Bae J.; De Rossi S.M.M.; O'Donnell K.; Hendron K.L.; Awad L.N.; Teles Dos Santos T.R.; De Araujo V.L.; Ding Y.; Holt K.G.; Ellis T.D.; Walsh C.J., "A soft exosuit for patients with stroke: Feasibility study with a mobile off-board actuation unit," *Rehabilitation Robotics (ICORR), 2015 IEEE International Conference on*, 11-14 August 2015.

A soft exosuit for patients with stroke: Feasibility study with a mobile off-board actuation unit

Jaehyun Bae[1,2], Stefano Marco Maria De Rossi[1,2], Kathleen O'Donnell[1,2], Kathryn L. Hendron[3], Louis N. Awad[1,2], Thiago R. Teles Dos Santos[1,2], Vanessa L. De Araujo[1,2], Ye Ding[1,2], Kenneth G. Holt[3], Terry D. Ellis[3], Conor J. Walsh[1,2]

[1] The Wyss Institute for Biologically Inspired Engineering, Harvard University, Cambridge, MA, 02138
[2] School of Engineering and Applied Science, Harvard University, Cambridge, MA, 02138
[3] Sargent College of Health & Rehabilitation Science, Boston University, Boston, MA, 02215

***Abstract*— In this paper, we present the first application of a soft exosuit to assist walking after stroke. The exosuit combines textile garments with cable driven actuators and is lighter and more compliant as compared to traditional rigid exoskeletons. By avoiding the use of rigid elements, exosuits offer greater comfort, facilitate donning/doffing, and do not impose kinematic restrictions on the wearer – all while retaining the ability to generate significant moments at target joints during walking. The stroke-specific exosuit adapted from previous exosuit designs provides unilateral assistance to the paretic limb during walking. This paper describes stroke-specific design considerations, the design of the textile components, the development of a research-focused, mobile off-board actuation unit capable of testing the exosuit in a variety of walking conditions, a real-time gait detection and control algorithm, and proof-of-principle data validating the use of the exosuit in the chronic stroke population. Ultimately, we demonstrate reliable tracking of poststroke gait, appropriate timing of assistive forces, and improvements in key gait metrics. These preliminary results demonstrate the feasibility and promise of exosuits for poststroke gait assistance and training. Future work will involve the creation of a portable, body-worn system based on the specifications obtained from such feasibility studies that will enable community-based rehabilitation.**

Keywords— wearable robot; exosuit; gait training; stroke rehabilitation; human-robot interaction

I. INTRODUCTION

Stroke is a leading cause of long-term disability, with over 33 million stroke survivors worldwide [1]. Over 80% of stroke survivors live with residual gait impairments despite extensive therapy [2]. In particular, gait after stroke is often characterized by a marked asymmetry [3], with the paretic side contributing little to forward progression and commonly exhibiting signs of neuromotor impairment such as foot drop, ankle and knee instability, and/or weakness at the hip [4][5]. These impairments often result in the development of compensatory patterns and a slow and energetically inefficient gait, causing more than 50% of stroke survivors to ambulate at a speed slower than that considered necessary for safe community ambulation (i.e. 0.8m/s) [6]. Such motor impairments also contribute to reduced physical activity and quality of life after stroke, and ultimately an increased risk for a second stroke and mortality.

The need for better walking therapies after stroke is apparent, and means for delivering effective rehabilitation in a community setting have gained recent popularity [6][7]. Exoskeleton technology is one of a number of potential solutions being explored to address this need [8]–[10]. Exoskeletons often come in two forms – clinic-based gait training systems or portable exoskeletons. Clinic-based systems, which are often treadmill-bound [9], [11]–[13] aim to complement traditional labor-intensive gait rehabilitation. Portable exoskeletons, on the other hand, have gained significant momentum in the last 5 years due to their ability to provide gait assistance both in the clinic and in community settings. Some of these systems [14] have recently received FDA approval and have been commercialized for personal

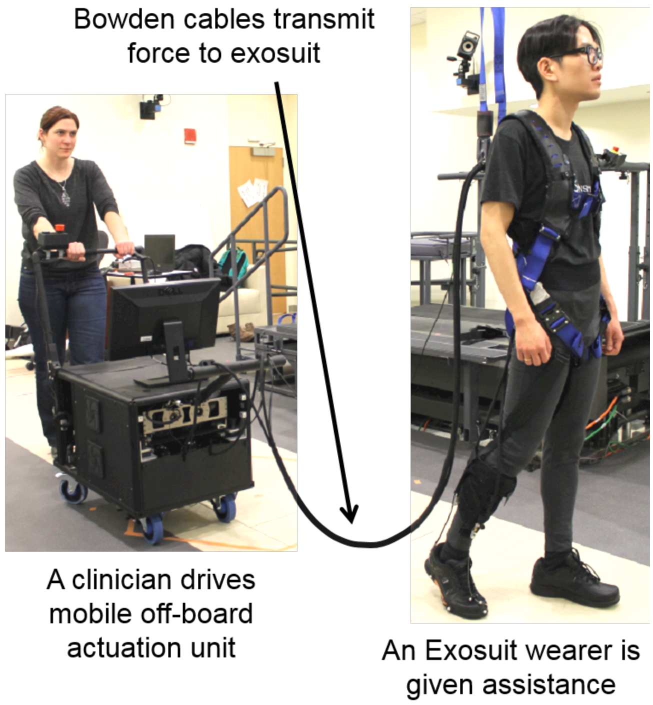

Figure 1. Overground walking wearing an exosuit actuated by a mobile off-board actuation unit. One end of the Bowden cable is connected to the off-board actuation unit, and the other end is connected to the exosuit. The off-board actuation unit contracts the Bowden cables by changing the motor position, resulting in torque generated at the ankle. The blue safety harness worn by the user in the figure is not the part of exosuit.

use. These rigid exoskeletons offer an incredible opportunity for mobility and independence in those needing total assistance – e.g. individuals who have suffered a spinal cord injury or for those with severe impairments following a stroke. However, the large majority of those after stroke regain the ability to walk – albeit with residual gait impairments. For these patients, rigid exoskeletons may not be an effective solution. Indeed, rigid exoskeletons unavoidably add significant mass to the wearer's limbs and impose restrictions on their kinematics, ultimately imposing a slow and inefficient gait that may not be better than a patient's baseline gait. Moreover, the additional significant mass attached to the wearer may cause injuries and pose a significant safety concern should a fall occur. Furthermore, misalignments of the exoskeleton's rigid joints with the user's biological joints can cause significant stress on the bones and soft tissue. Lastly, rigid exoskeletons face several practical challenges that may limit their adoption in a community setting. For example, the power requirements for such devices are relatively high and they would become impossible to transport by patients should they lose power.

