An Assistive Control Approach for a Lower-Limb Exoskeleton to Facilitate Recovery of Walking following Stroke

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***Abstract*— This paper presents a control approach for a lower-limb exoskeleton intended to facilitate recovery of walking in individuals with lower-extremity hemiparesis after stroke. The authors hypothesize that such recovery is facilitated by allowing the patient rather than the exoskeleton to provide movement coordination. As such, an assistive controller that provides walking assistance without dictating the spatiotemporal nature of joint movement is described here. Following a description of the control laws and finite state structure of the controller, the authors present the results of an experimental implementation and preliminary validation of the control approach, in which the control architecture was implemented on a lower limb exoskeleton, and the exoskeleton implemented in an experimental protocol on three subjects with hemiparesis following stroke. In a series of sessions in which each patient used the exoskeleton, all patients showed substantial single-session improvements in all measured gait outcomes, presumably as a result of using the assistive controller and exoskeleton.**

***Index Terms*—Assistive technology, cerebrovascular accident, gait rehabilitation, hemiparesis, lower-limb exoskeleton, rehabilitation robotics, stroke rehabilitation**

I. Introduction

EACH year approximately 800,000 people in the US suffer a stroke or cerebrovascular accident (CVA), of which approximately 660,000 survive [1]. Of these, approximately 200,000 annually are affected by lower-extremity hemiparesis to an extent that prevents walking without assistance six months after (i.e. by the time they enter the chronic stages of stroke) [2-4]. The inability to walk unassisted has an obvious impact on an individual's independence and community dwelling capability, and thus quality of life and continued health. Similarly, impaired balance and compromised walking ability increase the incidence of falls and resulting fractures [5-11].

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 (i.e., the potential of knee instability in flexion or hyperextension), and insufficient foot clearance on the paretic side during the swing phase of gait. In order to mitigate these deficits, post-stroke individuals typically employ compensatory actions. These include asymmetric spatial and temporal step lengths as well as a substantial frontal plane lean toward the non-paretic leg, both of which bias the individual away from loading the paretic leg in stance. Additionally, hip circumduction of the paretic leg during swing phase and ankle plantarflexion of the non-paretic ankle during stance (i.e., vaulting on the non-paretic leg), both facilitate foot clearance of the paretic leg during swing.

Given these biomechanical deficits exhibited by hemiparetic individuals, the biomechanical movement objectives of post-stroke gait training primarily entail improving load acceptance on the paretic leg during stance, which results in improved spatial and temporal step symmetry and generally greater stride length, and improving foot clearance of the paretic leg via increased hip and knee flexion of the paretic leg during swing. These therapeutic 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, which may be supplemented by assisted treadmill training. Two methods of assisted treadmill training are manually and robotically assisted body-weight-supported treadmill training (BWSTT). In the manual version of this therapy, a portion of a patient's body weight is suspended above a treadmill through an overhead suspension point, while one or more therapists manipulate a patient's pelvis and limbs as needed to facilitate treadmill walking. Robotic versions of this therapy incorporate robotic manipulation of the legs in place of manual manipulation. Such systems may provide more consistent

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interaction with a patient, and in most cases decrease the number of therapists required to provide BWSTT. As described in a recent review article [12], various methods have been proposed to control the patient-robot interaction in robotically-assisted BWSTT systems. Some representative methods include force-field-based control methods which guide the user along desired trajectories using simulated walls around a pre-selected footpath [13, 14]; record-and-replay impedance based methods to create subject specific trajectories [15]; and model-based methods which selectively target specific sections of the gait cycle [16, 17].

Recently, lower limb exoskeletons have begun to emerge. Unlike robotically assisted treadmill systems, lower limb exoskeletons are wearable robots, and as such enable overground rather than treadmill-based locomotion. Overground walking, particularly for severely hemiparetic individuals, can be characterized by a highly irregular gait speed, with considerable pauses between movements, as dictated by the movement volition, and balance and weight shifting needs of the individual. Treadmill-based systems can be adapted to provide adaptive speed capability (see, for example, [18], in addition to a large body of patent literature on the topic). Such systems, however, distort the dynamics of overground locomotion during periods of belt acceleration and deceleration when the belt speed changes. As such, for highly irregular gait, such as that which might be observed in a severely hemiparetic individual, a treadmill-based system is unable to accurately represent the dynamics of overground walking. In addition to resulting in unnatural perturbations in movement, the distortion in dynamics associated with an irregular belt speed also presents a distortion in vestibular information presented to the individual. The distortion in vestibular information, together with the associated lack of visual flow, further impairs the ability of a treadmill system to emulate overground walking with irregular gait speed.

