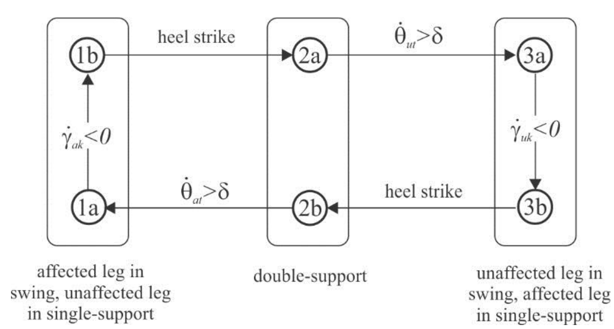

Fig. 2. Finite state machine switching conditions corresponding to the assistive controller.

joints, respectively; and $r \in [0,1)$ is a user-selectable gain that determines the extent of limb weight compensation during the affected-limb swing phase.

D. Feedforward Movement Assistance during Swing

Reducing the apparent weight of the swing limb reduces the burden of movement, while maintaining an energetically passive character of human/exoskeleton interaction. Such assistance, however, may not be sufficient to achieve suitable joint excursion during swing. In order to provide additional assistance without dictating joint trajectories, the controller allows the user to initiate a given movement, then supplements that movement with a brief torque pulse at the respective joint, as follows:

$$\boldsymbol{\tau}_{\mathbf{1a3}} = \begin{bmatrix} \tau_{ak} & \tau_{ah} & 0 & 0 \end{bmatrix}^T \tag{9}$$

$$\tau_{ak} = \begin{cases} \dfrac{P_{kf}}{2}\left(1+\sin\left(\dfrac{2\pi}{T_{kf}}t_a - \dfrac{\pi}{2}\right)\right) & \text{if } 0 < t_a < T_{kf} \\ 0 & \text{otherwise} \end{cases} \tag{10}$$

$$\tau_{ah} = \begin{cases} \dfrac{P_{hf}}{2}\left(1+\sin\left(\dfrac{2\pi}{T_{hf}}t_a - \dfrac{\pi}{2}\right)\right) & \text{if } 0 < t_a < T_{hf} \\ 0 & \text{otherwise} \end{cases} \tag{11}$$

$$\boldsymbol{\tau}_{\mathbf{1b3}} = \begin{bmatrix} \tau_{ak} & 0 & 0 & 0 \end{bmatrix}^T \tag{12}$$

$$\tau_{ak} = \begin{cases} \dfrac{P_{ke}}{2}\left(1+\sin\left(\dfrac{2\pi}{T_{ke}}t_b - \dfrac{\pi}{2}\right)\right) & \text{if } 0 < t_b < T_{ke} \\ 0 & \text{otherwise} \end{cases} \tag{13}$$

$$\boldsymbol{\tau}_{\mathbf{23}} = \boldsymbol{\tau}_{\mathbf{33}} = \begin{bmatrix} 0 & 0 & 0 & 0 \end{bmatrix}^T \tag{14}$$

where P_{kf} and T_{kf} are the torque pulse amplitude and duration, respectively, for the knee flexion torque pulse; P_{hf} and T_{hf} are the torque pulse amplitude and duration, respectively, for the hip flexion torque pulse; P_{ke} and T_{ke} are the torque pulse amplitude and duration, respectively, for the knee extension torque pulse; and t_a and t_b are the length of time since the controller entered sub-states 1a and 1b, respectively. Note that the amplitude and duration of each torque pulse are selected and adjusted as needed by a particular patient.

E. Reinforcement of Isometric Torques during Stance

The affected stance limb is often subject to instability, particularly at the knee joint, which can result in buckling, either in flexion or hyperextension. In order to prevent such buckling, the controller provides "soft" stops during single-support at the stance knee of the affected leg, which consist of emulated spring and damper couples as follows:

$$\boldsymbol{\tau}_{14} = \boldsymbol{\tau}_{24} = \begin{bmatrix} 0 & 0 & 0 & 0 \end{bmatrix}^T \tag{15}$$

$$\boldsymbol{\tau}_{34} = \begin{bmatrix} \tau_{ak} & 0 & 0 & 0 \end{bmatrix}^T \tag{16}$$

$$\tau_{ak} = \begin{cases} k(\gamma_{ak} - \gamma_{fss}) + b\dot{\gamma}_{ak} & \text{if } (\gamma_{ak} > \gamma_{fss}) \wedge (\tau_{ak} > 0) \\ -k(\gamma_{ak} - \gamma_{ess}) - b\dot{\gamma}_{ak} & \text{if } (\gamma_{ak} < \gamma_{ess}) \wedge (\tau_{ak} < 0) \\ 0 & \text{otherwise} \end{cases} \tag{17}$$

where k is the stiffness of the soft stop; b is the damping associated with the soft stop; and γ_{fss} and γ_{ess} are the angular positions of the flexion and hyperextension soft stops, respectively, at the knee. The composite assistive controller, is collectively described within each finite state i by summing the torque components enumerated in equations (1) through (17):

$$\boldsymbol{\tau}_{\mathbf{i}} = \sum_{j=1}^{4} \boldsymbol{\tau}_{\mathbf{ij}} \tag{18}$$