To alleviate some of these challenges, recent efforts have focused on the development of soft exosuits that use soft materials such as textiles and elastomers to provide a more conformal, unobtrusive, and compliant means to interface to the human body [15]–[24]. Previous work has shown that exosuits can securely and comfortably transmit forces to the wearer via textiles, actuators, and cable-based transmissions and generate biologically appropriate moments to assist with locomotion. The textiles are structured in a manner that anchors the exosuit to key locations on the body (e.g. pelvis, thigh, calf, and foot) and provide load paths that generate tensile forces across joints when an actuated segment reduces the relative length between two points in the suit. Depending on the load path of the textile, assistive torques can be provided at a single joint, or simultaneously across multiple joints, in a way that mimics the mono-articular and multi-articular function of biological muscles. Our group has conducted preliminary human subject testing on a multi-articular exosuit that generates torques to assist ankle plantarflexion and hip flexion simultaneously. We demonstrated that an exosuit can accurately deliver high forces to healthy subjects (up to 250N) walking at 1.25m/s in a manner that produces an average net metabolic cost reduction of 6.4% and does not interfere with normal sagittal plane kinematics when comparing exosuit worn powered versus unpowered [20]. Based on these results, it is clear that an exosuit can provide a positive effect to healthy wearers while allowing them to maintain their natural gait. These encouraging early results prompted study of the exosuit in the poststroke population.

From a clinical perspective, exosuits offer the unique opportunity for gait assistance across the continuum of care and have the potential to reshape the orthotic and rehabilitation landscape. For example, the exosuit can be used by a therapist to provide targeted assistance to individual joints during walking on the treadmill or overground in a manner similar to what is done in the clinic today, but in a more controlled and less labor intensive way.

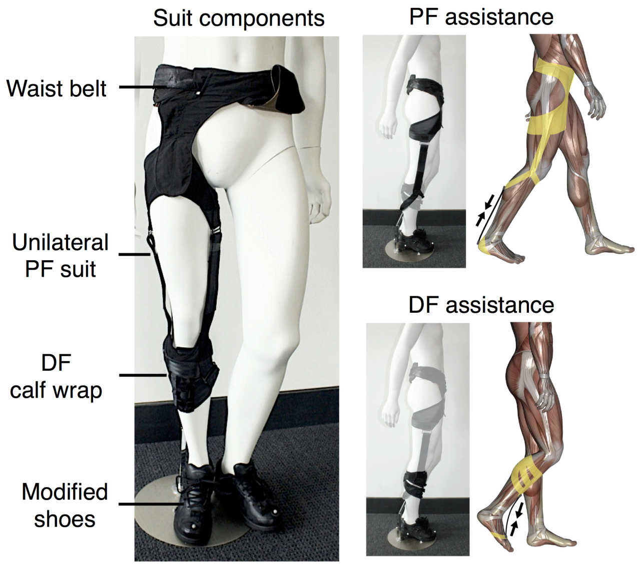

Figure 2. Stroke-specific exosuit textiles (Unilateral plantarflexion (PF) suit module, dorsiflexion (DF) suit module, and modified shoes) are worn on mannequin (Left). The new exosuit can assist PF by contracting a Bowden cable between the calf and the heel (Right top), and assist DF by contracting a cable between the anterior shank and the dorsal surface of the foot (Right bottom). Mediolateral cable alignment can be modified to provide additional assistance for weakness in foot inverters and everters.

In addition, due to their simple, unobtrusive, low power, highly portable, and lightweight design, exosuits offer a unique platform for continuous, targeted gait rehabilitation in the community. In particular, through the provision of well timed, small to moderate assistive forces to key joints (e.g. the ankle) during walking, we hypothesize more physiological and energy-efficient ambulation that encourages greater mobility and activity after stroke.

In the subsequent sections of this paper, we describe the first prototype of a unilateral soft exosuit for persons poststroke along with a mobile off-board actuation unit developed to test the feasibility of the clinical application of poststroke gait assistance on three patients in the chronic phase of recovery. Specifically, section II presents the design principles that governed the development of this first embodiment of a stroke-specific exosuit. In section III, we present the exosuit hardware, including the soft exosuit textiles and a mobile off-board actuation unit developed to facilitate laboratory-based feasibility testing of the exosuit. Section IV describes sensor implementation, a gait event detection algorithm, and a control strategy for assisting poststroke gait that facilitates input from clinicians. Finally, section V presents a preliminary experimental validation of the exosuit on three stroke patients.

II. Soft Exosuit for Poststroke Gait Assistance

When compared to rigid exoskeletons, exosuits offer three distinct advantages that increase their usability in the poststroke population. First, through the proximal-mounting of motors, electronics, and batteries (i.e. actuator packs) and flexible cable-based transmissions to the lower extremity, exosuits minimize the addition of mass on the extremities.

This allows for the components on the legs to be very light and non-restrictive – an important consideration for stroke survivors as any added mass or constraints could negatively impact their residual mobility and create discomfort/safety risks. Through the use of unobtrusive and conformal materials, exosuits also have a very low profile and can thus be worn underneath regular clothing. Considering that a major goal for such wearable technology is use in the community, a non-visible wearable robot is advantageous for aesthetic and psychosocial reasons. Lastly, due to their garment-like design, exosuits can be easily donned/doffed – another important consideration for those poststroke in whom upper extremity impairments are prevalent.