In addition to limitations associated with reproducing the dynamics of highly-irregular gait, treadmill-based systems are typically limited with respect to their ability to provide assistive forces that are fully consistent with the biomechanics of locomotion and balance. Specifically, in order to provide assistance that is fully consistent with the biomechanics of locomotion and balance, the assistive forces between the environment and the individual can only occur between the individual's feet and the ground. Since a wearable exoskeleton (as defined here) has no attachment points to the inertial reference frame, it must react assistive components that it provides exclusively between the individual's feet and the ground (which is fully consistent with the biomechanics of locomotion and balance). Treadmill-based systems, conversely, typically entail at least one point of constraint between the individual and treadmill (i.e., inertial reference) frame beyond the foot/floor contact points. The constraint between the treadmill frame and the robot will introduce a constraint force that is not consistent with the biomechanics of overground locomotion and balance, and therefore can presumably interfere with the relearning of or recovery of balance. In the case of a manually-assisted treadmill system, this constraint is typically an overhead suspension point, which imposes body-weight support from the overhead point down, and as such introduces an artificially stabilizing effect. In the case of a robotically-assisted treadmill, the nature of this constraint depends upon the extent to which the robotic portion is constrained relative to the treadmill (and inertial reference) frame. If the robotic portion of the treadmill is fully unconstrained relative to the treadmill frame (i.e., the robotic portion is essentially an exoskeleton mechanically decoupled from the treadmill frame), then no artificial constraint forces will be present, and as such no artificial force components will interfere with balance dynamics (i.e., the body-weight support will be provided in a manner fully consistent with balance dynamics, less the irregular belt speed issue previously discussed). If however, the robotic portion is coupled to the treadmill frame by at least one kinematic constraint, the system will introduce at least one artificial component of force that is not representative of the balance dynamics entailed in overground standing and walking. Depending on the rehabilitation objectives, such constraints could be an asset. If relearning balance for purposes of overground standing and walking is the primary objective, however, these artificial constraints constitute a distortion of overground balance dynamics, which presumably can interfere with the relearning of such balance.

Despite the efficacy of the aforementioned control methods [12-17] in governing interaction between the patient and robot in robotically-assisted BWSTT systems, such methods are less well-suited to walking overground in an exoskeleton. Specifically, these control methods either dictate or substantially influence the spatiotemporal nature of leg movement or foot path (i.e., they have a substantive influence on either step length or step time). In the case of treadmill walking, desired step length and/or time is consistent and generally known. Further, the presence of overhead body-weight support mitigates the need to maintain balance. In the case of overground locomotion, however, enforcing or encouraging a given leg movement or footpath will generally present a balance perturbation, which may interfere with a patient's ability to select step length and/or time, and thus interfere with the ability of the user to maintain balance when walking. As such, a control methodology for gait assistance for an exoskeleton should ideally assist movement, without governing the spatiotemporal nature of the footpath, such that the patient is able to provide the movement coordination required to maintain balance (i.e., the patient must select a step length and time that maintains his or her zero-moment-point within his or her support polygon). In this manner, the system facilitates balance recovery, and avoids substantial balance perturbations. This paper describes a control approach that provides this objective. Specifically, the approach provides floor-referenced walking assistance without substantially affecting a 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. Specifically, coordination is considered a mapping between sensory input and motor output in the sense of a neural network, wherein weights in the neural network are incrementally adjusted based on iterative error correction. Consistent with a Hebbian model of learning (i.e., “neurons that fire together wire together”), adjustment of synaptic weights requires associating an afferent pattern of neural information with an efferent response. Thus, it is conjectured that having the patient provide movement coordination, and allowing the patient to incur and correct for errors in that coordination, will facilitate neural recovery (i.e., will facilitate the formation of appropriately weighted coordination maps). As such, the objective of the control approach presented here is to provide to the patient movement assistance (to compensate for muscle weakness and to enhance stability), without providing a desired movement path or trajectory.