F. Structure of the State Machine

The switching conditions that describe movement between the finite states of the state machine are shown in Fig. 2. In particular, switching between sub-states 1a and 1b, or 3a and 3b, is based on a change in the sign of the knee angular velocity in the affected and unaffected swing leg, respectively. The controller switches from single-support to double-support states via detection of heel strike of the respective swing leg. Finally, the controller switches from double-support to swing (i.e., out of 2a or 2b) when the angular velocity of the respective thigh exceeds a given threshold.

III. Experimental Implementation and Assessment

A. Exoskeleton Prototype

The previously described assistive control approach was implemented on the Vanderbilt lower limb exoskeleton, Fig. 3a. [8]. The exoskeleton incorporates brushless DC motors and backdrivable transmissions at each of the four joints, is powered by a lithium polymer battery contained in the hip piece, and is used with a standard ankle foot orthosis (AFO). The combined weight of the exoskeleton and battery is approximately 12 kg.

B. Assessment Procedure

In order to provide a preliminary assessment of the efficacy of the exoskeleton controller, the authors implemented the assistive controller on the Vanderbilt exoskeleton, and conducted a preliminary evaluation on a subject with lower limb hemiparesis following stroke. Specifically, the subject was a 39-year-old female, 3 months post a left-side (i.e., right-affected) ischemic stroke at the time of the assessments. Prior to conducting the preliminary evaluations, the exoskeleton was fit to the subject, and the assistive control parameters were tuned such that the combined effort of the subject and exoskeleton achieved appropriate foot clearance during swing and knee stability during stance. Walking metrics were measured prior to using the exoskeleton, and immediately after doffing the exoskeleton. Three assessment metrics were utilized, including fast gait speed (FGS), step length asymmetry (SLA), and stride length (SL). Each session began with an approximately 5-minute warm-up which consisted of therapist-assisted overground walking. Following the warm-up period, each subject was allowed to rest if desired, after which the subject performed a ten meter walk test (10MWT). The subject was instructed to "walk as fast as you safely can" over a 14 m distance, with the middle 10 m segment being timed. Note that FGS was calculated using a stopwatch as the average speed during the (middle 10 m portion of the) 10MWT, while SLA and SL were both measured via video post-processing of the recorded 10MWT. SLA is defined as:

$$SLA = 1 - \frac{x_u}{x_a} \tag{19}$$

where x_u is the average step length of the unaffected leg, and x_a is the average step length of the affected leg. This definition of SLA is slightly modified from other similar definitions present in the literature to evaluate step length asymmetry [9, 10]. Specifically, in the definition given in equation (19), a smaller value indicates increased symmetry, while a larger value indicates reduced symmetry. A perfectly symmetric gait would have an SLA score of 0.

Following the "pre-session 10MWT" the subject donned the exoskeleton, and walked overground in the exoskeleton, with a physical therapist providing balance support as needed, as shown in Fig. 3b. Specifically, the subject walked for approximately 20-30 minutes, in approximately 5 minute segments, resting as needed between walking segments. Following the period of walking in the exoskeleton, the subject doffed the exoskeleton and conducted a post-session 10MWT. The full single-session protocol typically lasted approximately one hour. The single-session protocol was performed three times, each spaced three weeks apart to reduce the potential effects of carryover from previous sessions. The subject had baseline metrics, measured during the first pre-session 10MWT, of 0.33 m/s for FGS, 29% for SLA, and 88.7 cm for SL.

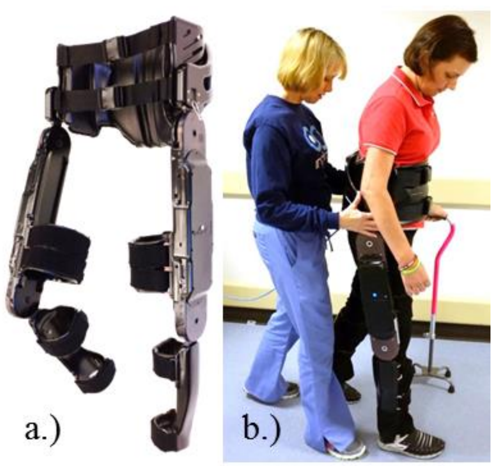

Fig. 3. a) Vanderbilt lower limb exoskeleton, and b) Experimental subject walking in the exoskeleton during a training session. A physical therapist offers assistance as needed.