The stroke-specific exosuit we present in this paper deviates from our previous designs [17]–[24] by providing only unilateral assistance to the paretic ankle during push off, and additionally provides dorsiflexion assistance during the swing phase via a new textile module. Poststroke gait impairments vary widely, and separate suit modules for different assistance targets provide the flexibility to address a wide variety of gait deficits. For example, a suit module for dorsiflexion assistance can prevent foot drop and facilitate better ground clearance while a module for plantarflexion assistance can improve the paretic limb's ability to generate forward propulsion. Moving to a unilateral design allows a substantial reduction in the weight of the total system while not sacrificing efficacy. Indeed, improving the function of the paretic limb – especially by targeting two key subtasks of walking (i.e. ground clearance and forward propulsion) known to be impaired after stroke [4][5] – can yield bilateral improvements in gait such as improved kinematic and kinetic symmetry. It should be noted that although an exosuit provides a platform for targeted unilateral assistance of the paretic limb, it could also be used bilaterally.

For exosuits to be practical in a community setting, it will be important for them to have a low overall mass and in particular a low-profile body-worn actuation system. The requirements for a body-worn actuation system (e.g. how much assistance to provide) and the sensors and controls needed to achieve these requirements are currently not known. Thus, as a first step, we have designed a mobile off-board actuation unit that enables feasibility testing with an exosuit both on the treadmill and overground. This actuation unit mimics the function of a body-worn actuation system, enabling the experimental testing of different sensing and control strategies. Our goal is to leverage this basic science research to guide the future development of body-worn actuators, textiles, and controllers.

III. Textile and Actuation Implementation

A. *Functional textiles*

As a first step toward the development of exosuits to assist walking after stroke, we implemented new unilateral 2 degrees of freedom (DoF) exosuit textiles. These textiles consist of two separate modules: a unilateral plantarflexion (PF) suit module and a dorsiflexion (DF) suit module (Fig 2).

The PF suit module was adapted from previous multi-articular exosuits described in [18]–[20]. A waist belt anchors the suit to the pelvis and supports most of the downward forces. Unlike previous exosuits, the waist belt of the new PF suit module is fabricated from highly stable woven textiles (Safety Components, Greensboro, NC) instead of polyester webbing straps to improve suit comfort and fit around the waist. These were deemed important to improve exosuit fit across diverse body types and shapes in persons poststroke. The connecting straps and thigh brace are routed to transmit the force from the waist to the leg in a manner that minimizes pressure on the skin and unwanted joint moments (e.g. the knee moment). The Bowden cable sheath attaches to the distal end of the suit near the calf, whereas its inner cable attaches at the heel of standard walking shoes (New Balance, Boston, MA) that were modified to include attachment points. With this multi-articular suit architecture, the cable contraction generates ankle plantarflexion and hip flexion torques simultaneously.

We developed the DF suit module to assist with ankle dorsiflexion because improving paretic limb ground clearance during swing phase is a stroke-specific requirement. The DF suit module consists of a single vertical reinforced strap on the shin connected to three separate straps that route forces posteriorly around the calf to a common anchor point located anteriorly at the tibial tuberosity. Rotational pivots on each strap enable a customized fit across a wide range of calf shapes, and overlapping hook and loop panels secure the straps in place to improve anchoring. Similar to the PF suit module, the Bowden cable sheath attaches at the lowest part of the DF suit module and its inner cable attaches on the front of the shoe over the metatarsophalangeal joint. Thus, with a contraction of the cable, dorsiflexion torque is generated. Furthermore, the cable can be aligned medial or lateral to the subtalar joint to additionally assist eversion or inversion of the foot when needed. The overall mass of the textile portion of the exosuit, which accounts for most of the weight imposed on the lower limbs, is less than 300g, making it significantly lighter than the lower body structure of rigid exoskeletons.

To date, the suit has been worn by several individuals in the chronic phase of stroke recovery. The feedback provided by these patients has led to the development of exosuit components (see Fig 2) that are customizable across a variety of body types. Because the suit anchors to the body via functional textiles that distribute forces evenly, pressure points are minimized and the suit is generally well tolerated by individuals. Participants have noted that the suit is snug, but still comfortable, around the paretic limb. Over time, participants generally report that this sensation of tightness is no longer perceivable. These reports of suit tolerance have been independently confirmed through skin checks performed by a physical therapist before and after prolonged suit wear.

B. *Mobile Off-board Actuation Unit*

A reconfigurable multi-joint actuation platform was previously presented to explore human-exosuit interaction in an efficient and controlled manner [22][25]. That platform contained three modular actuation boxes, each with two actuated linear DoFs that connected to Bowden cables. The actuation box translated the rotational motion of a brushless motor (Maxon, Fall River, MA) into linear motion of an aluminum carriage that connected to the inner cable of the Bowden cable. All of the electrical components necessary for programming of the actuators (desktop PC, motor controllers, and power supply) were included in the platform. The platform was designed to meet requirements based on the biomechanical data of healthy people during walking. In particular, the linear actuators move at a speed of 1.5m/s, enabling the ankle to rotate at a speed of 2.8 rad/s – which is the average ankle speed during walking. Ultimately, the actuators can exert forces up to 250N during walking at 1.25m/s, with sufficient control bandwidth for applying assistance to healthy individuals [22]. We previously demonstrated that biologically realistic torques could be provided either separately or simultaneously at the ankle and hip joints during walking. In this work, we extended the capability of this platform such that it was capable of simulating a body-worn exosuit actuation system during both treadmill and overground walking (see the *mobile off-board actuation unit* in Fig 1). The cart-like outer frame has four casters underneath and the handle at the back, therefore clinicians can follow the patient with the actuation unit to facilitate overground walking with exosuit assistance. It was designed to have a low height and a non-reflective surface so as not to disrupt motion capture data collection. The mobile off-board actuation unit provides all the necessary components to control the exosuit in its compact mobile frame. It contains two actuator boxes that are directly adapted from [25] and allow up to 4 DoF actuation, a mini fanless desktop (Advantech America, Milpitas, CA), motor controller (Copley controls, Canton, MA), power supplies with different voltages (5V, 15V, and 48V), and a monitor to display input and output data. To facilitate intermittent walking during overground experiments, a mode selector button was added to the handle to switch controller between walking and standby mode. An emergency stop button was also added to the handle in order to quickly and safely disable tension in the suit when required. Since the actuator box satisfies the biological requirements for a healthy subject's walking, and patients after stroke generally walk at a slower speed, we expect that the actuator unit is sufficient for this patient population.