The resulting controller, described subsequently, consists of the combination of three types of behaviors: gravity compensation, feedforward movement assistance during swing, and knee joint stability reinforcement during stance. The gravity compensation component consists of two sub-components: full gravity compensation for the mass of the exoskeleton, and partial gravity compensation for the patient’s leg mass during the swing phase of gait. The feedforward movement assistance consists of torque pulses that assist weak muscle groups when initiating or reversing joint movement at the beginning or middle of swing phase, as needed by the individual. The knee joint stability reinforcement takes the form of emulated spring-damper elements (similar to those used to simulate surfaces in haptic interfaces), which mitigates knee instability in flexion or hyperextension during the stance phase of gait. With regard to the previously stated control objectives (i.e., providing movement assistance without providing coordination or trajectory control), the gravitational components involve no prescribed trajectories. The torque pulse components during swing provide non-trajectory-based movement assistance, and specifically supplement movement already initiated by the user and vanish well in advance of the end of the respective movements. Finally, the knee joint stability reinforcement is a passive component that prevents knee joint buckling during stance, but otherwise involves no prescribed time-basis or trajectories. Thus, the combination of these control components provides the user with movement assistance, but relies entirely on the user to provide the coordination for movement (e.g., to select step length and time). The control approach also relies entirely on the user to initiate all movement. If the user is not constantly initiating movement, the user and exoskeleton will not move. Thus, the control approach relies on the user to be fundamentally engaged in the walking activity, and to provide appropriate coordination for it. The respective components of the control approach, and the state machine within which they operate, are described in the following sections.

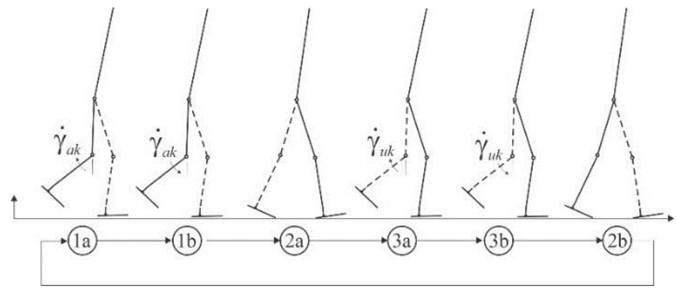

Fig. 1. Finite states corresponding to the assistive controller, where the affected leg is shown as a solid line and the unaffected leg as a dashed line. The three main states correspond to the 1) affected leg in swing, 2) double-support, and 3) unaffected leg in swing.

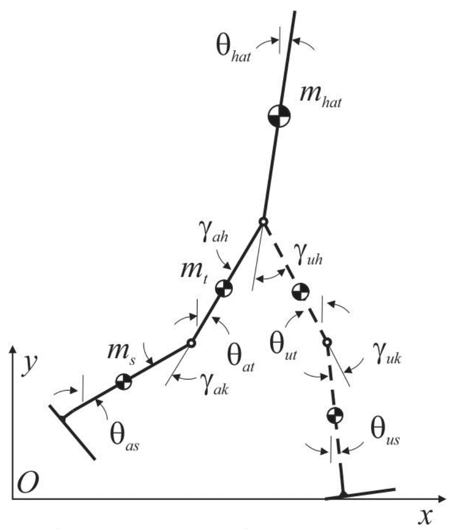

Fig. 2. Configuration parameters for assistive control approach.