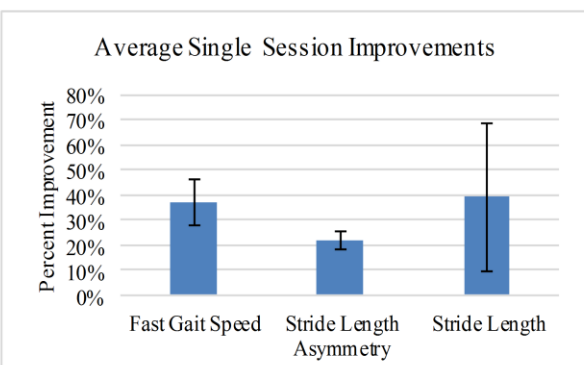

Fig. 4. Average single-session gains across all sessions for each measure. Error bars indicate plus/minus one standard deviation.

C. Single-Session Results

Single-session effects were assessed by comparing the pre-session and post-session measures of FGS, SLA, and SL, with the difference presumably attributed to the session of overground exoskeleton walking. Figure 4 shows the average improvement for each outcome measurement across the three trials. When comparing post-session to pre-session values, the percent change is indicated by the ratio of post and pre-session values of FGS and SL, while it is indicated by the difference between post and pre-session values of SLA, since SLA is already a ratio. The subject showed average improvements across the three trials of 37%, 22%, and 39% in FGS, SLA, and SL, respectively.

This preliminary assessment indicates the exoskeleton and assistive controller may have promise for assisting persons with hemiparesis in the recovery of walking. In future work, the authors will assess the exoskeleton and control method on additional subjects with hemiparesis following stroke.

IV. REFERENCES

[1] M. Kelly-Hayes, A. Beiser, C. S. Kase, A. Scaramucci, R. B. D'Agostino, and P. A. Wolf, "The influence of gender and age on disability following ischemic stroke: the Framingham study," *Journal of Stroke and Cerebrovascular Diseases,* vol. 12, pp. 119-126, 2003.

[2] V. L. Roger, A. S. Go, D. M. Lloyd-Jones, R. J. Adams, J. D. Berry, T. M. Brown*, et al.*, "Heart disease and stroke statistics—2011 update a report from the American Heart Association," *Circulation,* vol. 123, pp. e18-e209, 2011.

[3] L. Marchal-Crespo and D. J. Reinkensmeyer, "Review of control strategies for robotic movement training after neurologic injury," *Journal of NeuroEngineering and Rehabilitation,* vol. 6, Jun 16 2009.

[4] S. K. Banala, K. Seok Hun, S. K. Agrawal, and J. P. Scholz, "Robot Assisted Gait Training With Active Leg Exoskeleton (ALEX)," *Neural Systems and Rehabilitation Engineering, IEEE Transactions on,* vol. 17, pp. 2-8, 2009.

[5] H. Vallery, A. Duschau-Wicke, and R. Riener, "Generalized elasticities improve patient-cooperative control of rehabilitation robots," in *Rehabilitation Robotics, 2009. ICORR 2009. IEEE International Conference on*, 2009, pp. 535-541.

[6] J. L. Emken, S. J. Harkema, J. A. Beres-Jones, C. K. Ferreira, and D. J. Reinkensmeyer, "Feasibility of Manual Teach-and-Replay and Continuous Impedance Shaping for Robotic Locomotor Training Following Spinal Cord Injury," *Biomedical Engineering, IEEE Transactions on,* vol. 55, pp. 322-334, 2008.

[7] J. F. Veneman, R. Kruidhof, E. E. G. Hekman, R. Ekkelenkamp, E. H. F. Van Asseldonk, and H. van der Kooij, "Design and Evaluation of the LOPES Exoskeleton Robot for Interactive Gait Rehabilitation," *Neural Systems and Rehabilitation Engineering, IEEE Transactions on,* vol. 15, pp. 379-386, 2007.

[8] H. A. Quintero, R. J. Farris, C. Hartigan, I. Clesson, and M. Goldfarb, "A Powered Lower Limb Orthosis for Providing Legged Mobility in Paraplegic Individuals," *Topics in Spinal Cord Injury Rehabilitation,* vol. 17, pp. 25-33, 2011.

[9] J. H. Kahn and T. G. Hornby, "Rapid and long-term adaptations in gait symmetry following unilateral step training in people with hemiparesis," *Physical therapy,* vol. 89, pp. 474-483, 2009.

[10] C. K. Balasubramanian, M. G. Bowden, R. R. Neptune, and S. A. Kautz, "Relationship between step length asymmetry and walking performance in subjects with chronic hemiparesis," *Archives of physical medicine and rehabilitation,* vol. 88, pp. 43-49, 2007.