To our knowledge, this is the first demonstration of an off-board actuation unit for use in overground walking with wearable robots. Although the actuation unit is not suitable for use in a community setting, it can enable feasibility studies with poststroke patients walking on the treadmill and overground. This will guide the development of stroke-specific actuator requirements that will be used for the development of a body-worn actuator.

IV. Sensor and Controller Implementation

The stroke-specific exosuit we present here is equipped with a variety of sensors enabling a closed loop control system to deliver well-timed assistance to each DoF. The control system was designed to be robust to the highly variable gait patterns found in stroke patients. In addition to control, sensor data can be used to calculate gait temporal variables (e.g. step time) in real-time to provide quantitative feedback for clinicians during walking experiments.

A. *Sensor implementation*

Two load cells (Futek, Irvine, CA) were integrated into the distal ends of the PF suit module and of the DF suit module, where Bowden cable outer sheaths attach to the textiles. This configuration allowed measurement of the interaction force between the exosuit and the wearer. A gyroscope (SparkFun, Niwot, CO) was attached on the lateral part of each shoe to measure foot rotational velocities in the sagittal plane. Each actuator box inside the actuation unit was equipped with an incremental encoder (Maxon, Fall River, MA), a linear potentiometer (P3 America, Inc, San Diego, CA) to measure the displacement of the actuation cable, and a load cell (Futek, Irvine, CA) to measure the force at the proximal end of the Bowden cable. A data acquisition card and two DAQ connector blocks (National Instruments, Austin, TX) in the actuation unit collected all the sensor signals. The system was designed to be reconfigurable and so the number and type of sensors can easily be modified for future development of exosuit embodiments and walking experiments.

B. *Real-time gait event detection*

Control algorithms for walking assistance often rely on gait event data such as toe off (TO) and heel strike (HS) to generate assistance profiles [26]. However, the significant

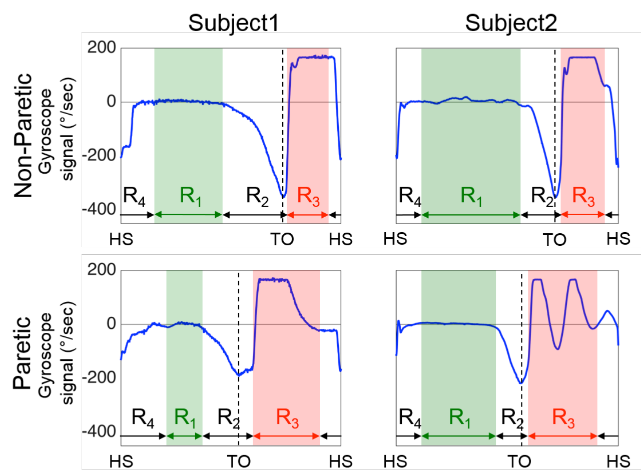

Figure 3 Gyroscope signals segmented by consecutive heel strikes (HS) from both legs for two different subjects. A comparison of the gyroscope signals to gait events segmented by kinematics data validated that the two negative peaks corresponded to HS and TO respectively. In addition, the green region of the gyroscope signal (R1), corresponding to foot flat during stance, was consistently at a value of zero for legs across subjects, while the illustrated red regions (R3), corresponding to early and mid-swing, were not consistent. The gait event detection algorithm used identifies R1 based on definitive conditions in equations (1) and (2). The algorithm searches for TO and HS during R2 and R4 respectively. It deliberately does not rely on the sensor signals in R3 since the sensor signals during the swing phase were inconsistent as shown. In addition, since the absolute magnitudes of peak signals were different in the non-paretic side and the paretic side, adaptive peak detection threshold is implemented in the algorithm in order to address the difference.

step-to-step variability of poststroke gait poses a challenge in implementing conventional gait event detection algorithms that heavily rely on gait periodicity [27]. To address these difficulties, we designed a gait event detection algorithm that uses gyroscope sensor signals from both feet and accounts for inconsistencies in the gait pattern and differences in sensor signals between the paretic and non-paretic limbs and from subject-to-subject. The algorithm can detect TO and HS from both sides independently, and divide the gait cycle into four phases (weight acceptance, paretic limb support, pre swing, and swing). Additionally, it enables evaluating temporal variables such as temporal symmetry between the paretic and non-paretic legs. Previous work [28] showed that the profile of the foot velocity in the sagittal plane as measured by the gyroscopes has two negative peaks in each gait cycle, which correspond to TO and HS. Figure 3 shows an example of measured foot velocity on both the paretic and the non-paretic feet for two different subjects. It can be seen that there is significant variability both in the paretic velocity profiles across different subjects, and between the paretic and non-paretic profiles. The algorithm is adaptive, thus it can be applied to both the paretic and the non-paretic sides without changing parameter settings. The algorithm divides each gait cycle into four regions (R1-R4) in real time, and searches the peaks only in specific regions (R2 and R4) as described:

R1: Foot flat during stance, with the foot velocity signal being close to zero. Mean (μ_{100ms}) and variance (σ_{100ms}^2) of the signal are computed over a sliding window of 100ms. This region starts when μ_{100ms} and σ_{100ms}^2 satisfies (1) and (2), and terminates when either (1) or (2) does not hold. No event is detected during R1.

$$|\mu_{100ms}| < 10 \tag{1}$$

$$\sigma_{100ms}^2 < 100 \tag{2}$$

R2: Pre-swing, beginning with the termination of R1 and ending with the detection of a negative peak that corresponds to TO. The peak is detected within this region in real-time based on an adaptive threshold method (see next paragraph). The time when the peak is detected is recorded (t_{TO}) and used in (3).

R3: Early and mid-swing, beginning with the termination of R2. No event is detected during this region as experimental data showed significant variation in sensor signals across subjects due to foot instability during swing phase. R3 terminates after the calculated duration based on an average of the last three swing time measurements (3). N in (3) is an index of gait cycle (i.e. $T_{swing}(N)$ is the swing time of Nth gait cycle). The same notation is used in the following equations.

$$\begin{aligned} T_{s3}(N+1) &= 0.8 \cdot \frac{1}{3} \cdot \sum_{i=N-2}^{N} T_{swing}(i) \\ &= 0.8 \cdot \frac{1}{3} \cdot \sum_{i=N-2}^{N} [t_{HS}(i) - t_{TO}(i)] \end{aligned} \tag{3}$$

R4: Late swing and early stance, beginning with the termination of R3. The negative peak corresponding to HS is detected in this region similarly to R2. The time when the peak is detected is recorded (t_{HS}) for use in (3), (5), and (6). R4 ends when the next R1 starts.