A. Control States and Notation

The exoskeleton controller is governed by a finite state machine consisting of six states, as illustrated in Fig. 1. Specifically, Fig. 1 depicts the exoskeleton configuration corresponding to each state, where the affected leg is shown as a solid line, and the unaffected leg as a dashed line. The six states of the state machine are comprised of three primary configurations as follows: state 1 corresponds to the swing phase of the affected leg; state 2 corresponds to the double-support phase of walking; and state 3 corresponds to the swing phase of the unaffected leg. Each state is further comprised of two sub-states, as follows: sub-state 1a corresponds to the portion of swing in which the affected knee is in a state of flexion; sub-state 1b corresponds to the portion of swing in which the affected knee is in a state of extension; sub-state 2a corresponds to double-support following heel strike of the affected leg; sub-state 2b corresponds to double-support following heel strike of the unaffected leg; sub-state 3a corresponds to the portion of swing in which the unaffected knee is in a state of flexion; and sub-state 3b corresponds to the portion of swing in which the affected knee is in a state of extension. The sequence of states through which the controller would transition under normal walking conditions is illustrated in Fig. 1. As per the subsequently described experimental implementation, 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} and τ_{ah} are the torque commands corresponding to the affected knee and hip joints, respectively, and τ_{uk} and τ_{uh} are the torque commands corresponding to the unaffected knee and hip joints, respectively. Since as previously mentioned the system is described by three configurational states, each with two sub-states, the torque vector within the i^{th} state can be denoted by $\boldsymbol{\tau}_i$. For cases in which the control torque changes as a function of sub-state, the torque commands can be further indicated by $\boldsymbol{\tau}_{ia}$ or $\boldsymbol{\tau}_{ib}$, corresponding to the appropriate sub-state. Within each state, the control torque may consist of the combination of multiple assistive torque components. If each assistive component of torque is identified by the subscript *j*, the composite control torque can be denoted by $\boldsymbol{\tau}_{ij}$. 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 referring to Fig. 2, is described by the following control law

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

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

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

where θ_{as} and θ_{at} are the angles with respect to the vertical of the affected shank and thigh segments, respectively; 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; g is the magnitude of the gravitational acceleration; and *c* is an abbreviation for the cosine function. Note that the mass of the hip segment is shared equally between the two legs in the double support phases of gait (i.e., state 2). Note also that the gravity compensation described by this control law assumes that movement occurs principally in the sagittal plane (i.e., neglects out-of-plane movements). Finally, note that in the single-support phases (states 1 and 3), the contralateral limb must provide reactive torques, since the gravitational loads are ultimately reacted through the support foot by the ground. Finally, note that this component of the control law does not vary with sub-state.

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, and therefore clearance between the foot and ground during the swing phase of gait. 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. Since the weight of the limb assists movement when movement of the limb is in the direction of gravity (i.e. when gravity is performing positive work on the limb), active compensation during these phases could potentially increase the energetic output required by the user. As such, the partial limb weight compensation component is only exerted by the controller when the control torque works against the energy gradient (i.e., when the exoskeleton joint is generating power), and is zeroed when the control torque is along the energy gradient (i.e., when the exoskeleton joint absorbs power). As such, the partial limb weight compensation controller is described by:

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

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

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

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

where γ_{ak} and γ_{ah} are the joint angles of the affected knee and hip joints, respectively, as identified in Fig. 2; 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 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. Note that the authors chose not to provide the corresponding reactive torques on the stance side of the exoskeleton, since it was assumed that these loads were most appropriately reacted by the user's unaffected leg (i.e., they would be reacted by the unaffected leg in the case that the affected leg was not in a weakened state).

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 swing-phase motion at the hip and knee joints, depending on the level of impairment in the affected limb, and also on the level of spasticity or tone present in the limb. Insufficient swing-phase motion at the hip and knee joints can consequently result in foot dragging during mid-swing, reduced step length, or inability to fully extend the knee prior to heel strike. In order to provide additional assistance without dictating joint trajectories, a control component is available to provide hip or knee joint torque pulses at the initiation of swing, and/or during mid-swing when the knee changes its direction of rotation from flexion to extension. Specifically, in order to avoid providing trajectory-based assistance, 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:

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

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

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

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

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

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

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. Knee Joint Stability Reinforcement during Stance

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

$$\boldsymbol{\tau}_{\mathbf{14}} = \boldsymbol{\tau}_{\mathbf{34}} = \begin{bmatrix} 0 & 0 & 0 & 0 \end{bmatrix}^T \tag{15}$$

$$\boldsymbol{\tau}_{\mathbf{24}} = \begin{bmatrix} \tau_{ak} & 0 & 0 & 0 \end{bmatrix}^T \tag{16}$$

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

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, which provides the movement assistance components as described individually above, is collectively described within each finite state i by summing the torque components enumerated in equations (1) through (17):

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

Recall that the subscript i in (18) represents the i^{th} state of the state machine, where i represents one of 6 states (1a/b, 2a/b, or 3a/b) as illustrated in Fig. 1 and discussed in the following section.