An adaptive threshold-based detection algorithm, similar to [29], was implemented to allow the robust detection of the TO and HS negative peaks. The algorithm detects the TO (or HS) peak when the sensor signal passes the threshold twice (from above to below, and back to above). Two different thresholds were defined for each the paretic and nonparetic limbs, one for detecting TO ($thres_{TO}$) and one for HS ($thres_{HS}$). Each threshold is computed based on the magnitude of the last three velocity peaks as described in (4). The adaptive thresholds allow the algorithm to be robust to the high variability of the impaired gait as well as to the differences between the paretic and non-paretic foot velocity patterns.

$$|thres_{TO/HS}(N+1)| = 0.8 \cdot \frac{1}{3} \cdot \sum_{i=N-2}^{N} |V_{peak_{TO/HS}}(i)| \tag{4}$$

As described in Section V, we demonstrated that the algorithm can accurately estimate the timing of the TO and HS events. In particular, the paretic stride time (the time period between consecutive HSs from the paretic side) calculated based on this algorithm was very close to that calculated based on motion capture data with an average error of less than 20ms, and the non-paretic stride time was also accurate with an average error of less than 10ms. Throughout the complete human subject testing sessions described, the algorithm detected 100% of HS and TO events from both lower limbs when the exosuit was active. However, the event timing errors slightly increased at faster speeds (i.e. the error calculated from the data of a subject walking at 0.5m/s is larger than that with the same subject walking at 0.3m/s).

C. *Feedback controller and adjustable trajectory generator*

Previous exosuits from our group generate assistive forces by contracting Bowden cables with a change in the position of a linear actuator as commanded by a position controller [18]–[20]. The position command results in an assistive force to the wearer due to the inherent human-suit series stiffness. Given the variability in poststroke gait, it is desirable to have the capability of delivering an individualized assistive profile to each patient. In response to this need, we developed a control strategy that enables the timing, magnitude and profile of the actuator position command trajectory (i.e. cable-pull command trajectory) to be adjustable for each patient, as well as within each session among different walking experiments. The resulting control system was implemented with a Matlab-based user interface (Matlab, Mathworks, MA) that enabled the cable-pull command trajectory to be adjusted manually in real-time. The patient-specific command trajectory is defined as a function of the gait cycle f(*GC*) (Fig 4a). In real time, clinicians can adjust key parameters of the command trajectory, such as the start timing of cable pull and release, pull and release speed, and maximum cable pull. The

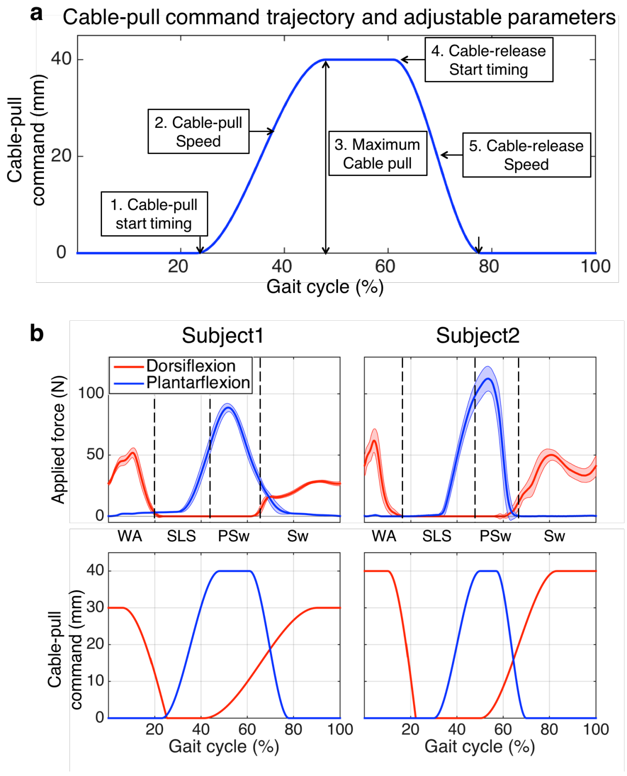

Figure 4 (a) PF cable-pull command trajectory and manually adjustable parameters (1-5). Clinician can change the start timing of cable pull and release, the pull and release speed, and the maximum cable pull. (b) PF and DF Cable-pull command trajectories (bottom) and corresponding applied forces for two different subjects (top). The gait sub-phases shown on force plots (dotted lines in Fig 4b) are identified using motion capture data. The four sub-phase indices under the applied force plots indicate weight acceptance (WA), single limb support (SLS), pre-swing (PSw), and swing (Sw). Based on the force plots, it is clear that different cable-pull commands corresponds to well-timed assistive forces that DF assistance is effective during swing and weight acceptance and PF assistance is effective during mid and late stance. In addition, since Subject1 is 50% lighter than Subject2 (table 1), less assistance is delivered to Subject1.

command trajectory is then scaled by estimated stride time and applied to the controller at the paretic HS of each gait cycle. Equations (5)-(7) give the details on the calculation of the estimated stride time (T_{stride_est}), the current percentage of the gait cycle ($GC_{current}$), and the cable-pull command ($P_{desired}$) at each time. T_{stride_est} is calculated as an average of the last three stride time measurements (5), and $GC_{current}$ is estimated by dividing the time elapsed after the last HS by $T_{stride\_est.}$ (6). Finally, the cable-pull command at current time ($P_{desired}(t_{current})$) is calculated by substituting the current gait cycle percentage to the command trajectory ($f(GC)$) adjusted by a clinician (7).

$$T_{stride_{est}}(N+1) = \frac{1}{3}\cdot\sum_{i=N-2}^{N} T_{stride}(i) = \frac{1}{3}\cdot\sum_{i=N-2}^{N}[t_{HS}(i) - t_{HS}(i-1)] \tag{5}$$

$$GC_{current} = \frac{t_{current} - t_{HS}(N)}{T_{stride_{est}}(N+1)}\cdot 100(\%) \tag{6}$$

$$P_{desired}(t_{current}) = f(GC_{current}) \tag{7}$$

A closed loop PID feedback controller was programmed in the xPC Target environment (Matlab, Mathworks, MA) with a closed-loop frequency of 1kHz so that the actuator could accurately follow the prescribed command trajectory. The feedback controller and adjustable trajectory generator described here were successfully used in the preliminary testing presented in Section V. The comparison between assistance force timing and gait phase segmented with motion capture data validated that we could deliberately control the timing of the provision of assistive force (Fig 4).