F. Structure of the State Machine

The switching conditions that describe movement between the finite states of the state machine are shown in Fig. 3. 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, as measured by angular encoders at the respective knee joints.

The controller switches from single-support to double-support states via detection of heel strike of the respective swing leg, which can be detected when the acceleration aligned with the respective leg, as measured by an accelerometer, exceeds a given threshold. Finally, the controller switches from double-support to swing (i.e., out of 2a or 2b) when the angular velocity of the respective thigh, as measured by a gyroscope, exceeds a given threshold (i.e., the user initiates swing by accelerating the thigh forward, until it reaches a detectable angular velocity).

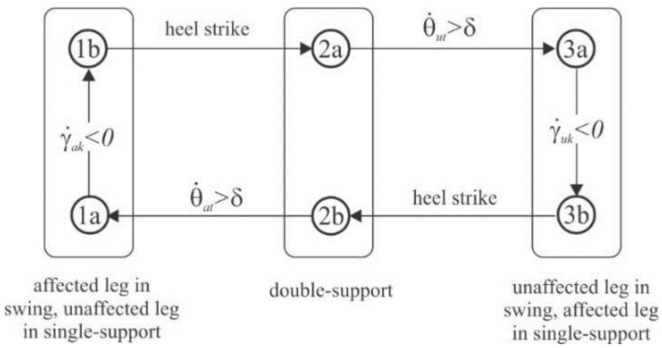

Fig. 3. Finite state machine switching conditions corresponding to the assistive controller.

III. Experimental Implementation and Preliminary Assessment

A. Exoskeleton Prototype

The previously described assistive control approach was implemented on the Vanderbilt lower limb exoskeleton, which is shown in Fig. 4. Design of the exoskeleton was previously described in the context of providing legged mobility for individuals with paraplegia [19, 20]. The exoskeleton incorporates four control actuators (brushless DC motors acting through speed reduction transmissions) that provide sagittal-plane torques at the right and left hip and knee joints (relative to the exoskeleton frame). The control actuators are capable of providing continuous torques at each joint of approximately 20 Nm, and peak torques of approximately 80 Nm for durations on the order of a few seconds (thermally limited). The exoskeleton is used with ankle foot orthoses (AFOs), which provide stability at the ankle joints and transfer the weight of the exoskeleton to the ground. Instrumentation (for measurement of configuration angles, Fig. 2, and of state machine switching conditions, Fig. 3) include absolute and incremental encoders at each joint, and one six-axis inertial measurement unit (IMU) in each thigh link (i.e., two total). The exoskeleton is powered by a 30 v, 120 W-hr lithium polymer battery with a mass of approximately 600 g. The total mass of the system, including the battery, is approximately 12 kg (26.5 lb).

Fig. 4. Vanderbilt lower limb exoskeleton.

TABLE I
BASELINE CHARACTERISTICS OF STROKE SUBJECTS

Subject	1	2	3
Age (yrs)	39	42	69
Mos Post-Stroke	3	10	17
Affected Side	Right	Left	Right
Stability Aids Used	Quad Cane, R AFO	Quad Cane, L AFO	Quad Cane, R AFO
Baseline FGS (m/s)	0.33	0.07	0.19
Baseline SLA (%)	29	115	27
Baseline SL (cm)	88.7	33.2	66.3

B. Preliminary Assessment Procedure

In order to provide a preliminary assessment of the efficacy of the exoskeleton controller, and in particular to assess the appropriateness and potential of the assistive controller to facilitate walking in individuals with lower limb hemiparesis following stroke, the authors implemented the assistive controller on the Vanderbilt exoskeleton, and conducted a preliminary evaluation on three human subjects with lower limb hemiparesis following stroke. Relevant information regarding each subject is summarized in Table I. Prior to conducting the preliminary evaluations, the exoskeleton was fit to each subject, and the assistive control parameters incorporated in equations (5), (10), (11), (13), and (17) were tuned according to each individual subject's needs, with the parameter tuning guided by a combination of physical therapist and subject input, such that once appropriately adjusted, the combined effort of the subject and exoskeleton achieved appropriate foot clearance during swing and knee stability during stance (as judged by the therapist). Specifically, the proportion of limb weight compensation r (5) was initialized at zero and iteratively incremented until appropriate hip flexion was achieved in swing. Note that a torque pulse at the hip in early swing, as given by P_{hf} and T_{hf} (11) could similarly be used to supplement hip flexion in swing, but was not employed for the assessments described here. The swing phase knee flexion torque pulse parameters P_{kf} and T_{kf} (10) were initialized at zero and iteratively incremented until appropriate knee flexion was achieved in early swing. Similarly, the swing phase knee extension pulse parameters P_{ke} and T_{ke} (13) were initialized at zero and iteratively incremented until appropriate knee extension was achieved in late swing. Finally, the stance knee soft stop locations γ_{fss} and γ_{ess} (17) were adjusted to provide a small range of unencumbered motion around a neutral angle of the knee during stance, prior to engaging the virtual soft stops. In this manner, each individual subject was required to provide knee stability during the stance phase of gait, with the exoskeleton providing support only when the knee travelled outside of this range. The angle of engagement of the soft stops were established based on the collective comfort level of the subject and physical therapist regarding an appropriate range of knee movement prior to engaging exoskeleton support. In particular, two of the subjects were comfortable with an unencumbered range of motion between zero and 8 deg (flexion), while one subject (whose knee was particularly prone to instability) preferred a range between 2 and 6 deg flexion. Because the level of impairment varied between patients, parameter selection was largely informed by what the physical therapist believed was an appropriate level of device assistance for each individual patient, rather than by predetermined goals for gait performance. Note that the stiffness and damping of the soft stops were determined by the investigators when constructing the controller, and therefore were not among the tunable parameters. Also, gravity compensation parameters (2-4, 6-7) were measured or estimated, and as such were not among the tunable parameters. The values for all tunable parameters used in the experiments for each subject are given in Table II.

Once the assistive controller was suitably parameterized for each subject, a series of preliminary assessments were conducted. In particular, the preliminary assessments evaluated single-session gains in walking achieved by each subject in three separate therapy sessions. The nature of each session involved the subject walking overground with the exoskeleton (with assistive controller) for a period of approximately 30 min. Walking metrics were measured at the beginning of each session (i.e., prior to using the exoskeleton), and at the end of each session (i.e., 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 (without the exoskeleton), during which each subject used his or her standard stability aids (in all cases this consisted of a quad-cane and unilateral AFO, see Table I). Following the warm-up period, each subject was allowed to rest if desired, after which the subject performed a ten meter walk test (10MWT). Subjects were instructed to "walk as fast as you safely can" over a 14 m distance, with the middle 10 m segment being timed to determine FGS.

TABLE II
TUNABLE CONTROL PARAMETERS FOR EACH SUBJECT

Subject	1	2	3
r	0.20	.85	0.75
P_{kf} (Nm)	17	0	17
T_{kf} (s)	0.5	0.5	0.8
P_{hf} (Nm)	0	0	0
T_{hf} (s)	0.5	0.5	0.8
P_{ke} (Nm)	10	5	12
T_{ke} (s)	0.5	0.5	0.8
γ_{fss} (deg)	8	6	8
γ_{ess} (deg)	0	2	0

Fig. 5. Experimental subject walking in the exoskeleton during a training session. A physical therapist offers assistance as needed.

Following this "pre-session 10MWT" the subject donned the exoskeleton, and walked overground in the exoskeleton, with a physical therapist providing balance assistance as needed (i.e., contact guard assist), as shown in Fig. 5. All subjects used a quad-cane when walking with the exoskeleton, as per their respective standard practices when walking without the exoskeleton. Subjects walked for approximately 20-30 minutes, in approximately 5 minute segments, resting as needed between walking segments. Figure 6 shows the hip and knee joint angles recorded on the paretic leg of subject 1 during a representative therapy session, averaged over ten consecutive strides, in addition to the hip and knee joint torque and power delivered by the exoskeleton. In the plots, positive angles indicate flexion and negative extension; positive torques indicate flexive, and negative extensive; and positive power indicates the exoskeleton is providing power to the