V. Human Subject Testing

A. *Experimental protocol*

Three participants with chronic stroke with broad age and different poststroke profiles, all of whom were community ambulators with the use of their regular assistive devices, were recruited for this preliminary study (Table 1). The study was approved by the Harvard Medical School Institutional Review Board. Participant inclusion criteria included being between 25 and 75 years old, at least 6 months poststroke, ability to walk independently for at least 6 minutes continuously, and sufficient passive range of motion at the paretic ankle joint to reach the neutral position with the knee extended. Participants were excluded if they were unable to communicate with research team members, if they had serious musculoskeletal, cardiovascular, or neurological comorbidities, if they had fallen more than twice within the previous month, or if they were currently participating in a physical rehabilitation program. Consent was obtained from the participants and a licensed physical therapist participated in every session. Data were collected at a baseline session (Baseline) and at a session where the suit provided assistance (Active). During Baseline testing, subjects walked on an instrumented treadmill (FIT Bertec, Columbus, OH) at their self-selected treadmill speed (Table 1), and full body motion capture data were collected through a 9-camera Vicon optical motion analysis system (Oxford Metrics, Oxford, UK) with 70 body markers. Between three and five acclimatization sessions with the exosuit followed the baseline session so that the patient could become familiar with the exosuit and the physical therapist could tune the actuator command trajectories. This was done through visual inspection by at least one expert in observational gait analysis of how the patient's gait changed in response to different types of assistance and through the interpretation of a select number of temporal gait variables that were exported from the exosuit and displayed on the interface. After the training sessions, the assistive trajectories were held constant for the remaining sessions.

B. *Data analysis*

Motion capture and ground reaction force (GRF) data were analyzed using Vicon Nexus and Visual 3D (C-motion Inc., Rockville, MD, USA) software. Raw data were filtered (fourth order low-pass Butterworth filter, 12Hz cut-off frequency) prior to processing, and spatial gait variables such as step length and foot trajectories were calculated based on motion capture data. The spatial gait variables were

TABLE I. SUBJECT DESCRIPTION

	Age (years)	Time post-stroke (years)	Gender	Height (m)	Weight (Kg)	Self-selected speed[1] (m/s)
Subject1	29	6	Female	1.62	49.9	0.3
Subject2	63	2	Male	1.83	98.4	0.5
Subject3	60	12	Male	1.83	95.3	0.45

[1]Self-selected speed on the treadmill without orthotic devices

segmented and normalized to 0-100% of the gait cycle, as defined by consecutive HSs. Gait events (HS and TO) were identified from GRFs and marker data. Spatio-temporal gait data were analyzed only for steady-state walking in the final minute of each condition, and the average across all measured strides in each final minute was used for analysis. Gait symmetry and progression metrics were calculated based on motion capture and GRF data (Table 2) since these are key parameters considered during poststroke gait rehabilitation [30]. Propulsive impulse was calculated as the integral of positive anterior-posterior GRF (AP GRF) during stance [31], and paretic propulsion was defined as the proportion of the total propulsive impulse produced by paretic side [32].

C. *Result and discussion*

Symmetry index (SI) [33] was defined as:

$$SI = (X_{np}-X_p) / [0.5\cdot(X_{np} + X_p)]\cdot 100\%$$

X_{np} and X_p were gait parameters from the non-paretic and paretic sides, respectively. We observed an improvement in step time symmetry index (i.e. SI when X_{np} and X_p are step time) with an average of 6.26% (Table 2). Stance time symmetry also improved by an average 3.52%. Moreover, stride time decreased by 11.43%. Fig 5b presents an example stance and stride time for a representative subject. In terms of spatial gait symmetry, two subjects had less significant improvement in their step length symmetry (Table 2). However, the abnormal lateral movement of the foot during swing phase was significantly reduced when a subject with circumduction gait (Subject 1) was assisted by the exosuit (Fig 5a). In terms of propulsion, the activation of the suit facilitated all subjects to have more propulsion symmetry. We observed an average 7.15% increase in propulsion symmetry. These initial preliminary results show a promising positive effect of the exosuit on temporal (step time and stance time) symmetry and paretic limb propulsion. In addition, because spatial symmetry (step length symmetry) also improved - albeit to a lesser extent - and excess lateral motion of the foot resulting from circumduction was reduced, these results imply an effect of the exosuit on compensatory gait patterns. These results suggest that the

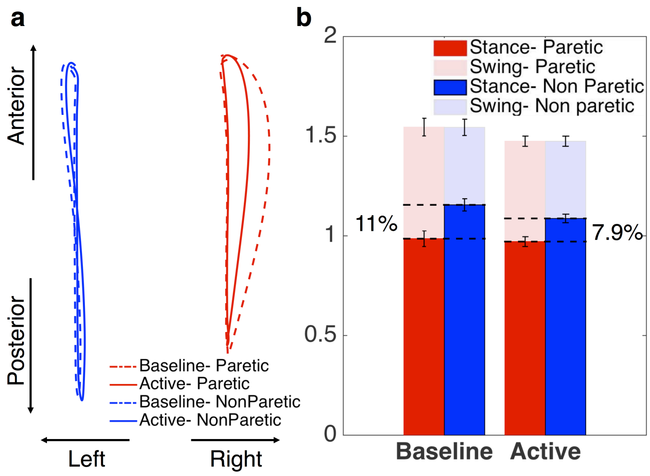

Figure 5 Examples of spatiotemporal symmetry improvement. (a) Foot trajectory of a subject with circumductory gait pattern (Subject 1) seen from above. The subject significantly reduced excess lateral foot motion when suit was active. (b) At the same time, stance time symmetry increased and stride time decreased.

exosuit may be able to promote a more efficient walking pattern in individuals poststroke. The decrease in stride length for all subjects may imply that subjects were unable to take long strides due to a restrictive nature of the suit. On the other hand, it could suggest that subjects increased their cadence with a desire to walk faster over a treadmill-constrained speed.

VI. CONCLUSION AND FUTURE WORK

In this paper, we presented the initial development and evaluation of a soft exosuit for poststroke gait assistance. New exosuit textile modules, a mobile off-board actuation unit, and control algorithms were described that enable human subjects testing in patients with stroke both on the treadmill and overground. Preliminary experiments were conducted on three patients with chronic stroke to validate the system robustness and evaluate the effect on walking biomechanics. Initial results demonstrated that a soft exosuit could improve patients' gait symmetry and paretic limb progression. This work thus presents proof-of-concept and motivates further effort in this area. Future work will be directed towards optimizing the design of the soft exosuit. In particular, we plan to optimize the design of the textile components to enable the application of higher forces and improve the controller to provide greater tuning of the assistance to the wearer. We will also focus on using the data from these initial studies to guide the development of a body-worn actuation system and a graphical user interface to

TABLE II. GAIT PARAMETER COMPARISION

	Step time SI (%) [a]			Step length SI (%) [a]			Paretic propulsion (%) [b]			Stride time (sec)		
	Baseline	*Active*	*Δ*	*Baseline*	*Active*	*Δ*	*Baseline*	*Active*	*Δ*	*Baseline*	*Active*	*Δ*
Subject1	24.91	15.71	**9.20**	-50.43	-48.81	**1.62**	10.27	10.54	**0.27**	1.55	1.47	**-0.08**
Subject2	9.68	3.26	**6.42**	21.36	25.60	**-4.24**	37.04	42.77	**5.73**	1.78	1.43	**-0.35**
Subject3	19.09	15.94	**3.15**	-13.90	-10.26	**3.64**	5.77	21.23	**15.46**	1.90	1.72	**-0.18**

[a] 0% SI is perfect symmetry [b] 50% paretic propulsion is perfect symmetry

facilitate clinicians' use of the exosuit. Finally, we plan more extensive human subject testing of the exosuit to demonstrate its potential across a larger number of patients with stroke.

ACKNOWLEDGMENT

The authors would like to thank Hani Sallum, Tiffany Wong, Gabe Greeley, Diana Wagner, Mona Qureshi, Kelly Cakert, and Alan T. Asbeck for their assistance with various aspects of the work. This work was supported by the by the Defense Advanced Research Projects Agency (DARPA) Warrior Web Program (Contract No. W911NF-14-C-0051-P00003), the National Science Foundation (Grant CNS-1446464) the Harvard University Star Family Challenge, the Wyss Institute for Biologically Inspired Engineering, and the School of Engineering and Applied Sciences at Harvard University.

REFERENCES

[1] D. Mozaffarian, E. J. Benjamin, A. S. Go, D. K. Arnett, M. J. Blaha, M. Cushman, S. de Ferranti, J.-P. Despres, H. J. Fullerton, V. J. Howard et al., "Heart disease and stroke statistics--2015 update: a report from the American Heart Association.," *Circulation*, vol. 131, no. 4, pp. e29–322, Jan. 2015.

[2] G. E. Gresham, T. E. Fitzpatrick, P. A. Wolf, P. M. McNamara, W. B. Kannel, and T. R. Dawber, "Residual disability in survivors of stroke--the Framingham study.," *N. Engl. J. Med.*, vol. 293, pp. 954–956, 1975.

[3] I. A. De Quervain, S. R. Simon, S. Leurgans, W. S. Pease, and D. McAllister, "Gait pattern in the early recovery period after stroke.," *J. Bone Joint Surg. Am.*, vol. 78, pp. 1506–1514, 1996.

[4] S. Olney and C. Richards, "Hemiparetic gait following stroke. Part I: Characteristics," *Gait Posture*, vol. 4, pp. 136–148, 1996.

[5] N. E. Mayo, S. Wood-Dauphinee, S. Ahmed, C. Gordon, J. Higgins, S. McEwen, and N. Salbach, "Disablement following stroke," *Disabil Rehabil*, vol. 21, no. 5–6, pp. 258–268, 1999.

[6] R. Dickstein, "Rehabilitation of gait speed after stroke: a critical review of intervention approaches.," *Neurorehabil. Neural Repair*, vol. 22, no. 6, pp. 649–660, 2008.

[7] J. E. Sullivan, L. E. Espe, A. M. Kelly, L. E. Veilbig, and M. J. Kwasny, "Feasibility and outcomes of a community-based, pedometer-monitored walking program in chronic stroke: a pilot study.," *Top. Stroke Rehabil.*, vol. 21, no. 2, pp. 101–110, 2014.

[8] I. Díaz, J. J. Gil, and E. Sánchez, "Lower-Limb Robotic Rehabilitation: Literature Review and Challenges," *J. Robot.*, vol. 2011, no. i, pp. 1–11, 2011.

[9] A. Pennycott, D. Wyss, H. Vallery, V. Klamroth-Marganska, and R. Riener, "Towards more effective robotic gait training for stroke rehabilitation: a review," *J. Neuroeng. Rehabil.*, vol. 9, p. 65, Jan. 2012.

[10] A. M. Dollar and H. Herr, "Lower extremity exoskeletons and active orthoses: challenges and state-of-the-art," *Robot. IEEE Trans.*, vol. 24, no. 1, pp. 144–158, 2008.

[11] S. Jezernik, G. Colombo, T. Keller, H. Frueh, and M. Morari, "Robotic orthosis lokomat: a rehabilitation and research tool.," *Neuromodulation*, vol. 6, no. 2, pp. 108–15, Apr. 2003.

[12] D. J. Reinkensmeyer, D. Aoyagi, J. L. Emken, J. a. Galvez, W. Ichinose, G. Kerdanyan, S. Maneekobkunwong, K. Minakata, J. a. Nessler, R. Weber, R. R. Roy, R. de Leon, J. E. Bobrow, S. J. Harkema, V. R. Edgerton, R. De Leon, J. E. Bobrow, S. J. Harkema, and V. R. Edgerton, "Tools for understanding and optimizing robotic gait training," *J. Rehabil. Res. Dev.*, vol. 43, no. 5, p. 657, 2006.