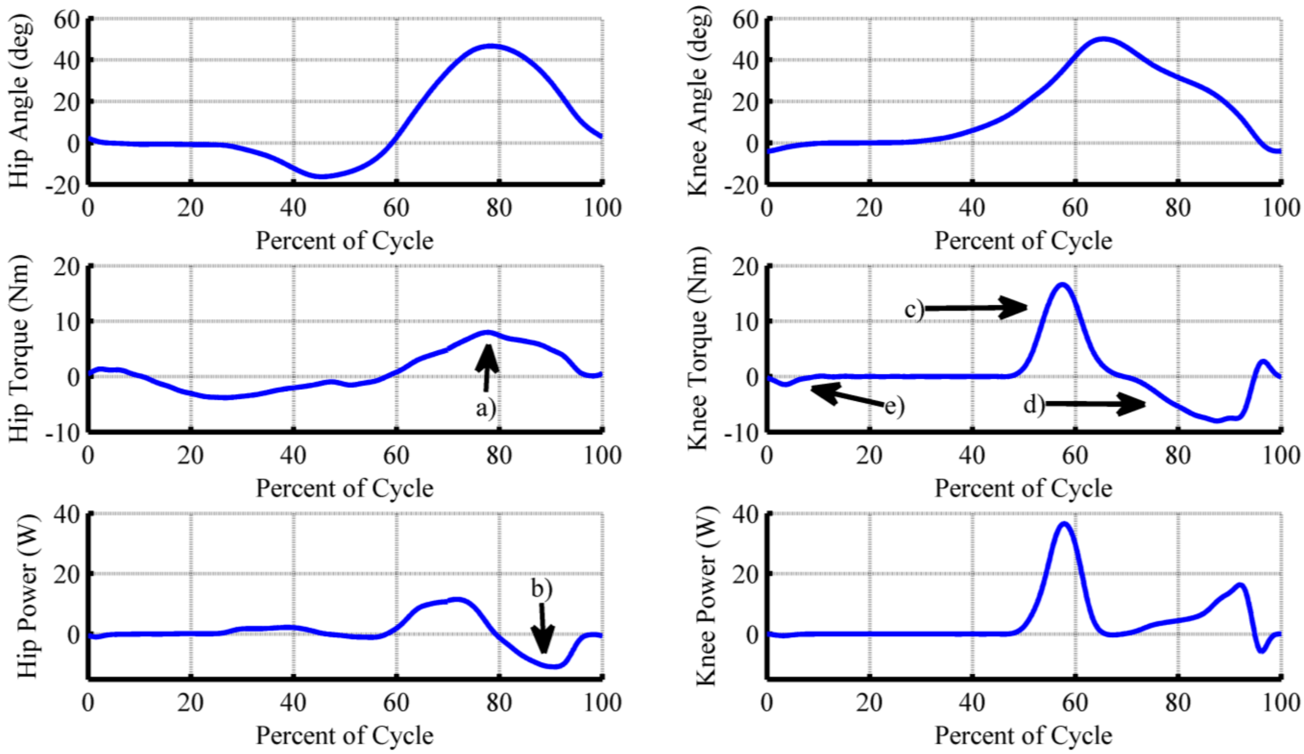

Fig. 6. Paretic leg hip and knee joint angles during exoskeleton walking from therapy session with subject 1, averaged over ten strides, and the associated torque and power at both joints imparted by the exoskeleton.

subject, while negative power indicates the exoskeleton is dissipating power. This data provides some indication of the nature of interaction between the exoskeleton and the subject. Some of the control components as indicated in the plots include: a) flexive hip torque associated with gravity compensation; b) power dissipation associated with gravity compensation of the exoskeleton mass; c) flexive knee torque associated with feedforward flexion assistance in early swing; d) extensive knee torque associated with feedforward extension assistance in mid swing; and e) knee joint torque assistance associated with the knee joint stability component during stance (i.e., immediately following heel strike). Note that in general the exoskeleton generates and dissipates power at different periods of the gait cycle, but on average provides net power to the user (i.e., on average is assistive rather than resistive).