[13] S. Banala and S. Kim, "Robot Assisted Gait Training With Active Leg Exoskeleton (ALEX)," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 17, no. 1, pp. 2–8, 2009.

[14] A. Esquenazi, M. Talaty, A. Packel, and M. Saulino, "The ReWalk Powered Exoskeleton to Restore Ambulatory Function to Individuals with Thoracic-Level Motor-Complete Spinal Cord Injury," *Am. J. Phys. Med. Rehabil.*, vol. 91, pp. 911–921, 2012.

[15] I. Galiana, F. L. Hammond, R. D. Howe, and M. B. Popovic, "Wearable soft robotic device for post-stroke shoulder rehabilitation: Identifying misalignments," in *IEEE International Conference on Intelligent Robots and Systems*, 2012, pp. 317–322.

[16] Y.-L. Park, B.-R. Chen, N. O. Pérez-Arancibia, D. Young, L. Stirling, R. J. Wood, E. C. Goldfield, and R. Nagpal, "Design and control of a bio-inspired soft wearable robotic device for ankle-foot rehabilitation.," *Bioinspir. Biomim.*, vol. 9, no. 1, p. 016007, 2014.

[17] M. Wehner, B. Quinlivan, P. M. Aubin, E. Martinez-Villalpando, M. Baumann, L. Stirling, K. Holt, R. Wood, and C. Walsh, "A lightweight soft exosuit for gait assistance," *Proc. - IEEE Int. Conf. Robot. Autom.*, pp. 3362–3369, 2013.

[18] A. T. Asbeck, R. J. Dyer, A. F. Larusson, and C. J. Walsh, "Biologically-inspired soft exosuit," in *IEEE International Conference on Rehabilitation Robotics*, 2013.

[19] A. T. Asbeck, S. M. M. De Rossi, I. Galiana, Y. Ding, and C. J. Walsh, "Stronger, Smarter, Softer: Next-Generation Wearable Robots," *Robotics & Automation Magazine, IEEE*, vol. 21, no. 4. pp. 22–33, 2014.

[20] A. T. Asbeck, S. M. M. De Rossi, K. G. Holt, and C. J. Walsh, "A biologically inspired soft exosuit for walking assistance," *Int. J. Rob. Res.*, pp. 1–19, 2015.

[21] A. T. Asbeck, K. Schmidt, I. Galiana, and C. J. Walsh, "Multi-joint Soft Exosuit for Gait Assistance," in *Robotics and Automation (ICRA), 2015 IEEE International Conference on*, 2015.

[22] Y. Ding, I. Galiana, A. Asbeck, B. Quinlivan, S. M. M. De Rossi, and C. Walsh, "Multi-joint actuation platform for lower extremity soft exosuits," in *Robotics and Automation (ICRA), 2014 IEEE International Conference on*, 2014, pp. 1327–1334.

[23] P. Polygerinos, Z. Wang, K. C. Galloway, R. J. Wood, and C. J. Walsh, "Soft robotic glove for combined assistance and at-home rehabilitation," *Rob. Auton. Syst.*, Sep. 2014.

[24] P. Polygerinos, K. C. Galloway, E. Savage, M. Herman, K. O'Donnell, and C. J. Walsh, "Soft Robotic Glove for Hand Rehabilitation and Task Specific Training," in *Robotics and Automation (ICRA), 2015 IEEE International Confernce on*, 2015.

[25] Y. Ding, I. Galiana, A. T. Asbeck, S. M. M. De Rossi, J. Bae, T. R. T. Santos, V. L. Araujo, S. Lee, K. Holt, and C. J. Walsh, "Biomechanical and Physiological Evaluation of Multi-joint Assistance with Soft Exosuits," *IEEE Trans. Neural Syst. Rehabil. Eng. under review*. .

[26] M. R. Tucker, J. Olivier, A. Pagel, H. Bleuler, M. Bouri, O. Lambercy, J. D. R. Millan, R. Riener, H. Vallery, and R. Gassert, "Control strategies for active lower extremity prosthetics and orthotics: a review.," *J. Neuroeng. Rehabil.*, vol. 12, no. 1, p. 1, 2015.

[27] J. Rueterbories, E. G. Spaich, B. Larsen, and O. K. Andersen, "Methods for gait event detection and analysis in ambulatory systems.," *Med. Eng. Phys.*, vol. 32, no. 6, pp. 545–52, Jul. 2010.

[28] H. Lau and K. Tong, "The reliability of using accelerometer and gyroscope for gait event identification on persons with dropped foot," *Gait Posture*, vol. 27, pp. 248–257, 2008.

[29] B. R. Greene, D. McGrath, R. O'Neill, K. J. O'Donovan, A. Burns, and B. Caulfield, "An adaptive gyroscope-based algorithm for temporal gait analysis.," *Med. Biol. Eng. Comput.*, vol. 48, no. 12, pp. 1251–1260, Dec. 2010.

[30] K. K. Patterson, I. Parafianowicz, C. J. Danells, V. Closson, M. C. Verrier, W. R. Staines, S. E. Black, and W. E. McIlroy, "Gait asymmetry in community-ambulating stroke survivors.," *Arch. Phys. Med. Rehabil.*, vol. 89, no. 2, pp. 304–310, Feb. 2008.

[31] M. G. M. Bowden, C. K. C. Balasubramanian, R. R. Neptune, and S. a. Kautz, "Anterior-posterior ground reaction forces as a measure of paretic leg contribution in hemiparetic walking," *Stroke*, vol. 37, no. 3, pp. 872–876, Mar. 2006.

[32] L. N. Awad, D. S. Reisman, T. M. Kesar, and S. A. Binder-Macleod, "Targeting paretic propulsion to improve poststroke walking function: a preliminary study.," *Arch. Phys. Med. Rehabil.*, vol. 95, no. 5, pp. 840–848, May 2014.

[33] S. Nadeau, "Understanding Spatial and Temporal Gait Asymmetries in Individuals Post Stroke," *Int. J. Phys. Med. Rehabil.*, vol. 02, 201.