Following the period of walking in the exoskeleton, the subject doffed the exoskeleton and conducted a post-session 10MWT. Note that both the pre-session and post-session 10MWT were conducted without the exoskeleton. The full single-session protocol typically lasted approximately one hour. For each of the three subjects, the aforementioned single-session protocol was performed three times, each spaced three weeks apart to reduce the potential effects of carryover from previous sessions.

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. 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 [21, 22]. Specifically, in the definition given in (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, while an exact “step-to” gait (i.e. the unaffected limb is brought even with the affected limb during swing) would have an SLA value of 1. 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. Figure 7 shows the average improvement for each outcome measure across the three trials, grouped by subject. As is evident in Fig. 7, all subjects showed improvements in all outcome measures in each of the trials. Figure 8 shows the single-session improvements for each outcome measure averaged across all subjects. Subjects demonstrated average improvements of 26%, 26%, and 30% in FGS, SLA, and SL, respectively.

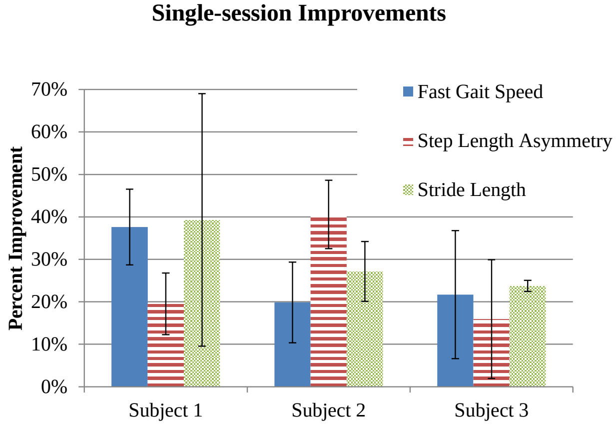

Fig. 7. Average single-session gains across all sessions for each measure grouped by subject. Error bars indicate plus/minus one standard deviation.

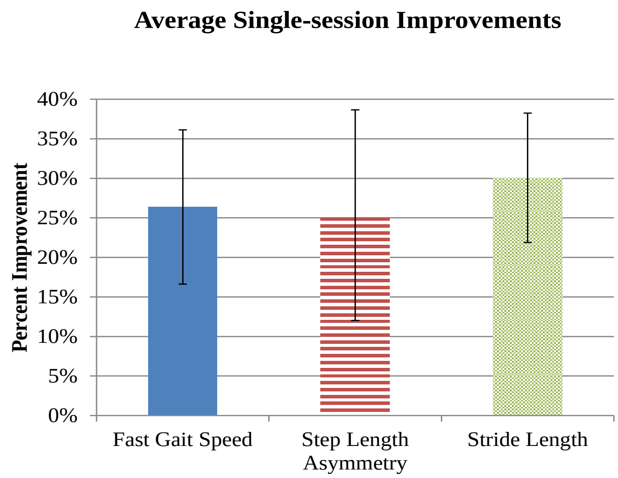

Fig. 8. Average single-session gains for each outcome measure averaged for all subjects and all sessions. Error bars indicate plus/minus one standard deviation.

IV. Conclusion

The authors present the implementation of an assistive controller for a lower limb exoskeleton, intended to facilitate recovery of walking function to persons with hemiparesis following stroke. The authors hypothesize that such recovery is facilitated by allowing the patient rather than the exoskeleton to provide movement coordination. As such, the objective of the control approach presented here is to provide to the patient movement assistance, without providing a desired joint angle path or trajectory. Accordingly, the authors developed and describe here a controller that provides walking assistance to the user, without dictating the spatiotemporal nature of a given movement, such that the user is required to provide the coordination of movement. In order to provide a preliminary assessment of efficacy, the authors implemented the controller on an exoskeleton prototype, and studied single-session improvements in walking in three subjects with lower limb hemiparesis following stroke. All subjects showed substantial improvement in three walking metrics in all sessions, indicating that the assistive control approach may have promise with respect to facilitating walking recovery. Future studies with a larger number of subjects and with longer periods of dosing will be required to fully assess the efficacy of such a system in providing recovery of walking following stroke.

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